OPTICAL AND PHYSICAL HEART STABILIZATION FOR CARDIAC SURGERY

By

Terence Gilhuly

B. A. Sc. (Systems Design Engineering) University of Waterloo, 1995

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF
THE REQUIREMENTS FOR THE DEGREE OF
MASTER OF APPLIED SCIENCE

in
THE FACULTY OF GRADUATE STUDIES
ELECTRICAL ENGINEERING

We accept this thesis as conforming
to the required standard

THE UNIVERSITY OF BRITISH COLUMBIA
April 1998
© Terence Gilhuly, 1998
In presenting this thesis in partial fulfilment of the requirements for an advanced degree at the University of British Columbia, I agree that the Library shall make it freely available for reference and study. I further agree that permission for extensive copying of this thesis for scholarly purposes may be granted by the head of my department or by his or her representatives. It is understood that copying or publication of this thesis for financial gain shall not be allowed without my written permission.

Electrical Engineering
The University of British Columbia
2075 Wesbrook Place
Vancouver, Canada
V6T 1W5

Date: April 29, 1998
Abstract

Cardiopulmonary bypass (CPB) machines impose a great deal of harm to the patient during cardiac surgeries demanding their use. The damage they cause ranges from inflammation of the immune response and other chemical imbalances to mechanical problems such as the destruction of platelets and red blood cells by the pump's rollers as the blood cycles through the pump. If this damage is eliminated, patient recovery time both in and out of the hospital could be reduced at great savings to patients, taxpayers and hospitals alike.

This thesis proposes, details and evaluates surgical devices and techniques that would permit operation on a beating heart to occur without the use of a CPB machine. Specifically designed for Coronary Artery Bypass Grafting (CABG) surgery, but applicable to other surgeries as well, these devices and techniques concentrate on stabilizing, or making apparent the stabilization, of anastomosis sites on the heart so that surgery may proceed unimpeded.

Optical strobing devices are presented first. Theoretically, by strobing the heart according to the patient's EKG signal, the surgeon will only see the heart at a certain time in the cardiac cycle making the heart seem frozen in that position. Thus, with this frozen image, the surgeon will be able to suture to the heart as if it were still. The devices presented here trigger off of a patient's EKG R-wave, producing a pulse of light that is both adjustable in delay to start and length of time on. The adjustments allow the timing of the light so that it can be turned on at any point in the cardiac cycle and left on for as much time as the surgeon desires. Typically these are set so that the strobing occurs at the end of the diastolic period when the heart is nearly filled with blood, for a length of time that is short enough to maintain a frozen image of the heart, yet long enough to allow the surgeon to make their stitch.

A physical method of stabilizing the immediate section of the heart where the graft would take place is also put forward here. This stabilizing device consists of a rigid platform that fixates to the heart with vacuum pressure and is then rigidly connected to the patient at their retractor with a surgical arm newly developed for this purpose and also presented in this document. The stabilization device is a "C"-shaped ring. Its vacuum is provided in a continuous field to permit greater hold at lower pressures with a smaller sized device than could occur while using discrete suction cups. The arm is a seven link, nine DOF redundant serial linkage. It features a lightweight, quick to activate, highly dextrous, unobtrusive
design relying on pneumatic pressure for fixation. It mounts to the retractor to couple its motion to the patients should they cough or otherwise move during the surgery. This mounting eliminates the danger of injury from this source and is unobtrusive relative to floor or table-mount designs. A number of other novel stabilization devices and surgical arms are presented, as well.

In designing the stabilizing device and surgical arm, required end effector force and torque estimates were determined by scaling force and torque measurements taken from pigs to representative force and torque values for large humans.
Table of Contents

Abstract ii

List of Tables vii

List of Figures viii

List of Acronyms x

Acknowledgements xii

1 Introduction 1

1.1 Coronary Artery Bypass Procedures 2

1.1.1 The Cardio-Pulmonary Bypass Pump 3

1.2 Alternative Methods to Standard CABG 3

1.2.1 Non-Sternotomy Surgeries 3

1.2.2 Off Pump Surgeries and Cardiac Stabilization 4

1.2.3 Prior Art in Optical Stabilization 6

1.3 Objectives of this Research 7

2 Optical Stabilization of the Heart 9

2.1 Circuitry 11

2.1.1 R-Wave Detection 11

2.1.2 Strobe Driving Circuitry 12

2.2 Lighting Methods 14

2.2.1 The LED Strobes 15

2.2.2 The Incandescent Light Strobe 16

2.2.3 The Flash Tube Strobes 17

3 Optical Stabilization Testing 21

3.1 The Testing Apparatus 21
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.1.1</td>
<td>Control</td>
<td>23</td>
</tr>
<tr>
<td>3.1.2</td>
<td>The Servoing Signals</td>
<td>24</td>
</tr>
<tr>
<td>3.1.3</td>
<td>Strobe Triggering</td>
<td>26</td>
</tr>
<tr>
<td>3.2</td>
<td>Suturing Tests</td>
<td>26</td>
</tr>
<tr>
<td>3.2.1</td>
<td>Test Evaluation</td>
<td>27</td>
</tr>
<tr>
<td>3.3</td>
<td>Results</td>
<td>28</td>
</tr>
<tr>
<td>3.4</td>
<td>Recommendations</td>
<td>31</td>
</tr>
<tr>
<td>4</td>
<td>Physical Stabilization of the Heart</td>
<td>38</td>
</tr>
<tr>
<td>4.1</td>
<td>Stabilizer Design Requirements</td>
<td>38</td>
</tr>
<tr>
<td>4.1.1</td>
<td>Maintaining a Stable Suturing Area</td>
<td>40</td>
</tr>
<tr>
<td>4.2</td>
<td>Cardiac Stabilizer Design Alternatives</td>
<td>44</td>
</tr>
<tr>
<td>4.2.1</td>
<td>Proof of Concept</td>
<td>47</td>
</tr>
<tr>
<td>4.2.2</td>
<td>Further Experimentation</td>
<td>47</td>
</tr>
<tr>
<td>4.2.3</td>
<td>The Final Prototype Stage</td>
<td>48</td>
</tr>
<tr>
<td>4.2.4</td>
<td>Testing the Stabilizer</td>
<td>49</td>
</tr>
<tr>
<td>4.3</td>
<td>Future Designs</td>
<td>50</td>
</tr>
<tr>
<td>4.3.1</td>
<td>Future Stabilizer Shape</td>
<td>50</td>
</tr>
<tr>
<td>4.3.2</td>
<td>Control</td>
<td>52</td>
</tr>
<tr>
<td>5</td>
<td>Surgical Arm Development</td>
<td>53</td>
</tr>
<tr>
<td>5.1</td>
<td>Surgical Arm Design Constraints</td>
<td>53</td>
</tr>
<tr>
<td>5.2</td>
<td>Design Alternatives</td>
<td>54</td>
</tr>
<tr>
<td>5.2.1</td>
<td>The Rod and Snug Experimental Arm</td>
<td>57</td>
</tr>
<tr>
<td>5.2.2</td>
<td>The Flexible Arm</td>
<td>58</td>
</tr>
<tr>
<td>5.2.3</td>
<td>The Electromagnetic Arm</td>
<td>59</td>
</tr>
<tr>
<td>5.2.4</td>
<td>The Pneumatic Arm</td>
<td>60</td>
</tr>
<tr>
<td>5.2.5</td>
<td>The Pistonless Pneumatic Arm</td>
<td>66</td>
</tr>
<tr>
<td>5.3</td>
<td>Future Surgical Arm Development</td>
<td>70</td>
</tr>
<tr>
<td>6</td>
<td>Conclusions</td>
<td>72</td>
</tr>
<tr>
<td>6.1</td>
<td>Contributions</td>
<td>72</td>
</tr>
</tbody>
</table>
List of Tables

3.1 Group means and standard deviations of subject perceptions during the strobing tests (scale of 1 to 10 questionnaire replies) .................................................. 28
3.2 Group means and standard deviations for subject performance during the strobing tests ................................................................. 29
3.3 Significance and power testing of the unpaired data ................................................................. 30
3.4 Paired testing means and standard deviations for subject performance during the tests ................................................................. 30
3.5 Significance and power testing results of the paired data ................................................................. 31

4.6 Maximum force (N) and torque (Nm) measurements from the first pig, scaled to large human sizes ........................................................................ 46
4.7 Maximum force (N) and torque (Nm) measurements from the second pig, scaled to large human sizes ........................................................................ 46
4.8 Comparison of dimensions and performance for the stabilizer models (dimensions are in mm for lengths, mm$^2$ for areas and N for force) ................................................................. 47

5.9 Qualitative comparison of the arm design alternatives ........................................................................ 57
5.10 Typical heart dimensions (from [32]) ......................................................................................... 61
5.11 Maximum and minimum braking torques (in Nm) calculated for the pneumatic arm ................................................................. 64
5.12 Maximum and minimum braking torques (in Nm) calculated for the pistonless pneumatic arm ......................................................................................... 70

A.13 Heart chamber displacements (from [32]) ......................................................................................... 77
List of Figures

2.1 Heart cycle EKG, pressure, volume, heart sound and recommended times for strobing relationships (modified from [3]) .................................................. 10
2.2 General implementation of the optical stabilization method ............................................. 11
2.3 EKG R-wave detection circuit .................................................................................. 13
2.4 Block diagram for EKG R-wave detection circuit ....................................................... 14
2.5 LED triggering circuit ............................................................................................... 15
2.6 Strobe triggering circuit ............................................................................................ 18
3.7 Test apparatus setup: strobe (top), EKG filter, power supplies, signal generating computer (bottom, left to right at rear), rotating platform (foreground) .................. 22
3.8 Strobing test platform overview ................................................................................ 23
3.9 Electronic circuitry for testing platform .................................................................. 33
3.10 Heart waveforms: measured (top graph), and artificial: simple (middle) and complicated (bottom) ................................................................. 34
3.11 The heart tissue position sensing setup .................................................................... 34
3.12 Unmarked target sheet on platform for testing of left handed subject ..................... 35
3.13 Subject performing a visual stabilization test .......................................................... 35
3.14 Subject perceptions from performing the optical stabilization tests ..................... 36
3.15 Subject performance during the strobing tests ......................................................... 37
4.16 Application of the cardiac stabilizer ....................................................................... 39
4.17 The force torque sensor and experimental arm above the pig's thoracic cavity .... 41
4.18 The force torque sensor ......................................................................................... 42
4.19 200 kg human left ventricle forces and torques ...................................................... 44
4.20 Cardiac stabilizer designs (dimensions in mm) ......................................................... 45
4.21 Applying the C-ring (with experimental arm) for grafting ....................................... 49
4.22 Grafted vein held by surgical pliers ....................................................................... 50
List of Acronyms

bpm (Heart) Beats per minute
inHg Inches of mercury (pressure measurement)
mmHg Millimeters of mercury (pressure measurement)
psi Pounds per square inch (pressure measurement)
ADC Analog to Digital Conversion
AV Atrio-ventricular
BSA Base Surface Area
CA Coronary Artery
CAB Coronary Artery Bypass
CABG Coronary Artery Bypass Graft
CHF Congestive Heart Failure
CNC Computer Numerically Controlled
CPB Cardio-Pulmonary Bypass
CV Cardio-Vascular
DISS Diameter Index Safety System
DOF Degree(s) of Freedom
DH Denavit-Hartenberg
ECG, EKG Electrocardiogram
EE End Effector
FET Field Effect Transistor
FDA Federal Drug Administration
FTS Force Torque Sensor
LA Left Atrium
LAD Left Anterior Descending (Coronary Artery)
LIMA Left Internal Mammary Artery
LCD Liquid Crystal Display
LV Left Ventricle
LVAD  Left Ventricular Assistive Device
MIDCAB  Minimally Invasive Direct Coronary Artery Bypass
OR  Operating Room
PCTA  Percutaneous Transluminal Coronary Angioplasty
PID  Proportional Integral Derivative (controller)
PWM  Pulse Width Modulation
RA  Right Atrium
RBC  Red Blood Cell
RCA  Right Coronary Artery
RF  Radio-frequency
RIMA  Right Internal Mammary Artery
RV  Right Ventricle
SA  Sino-Atrial
SF  Safety Factor
TAH  Total Artificial Heart
VAD  Ventricular Assist Device
VOGL  Very Ordinary Graphical Language
Acknowledgements

General thanks are due to all those around me making this planet what it is. Were it not for God, my parents, my family, my wonderful girlfriend, for whom I should do more dishes\(^1\), and my friends, I would not be here as I am. Were it not for my advisors in surgery, academic supervisors, support staff, and technicians and the like, the projects that were constructed would not be as they are. And were it not for the society I live in, the words would not flow as they do and this thesis would not be as it is.

Specifically I would like to thank my advisors Drs Tim Salcudean, Sam Lichtenstein, Kassam Ashe and Peter Lawrence. I would like to thank Don Dawson and Simon Bachman for their machining and mechanical design advice and criticism. Kudos to the rest of the robotics group for their support and necessary distraction.

Last but not least, thanks to the Government of Canada and its setting up of technology funding through IRIS (the Institute for Robotics and Intelligent Systems) which made it possible financially for me to study, research and produce this work.

Thank you all.

\(^1\)She made me say that.
Chapter 1

Introduction

Cardio-vascular disease, and coronary artery disease in particular, is very prevalent and becoming increasingly so due to our aging population. According to the American Heart Association [1], in 1994 in the United States alone there were 501,000 coronary artery bypass grafting (CABG) procedures performed on 318,000 patients. Assuming bypass operation incidence is proportional with population, approximately 36,500 Canadians endured a total of 57,600 CABG procedures. At an average cost of US$44,200, a total of US$22 billion was spent in the United States and about US$2.5 billion was spent in Canada for bypass procedures. This cost quadruples when costs for diagnosis, nursing home services, medications, home health care and productivity loss from morbidity and mortality are taken into account [1].

A great proportion of these costs are due to the use of the cardiopulmonary bypass (CPB) machine (heart-lung machine) and the subsequent negative effects it has on the blood. If this machine could be removed and the surgery performed directly on the heart while it is still beating, costs could be reduced and patient well-being increased (as will be discussed in Section 1.1.1). This thesis explores two separate methods of stabilizing the heart to permit this to occur: one visual and the other physical.

Before these are described, however, details of current bypass surgery will be explained, including the use and impact of the CPB machine. Then, alternative methods for performing CABG will be discussed, along with related prior art in the field of stabilization and surgical arm design. Finally, the objectives of this research will be presented.

To aid understanding of all this, a basic discussion of heart anatomy and physiology can be found in Appendix A.
1.1 Coronary Artery Bypass Procedures

Coronary artery disease involves the calcification of the coronary arteries, which supply blood to the muscular tissue of the heart. This calcification leads to ischemia (narrowing of the arteries) and eventually, to a complete blockage of blood flow to the tissue these arteries serve. Without a supply of fresh blood the heart tissue dies, resulting in a heart attack and potentially death. To allow the patient to survive and resume leading a happy, healthy life, these blockages in the heart need to be either removed or bypassed.

In simple cases where the degree of calcification is small, the arteries can be widened through use of percutaneous transluminal coronary angioplasty (PCTA). This technique involves the passing of a balloon catheter to the location of the blockage in the artery. It is positioned to surround the blockage and then inflated to compress the calcification and open up the artery. This method is simple and effective in the short term, however blockages often form again.

Should the blockages become substantial enough that PCTA is not deemed to be effective, cardiology gives way to cardiac surgery and the blockages are bypassed. In CABG procedures, the heart is revascularized by bypassing all severe stenoses (greater than 50% diameter reduction) in all coronary branches and arterial trunks greater than 1 mm in diameter [2], to provide the tissues behind the ischemic areas of the arteries with blood. Redundant veins or arteries, generally the saphenous vein and the left and right internal mammary arteries (LIMA and RIMA), serve as the bypassing media. They are attached to the coronary arteries distal to the stenosis with either side to side or end to side anastomoses (attachment with the intention of providing fluid flow). For veinous grafts (which involve completely harvested veins) the veins are attached to a blood source, usually the aorta. Redundant arteries already provide fresh blood and are therefore never completely removed and need only be attached to the grafting site.

This operation is complicated by the small size of the blood vessels concerned and the relatively large and quick motions of the heart. The arteries are generally between one and 2.5 mm in diameter and have wall thickness less than 0.8 mm. The veins used are similarly diminutive, having wall thicknesses of approximately 0.5 mm and being 5 mm in diameter. Heart motion can be as much as 4.4 cm (from the measurements of Section 3.1.2) for a large human heart during the contraction of the ventricles, an event requiring less than 0.3 s for a heart beating at 75 bpm [3].
1.1.1 The Cardio-Pulmonary Bypass Pump

Due to the described motion of the heart and size of the target arteries and graftee blood vessels, the heart and lungs are usually stopped so that the surgeon may make the anastomosis on a still target. When the heart is stopped, the function of the heart and lungs are replaced by a cardio-pulmonary bypass (CPB) machine. Blood flow that would normally be to the lungs is instead diverted from the vena cava through a cannula to the CPB machine. There it is warmed, purged of carbon-dioxide, oxygenated and then returned through another cannula into the aorta.

While the CPB machine is very effective in its duties and thereby effective in keeping the patient alive, it can be extremely harmful to the patient as well. Example physiological problems due to prolonged CPB exposure include: hemolysis, anemia, constriction and ischemia of major arteries, red blood cell aggregation, impaired renal function, released renin, and cellular hypoxia [2]. The impact of this on the patient is an extended hospital recovery time from several days to a few weeks at great cost in terms of time and money. Hospitals are similarly burdened. In a study comparing simple (as indicated by the much less than the previously quoted US$44,000 per case average cost) on-pump CABG with minimally invasive (the surgeries were performed through left anterior small thoracotomy as opposed to a sternotomy) cases performed off-pump, for two groups of patients with similar preoperative conditions, length of stay in the hospital was seen to decline from 5.9 ± 2 days for on-pump cases to 2.5 ± 0.8 days for the off-pump cases. Total cost was seen to fall from US$21,260 ± US$5,497 to US$12,885 ± US$1,511 [4].

A more detailed elaboration on these effects and the positive results of not using the pump can be found in Appendix A.2.1.

1.2 Alternative Methods to Standard CABG

Alternative methods of surgeries performed focus on reducing harm to the patient through one or both of two actions: 1) reducing the intrusiveness of the procedure by not using a sternotomy, and 2) reducing the overall damage by eliminating the pump completely from the procedure.

1.2.1 Non-Sternotomy Surgeries

Traditionally CABG operations have involved performing a sternotomy and then spreading the ribs with a chest retractor to gain access to the heart. This involves cutting through the sternum lengthwise.
Chapter 1. Introduction

Minimally invasive direct coronary artery bypass (MIDCAB) surgery reduces the wound size to the patient by operating on the patient through a limited thoracotomy - cutting through one or two ribs above the site of interest and again spreading them apart. However, the pump is still often used as the limited access only offers good exposure to the coronary arteries directly beneath the port. This only allows grafting to the left anterior descending (LAD) coronary artery off-pump and not to locations on the posterior side of the heart. The pump is also used to avoid the danger of complications such as fibrillation occurring. In this instance the surgeon may be helpless to prevent the patient from dying due to their inability to massage the heart or perform other basic resuscitation techniques. For these reasons, these cases need to be chosen very selectively.

Saphenous vein graft harvesting, traditionally done through large incisions in the lower leg, is now done in a minimally invasive fashion as well. Small holes only are required through the skin to fit a laparoscopic camera, surgical tools and piping for inflation with CO₂. The inflation creates a larger workspace to permit better maneuverability for the tools. Guidant [5] is one organization that produces tools for this procedure.

1.2.2 Off Pump Surgeries and Cardiac Stabilization

The main problem with operating on the beating heart is that the surgeon must then stitch to a moving target. To allow this to occur the medical teams have stabilized the heart physically - generally just a small section of it so that fibrillation would not be induced. In the first attempts at coronary bypass surgery off-pump, the surgeons involved simply tethered the heart with thick sutures to the retractor. They did this about the location of the stenose (blockage) to hold its artery and the surrounding area still. There have been many studies ([6], [7], [8], [9], [10], and [11]) showing excellent results in terms of graft patency and patient recovery, however the motion of the heart makes these operations lengthy and difficult to perform.

As an improvement on these initial efforts, mechanical stabilizer devices were developed to provide more accurate stabilization of the heart. The ring idea that started the research of this thesis was developed by Dr. Kassam Ashe in 1990 [16]. He constructed a prototype consisting of a machined ring similar to the O-ring (see Section 4.2.1). However, it was a simpler version than that of this thesis, larger and with poorly placed vacuum port and immobilizing connection.

Other stabilization devices developed near the same time as the ring devices of this thesis include the commercial devices made by Cardio-Thoracic Systems (CTS) Inc. ([12], [13]), Guidant ([5]) and
Medtronic ([14]).

The CTS and Guidant devices are very similar, both relying on a horseshoe shaped mechanism to push down the region of the coronary artery, trapping the heart at this location. The CTS horseshoe is attached to a vertical rod which is screw-tightened to a slide which is screw mounted to a custom made retractor. This appears to give it three degrees of freedom. The Guidant device's horseshoe is fixated to the chest retractor through a flexible cable clamp mechanism screw tightened to the retractor.

In contrast to the CTS and Guidant designs, the stabilizer of this thesis uses vacuum pressure to attach to the heart tissue. This permits it to be applied to any location on the heart. The systems of CTS and Guidant are restricted in this sense, as they can only work upon areas that they can push down upon.

The solution of Medtronic, known as the “Octopus”, relies on vacuum suction applied through discrete suction cups on rods to grip the heart. The rod with the suction cups is screw tightened to a cable holder, which is then screw fastened to a rod mounted to the operating room table.

In contrast to the Octopus, this thesis' stabilizer applies its vacuum pressure over a continuous surface. The continuous field allows for greater hold at low pressures which results in less stress on the myocardium in comparison to the discretized vacuum field of the Octopus' suction cups. This continuous vacuum field also contributes to better vision in the surgical field as it permits “inhalation” of the myocardial tissue surrounding the suturing site into it. If placed across the coronary artery being worked on, this action pinches off the coronary artery where inhaled, preventing bleeding within the area the stabilizer surrounds.

Another difference between the stabilizer of this thesis and the Octopus is the chosen method of holding the device in the surgical field. The Octopus is mounted to the operating room table, while the stabilizer of this thesis is mounted to the patient's retractor. This is important as mounting to the chest retractor couples the motion of the stabilizer to the patient. This removes the potential for injury to the patient should they move (perhaps due to coughing) during the surgery.

Finally, the stabilizer system of this thesis has the advantage of quick application. The methods of CTS, Guidant and Medtronic rely on screw tightening to fixate their various mechanisms. This is easy to construct mechanically, but is problematic due to the potential for joint slip, and inconvenient due to tedious and time consuming setup and application [15]. This is particularly true of the Octopus because of its cumbersome supporting arm and table mounting. This is known from experience gained in using the initial arm for this research, the rod and snug experimental arm (detailed in Section 5.2.1), which
required hand tightening of joints to lock in place, and from the experience of surgeons having used the other devices ([15], [16]). The arm of this thesis is extremely quick to activate as it relies on pneumatic power for joint braking and its small size requires little air to be pressurized.

Surgical Arms

Other surgical arms have been developed by such companies as Andronic Devices and Computer Motion. Andronic Devices developed the Andronic Endex arm ([17], [18], [19]) for holding, positioning and supporting patient limbs during operation. The intended application requires an arm that can withstand much larger loading than that required in this application. To accommodate the larger loads, the Andronic arm was endowed with force multiplication levers on its pneumatic actuators and a great deal more bulk material to withstand the stresses from its weight and that of the patient. As such, although the arm is similar in that it uses pneumatic actuation, the arm is substantially different in its complexity of design and size. In contrast to the surgical arm presented herein, this arm cannot be mounted to the patient but is instead mounted to the operating room table.

Computer Motion ([20]) has also made an arm, the AESOP (Automated Endoscopic System for Optimal Positioning) 3000. It is a voice controlled active arm used for maneuvering and positioning an endoscope during minimally invasive heart surgery procedures. This arm is not used for cardiac stabilization.

Another arm of interest is the "automatic, medical holding device" described in [21]. This holding device was developed for the purpose of relieving surgical staff of the tedious duties of holding endoscopic and orthopaedic tools. It is a passive arm similar in design to the pneumatic and pistonless pneumatic arms of Chapter 5 in its use of ball joints between the links for maneuverability. However, the holding device differs in its locking mechanism. Instead of pneumatic air pressure, it uses piezoelectric actuators as wedges to jam the balls in their desired locations. While this provides an effective holding device, it poses a potential hazard due to the high voltages required to expand the piezoelectric actuators. Other concerns with devices such as this would be its high power requirements, heat dissipation for extended use, electromagnetic interference with electronic operating room equipment, sterility issues, and the presence of wires in the surgical field.

1.2.3 Prior Art in Optical Stabilization

Currently there is no method of heart stabilization similar to the optical stabilizer presented in Chapter 2.
However, something similar (in the sense that it relies on a still view of the heart being presented to the surgeon) has been proposed in [22] and [23]. These sources propose that a "virtually immobilized" view of the heart could be provided to the surgeon by using cameras mounted to a slave robot moving in a cardio-stationary orbit. The orbit would be followed by tracking and maintaining a constant distance to beacons or anatomical features on the heart. It is also proposed that this would be small enough to be placed in the thoracic cavity through a port. The tracking would require expensive laparoscopic cameras and stereoscopic vision systems, high speed image processing, extremely accurate controllers, and motors and actuators with very small response times. Should it prove possible though, this robot would effectively cancel the motion of the heart for the surgeon and allow them to operate telerobotically on what would effectively be a still object.

Optical stabilization similar to this thesis' method, but in a more sophisticated implementation, would be through the control of stereoscopic camera images presented to the surgeon. A view of the heart could be captured and presented to the surgeon at a certain point each heart cycle and held frozen until the same point of the next cycle. Only the image of the heart would be taken from the frozen view. Overlaid on top of that image would be images of the surgeon's hands and tools moving in real time, so as to prevent the disorientation that would come from losing their tools due to motion occurring between successive frames. Image processing here would be required for object extraction (the tools, hands and environment from the still shot), insertion (the moving tools, hands and environment would have to be added back in) and embellishment (previously covered areas on the 'still' heart that become exposed would have to be filled in). Computation demands and necessary information bandwidth for this would be tremendous and costly.

The research into optical stabilization presented in this thesis is a precursor to these more complicated and expensive methods.

1.3 Objectives of this Research

As stated previously, for reasons of ease of operation and reliability, the majority of coronary artery bypass procedures are performed while the CPB pump is in use. This is a process which causes damage to and expense for the patient that can be avoided through the use of techniques and devices permitting the operation to proceed while the heart is beating.

The overall objective of the research of this thesis was to investigate and develop techniques and
devices for optical and physical heart stabilization during CABG surgery. The optical devices developed are heart strobes to be turned on at the end of the diastolic period when the heart is at its fullest and moving the least. This provides the surgeon with a stable view of the heart so that suturing can proceed as if the heart were still. The device developed for physical stabilization of the heart was a rigid platform that adhered to the heart at the location of grafting, halting the motion of the heart at that point. To support this stabilizer, the design and construction of an appropriate surgical arm was undertaken.

The design process for the optical methods of cardiac stabilization and evaluations of the devices realized follow immediately in Chapter 2. Details of the testing of this concept follow this in Chapter 3. A discussion of the design, implementation and evaluation of the physical cardiac stabilizer and the surgical arms developed follows in Chapters 4 and 5 respectively. Conclusions gathered from this research appear in Chapter 6. In the appendices can be found a discussion of heart function (A), an alternate ring design (B), detailed calculations for the braking torque produced in the joints of the pneumatic surgical arm (C), and the forms used in the testing procedures (D).
Chapter 2

Optical Stabilization of the Heart

Optical stabilization through strobing of the heart was an idea of Dr. Samuel Lichtenstein [15]. It is an attempt to stabilize the heart in the surgeon's mind by providing them with a view of an immobilized heart. This allows the surgeon to make grafts without the heart motion distracting or impeding their efforts. While the advanced methods of master-slave heart tracking robots and computer controlled vision systems described in the previous section, Section 1.2.3, would perform their stilling of the heart with much more expensive equipment, the work of this chapter provides a simple and inexpensive method to achieve the same result. All that is required here is a bright light synchronized with the patient's EKG signal to flash for a brief period of time each heart cycle. The flashing is made at the end of the diastolic period when the heart is at its maximum volume, moving the slowest and its surface is closest to the surgeon. This time period is indicated on the chart of EKG, volume, pressure and heart sounds for the heart cycle, Figure 2.1, as e), the “Range of Best Strobing Times”. Strobing during this period presents the surgeon with a guideline of the safest times to stitch.

To flash the strobe, the patient's EKG signal is first input to the strobe's control circuitry where it is filtered to detect the R-wave. The R-wave then triggers the strobe according to the surgeon adjustable parameters of delay, ON-time and strobing frequency. Implementation of this is pictured in Figure 2.2. The delay parameter sets the time between R-wave detection and turning on the strobe. It allows the triggering of the strobe at different times in the cycle which can be useful for handling the variation of systolic periods and cases with irregular heartbeats. The ON-time parameter sets the duration time of the strobe light. The strobing frequency is the frequency at which the strobe is triggered during the ON-time. It is required for flash tube strobes only and is used to make the light which needs to be a series of discrete light pulses as dictated by the charge-discharge cycling time. The result of this was the driving signal to control the strobe. The relationship of the ON-time and delay (labelled EKG detector out) signals to the EKG wave can be seen in Figure 2.3.

This chapter will begin with a discussion of the circuitry used to detect the R-wave and to drive
Figure 2.1: Heart cycle EKG, pressure, volume, heart sound and recommended times for strobing relationships (modified from [3])
Chapter 2. Optical Stabilization of the Heart

2.1 Circuitry

The circuitry implemented to perform the optical stabilization is divided into two segments: the circuitry for R-wave detection and the circuitry to produce the signals to drive the strobes.

2.1.1 R-Wave Detection

The most distinct, and thus the easiest element of the EKG to use as a trigger source, is the sharp R-wave peak (see Figure 2.1). Filtering could be performed to find other elements but these would require more complicated signal processing most easily implemented with data acquisition hardware and software based signal processing. Software based signal processing of the EKG signal was considered for its capability and flexibility in sampling, storing, filtering and analyzing the signal. However, it was not implemented for reasons of complexity, time consumption and cost - particularly as it demands staff computer training, and robustness and reliability issues for the computer hardware in the operating...
Because of these problems, a hardware approach to detection was taken using discrete electronic components. The implementation used was a modified version of a circuit first published in [24]. The version used in this research is shown in Figure 2.3. This circuit was simple yet reliable, kept simple for the reason that patients produce a relatively stable, high quality EKG signal during surgery. As a result, the author believed complex detection circuitry such as adaptive filtering or first and second derivative based thresholding were not required, but amplitude detection alone would suffice [24]. Other circuits performing the same task but with some advancements and differences dependent upon their specific application can be found in [25], [26], [27], and [28].

The circuit used here takes as input a one to three volt peak-to-peak EKG signal from a mono-output jack. The signal is filtered with a passband filter centered in the range of fifteen to seventeen Hertz to maximize the QRS complex energy [29]. It is then rectified to remove negative components. R-wave spikes are then found by comparing the signal voltage level to a voltage level of sixty percent of the previous spike. Should it be larger than this the value is sampled for comparison with future potential spikes and passed through a switch to trigger a monostable. The output of this is a square pulse of 5 V amplitude and variable width according to the setting of the potentiometer. The leading edge of this pulse marks the onset of the R-wave. This is illustrated in Figure 2.4.

2.1.2 Strobe Driving Circuitry

The circuit from [24] was used mainly to provide a reliable means of detecting the EKG R-wave. From this was required further circuitry to produce the signals for delay and ON-time that would be required to properly drive the strobe. To produce delay, the monostable producing the R-wave detection pulse had an adjustable output length (controlled by the time constant of the resistor-capacitor network at the monostable’s R/C pins, as labeled in Figure 2.3). This signal triggered the monostable for the ON-time parameter with its falling edge. The ON-time determines the length of time per heartbeat that the light is activated. As with the delay monostable, the ON-time output was also variable in length due to a variable resistor-capacitor network on its timing pin.

The delay pulse and the ON-time pulse both have adjustable lengths up to the period of the heartbeat, beyond this the pulses overlap and form a DC output. Thus, the surgeon can position the strobing period (the period of the ON-time signal) wherever they like during the EKG cycle and can view the heart in a time window of whatever length he or she feels comfortable suturing in.
Figure 2.3: EKG R-wave detection circuit
Interfacing to the Hospital Monitors

To adjust the strobe with respect to the patient’s EKG, the signals need to be displayed concurrently. In a lab setting this is done easily by comparison of oscilloscope traces of the EKG signal and the output ON-time signal. However, this is bulky, expensive and non-intuitive to hospital personnel untrained in electronics. As such, circuitry was implemented to display the output waveform on the hospital monitor.

This was done by creating a circuit to imitate a pressure sensor, an arterial line pressure sensor, that can be interfaced to the monitor. These sensors are strain gauges that output 50 μV/mmHg. When activated they generate a differential signal of 1 mV or less. This was duplicated by passing it through an amplifier circuit with fractional gain to the millivolt range, and then filtering it for noise and buffering it to match the high impedance inputs at the monitor.

The circuit was seen to produce the desired voltage waveforms in lab testing, but has yet to be interfaced to an OR monitor.

2.2 Lighting Methods

The lighting implementation was next considered. It was required to be bright, of quick response both for turning on and turning off, and safe to operate in the operating room. Different types of lighting such as light emitting diodes (LEDs), incandescent halogen bulbs, and flash tubes were tried against these constraints. As will be detailed below, the best solution was one using a digitally triggerable flash tube strobe light.
2.2.1 The LED Strobes

The first lighting methods attempted used LEDs. This was to take advantage of their ease of operation and implementation, and their low power requirements.

LEDs are composed of two types of semiconductors, of p and n type material, meeting at a junction. They convert electrical energy, in the form of a voltage across this junction, into light. The light energy produced is from the rise and fall of electrons in their electron shells. This causes the emission of photons. LEDs only produce light when this energizing voltage is present, making them extremely quick to turn on and off. The photons appear at very specific wavelengths according to the shells they move between. This gives the light produced a very narrow bandwidth. They require little power, due to low current consumption (generally between 10 and 40 mA) and low voltage requirements (between 1.4 and 1.8 V).

Three attempts were made at implementing the LED strobes. The first two methods used visible light - one scheme using high candlepower LEDs of one wavelength and the other low candlepower LEDs of different colours to produce a fuller spectrum of light. The third method was an implementation using infra-red LEDs.

Triggering for all of these methods used the circuitry displayed in Figure 2.5. The ON-time signal was used to drive an n-channel MOSFET. This then sank the appropriate amount of current through the LEDs allowing them to turn on.

![LED triggering circuit diagram](image)

Figure 2.5: LED triggering circuit
Chapter 2. Optical Stabilization of the Heart

The first strobe was constructed using high efficiency AlGaAs LED lamps, each capable of 2500 mcd (millicandella) of light output with driving currents of only 30 mA. In total the light used sixteen LEDs, in four strings of four LEDs each, for a total maximum output of forty candle power of light. The candle power produced was bright enough to light up the heart and to strobe well, but due to the mono-colour nature of the LEDs, the discernment of the anatomy and the recognition of bleeding became much more difficult, particularly as the light produced was only at the one frequency, approximately 650 nm (red). Currently industry research is being conducted to reduce the wavelengths of high powered, high efficiency LEDs but 590 nm (amber) is currently the smallest wavelength.¹

As an attempt to improve the discernment of the anatomy, low powered LEDs of red, green and yellow were made into a strobe.² Eight strings of five LEDs, alternating in colour were used to produce a total of 1200 mcd of light power. These lights were configured as four strings arranged as straight lines to form a “plus sign” shape and four strings forming squared circles placed in each of the plus’ quadrants. The result was an improvement in discernment of blood and anatomy, but a strobe that produced too little light to be useful in an operating room.

The third LED strobe is the infra-red strobe. It operates as the other LED strobes do, except that the light produced is outside of the human visible spectrum. In order to see the strobe on the heart, a CCD camera and television monitor are used. CCD cameras use silicon to receive incoming light. Silicon has a bandwidth extending into the infra-red range enabling it to register the light energy output of the strobe that reflects off of the heart. The camera converts the registered light energy into a brightness that is displayed within a range visible to human observers.

This system has been tested for concept and seen to work well, but has yet to be tested in a hospital setting. Due to the results of Section 3.3 however, this implementation will most likely not be pursued.

2.2.2 The Incandescent Light Strobe

An incandescent light using a halogen bulb was also created for strobe testing. It used a 50 W bulb powered by regular AC current from a wall socket and was switched on and off by a semiconductor relay driven by the R-wave detection circuit ON-time signal (see Figure 2.3). As this was the same ON-time signal used by the LED strobe, implementation was simple. As well, the light produced was full spectrum and bright, and operation was silent.

¹These are produced by Hewlett Packard Corporation for use in traffic lights. They are constructed from AlInGaP.
²Blue LEDs are also in production but of limited availability and high cost due to low production yield.
The negative aspects of this bulb are the heat it generates and its response time - it is too slow in turning on and off as compared to the LEDs or flash tubes. Thus, the light cannot provide a sharp, frozen image of the heart. As a result, this form of strobe was not pursued any further.

2.2.3 The Flash Tube Strobes

To get around the problem of providing a full spectrum of light while maintaining good lighting power, it was decided to use flash tubes. These flashes have three terminals: a cathode and anode at either end of the tube, and a base terminal, located between the cathode and anode. They are filled with an inert gas (in this case Xenon) which is ionized by placing a high voltage (4 kV or more) at the base. The ionization allows stored charge to flow between the cathode and anode, which normally have a difference potential of 400 V. The flowing charge causes the ionized gas to release energy in the form of light.

This cycle of ionizing and de-ionizing requires a great deal of energy to occur. As such, the light produced cannot be constant, as would be ideal, but can be produced only in discrete flashes. To make this flashing appear as a constant, continuous source, the driving signal is formed by modulating the ON-time signal with a square wave pulse train of higher frequency. The frequency of this train is controlled by the aforementioned third parameter, frequency. It is set high enough that the flashing would be imperceptible to the eye - similar in concept to viewing movies using film projection.

Two strobes using xenon flash tubes were implemented: a modified engine timing strobe and a lab strobe that could be triggered digitally. They were both driven simultaneously with the circuit pictured in Figure 2.6. In this circuit, the signal for the light’s period of activity (the ON-time signal as created in the circuit of Figure 2.3) is Boolean AND’ed with a generated square wave pulse train, which creates the modulated pulse train needed to drive the strobes. The potentiometer in the pulse train generator allows for tuning of its frequency.

These lights have the advantage over the LEDs of being brighter and of fuller spectrum. Their disadvantages are light flicker, noise due to their charging and discharging cycle, and higher power consumption.

Engine Timing Strobe

Engine timing strobes are handheld gun shaped devices used to determine the proximity of an engine’s pistons to top dead centre when the spark plugs are fired. Triggering is performed through use of an inductive pickup placed around the spark plugs cable. The pickup reads the voltage spike created by the
Figure 2.6: Strobe triggering circuit
ignition coil when it transforms the low tension car battery voltage to a high tension voltage sufficient to promote electrical discharge across the spark gap at the plug. The range of the voltage is from a few hundred volts in the primary winding of the ignition coil, to as much as 25 kV at the gap [30]. Instead of attempting to generate these voltages which could be hazardous in a hospital room, the engine timing strobe was dismantled, its circuit board reverse-engineered and an appropriate location and means of directly interfacing to the circuit board in order to trigger the light was determined.

To trigger the timing light the modulated square wave passed to the digital strobe was buffered and then differentiated to obtain a string of voltage spikes. The negative voltage spikes were removed, the signal buffered again, passed through a variable gain block, high-pass filtered (3.3 Hz cutoff frequency for DC blocking) and then interfaced to the strobe. The timing light interface required a high impedance path to ground for passing the signal correctly. Again, this is shown in Figure 2.6.

This implementation of the strobe worked but had a significant number of disadvantages, including awkwardness in application, low light power, high power requirements, safety and speed of operation. The shape of the strobe gun makes them useful for automotive applications but awkward for surgery, where they'd require an extra set of hands to use. The light power was weak, being visible only in complete darkness. In use they require connection to a power source capable of supplying a great deal of current to make it strobe quickly and repeatedly - generally a car battery, which is a cumbersome, bulky power source that would be awkward in an operating room and definitely unsafe if it was to be suspended. As well, it has the potential danger of leaks and explosion if incorrectly hooked up. Finally, the light was also slow to trigger having a maximum frequency of 45 Hz as timed by the quicker digital strobe.

**Digital Strobe**

The digital strobe is similar to the engine timing strobe in its function but much more capable in its design. It was able to flash at a much higher frequency than the engine timing strobe and do so while powered with a conventional 115 VAC supply. This makes it much safer to use than a strobe powered by a car battery. It has a compact size and can be easily suspended above the patient during operation. Finally, it is easily triggered, requiring only simple TTL logic input. The trigger input used here was a high frequency square wave modulated with the ON-time signal from the EKG detector, as shown in Figure 2.6.

When tested in the hospital animal lab and OR, the digital strobe was seen to work well, adequately
lighting and visually stopping the heart. Its light was brighter than the LEDs and provided for full
discernment of everything in the operating field with its “white” light. The motion of the muscle was
completely removed visually when the heart strobe light was applied and stitching was judged to be
possible.

With these results it was decided that the digital strobe was the most successful of the flash tube
strobes, and of all the lighting methods tested. The digital strobe was the light used in the testing of
optical strobing to be described next in Chapter 3.
Chapter 3

Optical Stabilization Testing

On the way to development of the cardiac stabilization strobe two animal tests and one human feasibility test were conducted. The first trial was useful for assessing the effectiveness of the red LED and low-power multi-colour strobos, as described in Section 2.2.1, and for testing and debugging the strobe's R-wave detection circuit. In the second animal trial and in the only human test, the flash tube strobos were tested to determine which was the better strobe (the digital strobe, for the reasons outlined in Section 2.2.3) and to determine whether or not optical stability could be achieved with the strobos.

Due to the nature of the circumstances and the ramifications of problems should they have occurred, the effect of the strobe on the motion of the heart was only observed, and grafting was not actually performed. Optical stabilization was observed, so the next step was to determine whether or not it would be useful for the surgeon. The chosen route for this was through the design and construction of a test apparatus and testing procedure, as will be detailed in this chapter.

3.1 The Testing Apparatus

The device constructed was a simple one DOF device that rotated a platform back and forth as a simplistic mimic of the heart's motion while beating. On the platform was a stretched piece of latex to duplicate cardiac tissue for suturing purposes.

The test apparatus consisted of several components: the rotatable suturing platform, the motor to rotate it, an encoder for sensing motor position, mounting brackets, connecting rods and an aluminum plate base to which it was all mounted. This, excepting the rotating platform, was all covered by a plastic sheet to provide hand rests for the operator and to cover the circuitry. As well, a roughly 60 cm tall table was constructed to stand over this apparatus supporting the strobe. A square hole was cut in the table through which the strobe light could shine onto the platform. A view of the platform hardware appears in Figure 3.7.

In Figure 3.8 a block diagram for signal flow in the test apparatus system can be seen. It can be
Figure 3.7: Test apparatus setup: strobe (top), EKG filter, power supplies, signal generating computer (bottom, left to right at rear), rotating platform (foreground)
described as follows. In operation, computer software generates voltage levels proportional to the desired angle for the platform. These are then converted into PWM outputs by a National Instruments DAQ700 data acquisition card. The PWM signal is low pass filtered to reproduce the desired angle\(^1\). The desired position voltage input is compared to a reference voltage to trigger the strobe and to a voltage reading across a potentiometer attached to the motor’s drive shaft to determine the platform’s angle error. The angle error signal is then inputted to an analog PID controller whose output is used by a current amplifier to drive the motor.

![Strobing test platform overview](image)

Figure 3.8: Strobing test platform overview

The motor that was chosen for the rotation was a 90 W Maxon motor. It was chosen for availability and its maximum torque rating of 1.1 Nm, which exceeded the required torque for the application. Indeed, approximating the distance to the application of the surgeon’s stitching to be about 10 cm and the force of the heart there (a typical maximum for the range obtained in the force torque measurement and scaling of Section 4.1.1 and displayed in Figure 4.19) to be 6 N, the required motor torque was 0.6 Nm.

3.1.1 Control

The controller for the motor current was an analog PID control loop. It was implemented with electronics, using operational amplifiers to handle the necessary addition and subtraction to produce the error signal, summation of gains, integration and differentiation. Inputs to the controller were for power, desired angle (servoing position) for the platform, actual angle, and an offset input to provide for variable platform initial positions. The actual angle of rotation of the platform was a voltage measured across a

\(^1\)The signal generation could not be done directly as the data acquisition card did not have the capability for analog output.
potentiometer attached to the motor’s drive shaft.

The error signal was calculated as the sum of the desired angle and offset subtracting the feedback actual position. It was amplified according to the PID gains and the result was used as input to the differential amplifier input of the current amplifier for the motor.

A circuit diagram of the controller appears in Figure 3.9.

3.1.2 The Servoing Signals

Three signals were used as input to the controller. The first was a measured waveform for motion of the medial LAD coronary artery. The second and third were artificial waveforms with characteristics based on the first. The second was a simple exponential rise and fall waveform based on the maximum measured motion of the medial LAD. The third was a more complicated motion made of a series of exponential rise and fall waveforms modulated by a sinusoidal signal to better represent the true motion of the coronary artery. These all appear in Figure 3.10.

Heart Motion Measurement and the Measured Waveform

In the top graph of Figure 3.10, a period of the measured motion of the medial coronary artery of a pig can be seen. This measurement was made in a lab experiment on two pigs using a linear spring potentiometer held in place by the rod and snug experimental arm. The linear potentiometer was set against the pig’s beating heart such that the motion of the heart pushed against it to create a stroke that was transduced into a recordable voltage. This apparatus can be seen in Figure 3.11. There, it will be noted that the pig lies in the bottom of the picture with heart exposed by the chest retractor. Mounted on the chest retractor is the experimental rod and snug arm forming a loop, at the end of which is the spring loaded linear potentiometer contacting the heart.

The medial LAD motion was noted to be a waveform of good magnitude and of a more regular nature (in terms of repeated heart and lung motions being near identical) than the other measurements. As such, it was used as the basis for the artificial waveforms that were used to drive the platform. It was also the waveform with the largest magnitude motion, at its maximum showing movement of 2.22 cm. The other measurements taken were similar in shape but smaller in maximum range.

The measured signal was not used to drive the platform directly during the testing trials. This was because a simpler, more easily described waveform, with more definite and distinct motions was requested by the surgeons previewing the test apparatus.
Chapter 3. Optical Stabilization Testing

The Artificial Waveforms

For the testing, simple, easily described waveforms were desired. These were created by using the measurements of the medial LAD for deciding the relative motion range and contraction-expansion timing. Two waveforms were created: a simple exponential growth and decay motion and a more complicated, more heart-like motion.

The waveform displayed in the middle graph of Figure 3.10 is a simple exponential rise and fall waveform with rise time double the fall time, to represent a typical diastolic-systolic relationship. This waveform was used for the simple motion tests and as such did not have modulation due to the lung motion. Mathematically it is described as:

\[
y(t) = 1 - e^{\frac{-a}{T^2}}, \text{ during diastole, and } \quad (3.1)
\]

\[
y(t) = e^{\frac{-12(a-\frac{2}{7})}{T^2}}, \text{ during systole. } \quad (3.2)
\]

The term \(a\) is a counter for the sample number in the current heart beat. The term \(t\) represents the sample number for the current lung motion. \(T\) represents the period of lung motion.

The last waveform in Figure 3.10 is the complicated motion that includes lung action effects on the heart’s position. To do this the exponential rise and fall of the simple test was modulated by a sinusoid of period seven heart beats, in accordance with the medial LAD waveform’s lung action period. The equations for this are:

\[
y(t) = (2 - \left| \sin\left(-\frac{\pi}{2} + \frac{\pi}{T}t\right) \right|)(1 - e^{\frac{-a}{T^2}})), \text{ during diastole, and } \quad (3.3)
\]

\[
y(t) = (2 - \left| \sin\left(-\frac{\pi}{2} + \frac{\pi}{T}t\right) \right|)e^{\frac{-12(a-\frac{2}{7})}{T^2}} \text{ during systole. } \quad (3.4)
\]

It will be noticed that the graphs for these waveforms have unitless magnitude ranges between zero and one. The software written to produce these outputs provided for programmable gains so that the output waveform could be tuned should variation in the output be desired by the surgeons verifying the test setup. The motion of the endpoint of the platform was scaled according to the interspecies scaling principles outlined in Section 4.1.1 for lengths and distances. Scaling was done from a pig mass of 25 kg to a human mass of 200 kg (again, refer to Section 4.1.1). This resulted in a doubling of the values of the measurements made on the pig, meaning a maximum displacement of 4.44 cm at the tip of the platform.
3.1.3 Strobe Triggering

The circuitry that triggers the strobes normally uses as input an EKG signal, typically as appears in Figure 2.1. As there was no live heart producing an EKG signal during the testing another signal had to be used to trigger the strobe. This was the computer generated desired position signal.

In order that it could be used to trigger the strobe this signal was integrated to produce the desired position as a continuous non-PWM waveform and then a voltage comparator was used to compare the desired position with a set threshold. This produced a train of square pulses which were input to the R-wave detection circuit. The R-wave circuit then detected the rising edges of the pulses and triggered the strobe accordingly.

3.2 Suturing Tests

In pre-trial testing of the test apparatus, the surgeons judging the platform determined that suturing to the platform would be too difficult to do for testing purposes. Instead, the suturing procedure was replaced with marking targets on a sheet of paper attached to the moving platform, with a pen. To help the subjects by preventing them from losing track of the pen during the strobe OFF-time, glow-in-the-dark stickers were attached to the pen. This was to remedy the problem that arose during pre-trial testing, where the surgeons found it difficult to maintain concentration between strobings of the heart as they would “lose” their instruments in the dark.

The subjects were to mark six targets during each test: six dots configured in two columns of three dots each. Dots in the column were spaced apart by 1.25 cm. The columns were separated by 2.5 cm. To maintain consistency between subjects and between tests, the pattern was placed in the corner closest to the subject’s hand of preference. In the picture of Figure 3.12 the platform with target sheet is seen set up for a left handed person. The subjects were given as much time as they desired to practice and perform each test. Test subjects sat in a chair placing them at a height from which they could comfortably rest their arms on the platform if they desired. A picture of a subject performing a test can be seen in Figure 3.13.

There were five tests, varying according to platform motion and use of strobing, that the subjects performed. These tests were:

- Test 1: Stable platform.
- Test 2: Moving platform with simple motion (middle waveform of Figure 3.10).
Chapter 3. Optical Stabilization Testing

- Test 3: Moving platform with simple motion and optical stabilization in effect (strobing).
- Test 4: Moving platform with complicated, heart-like motion (bottom waveform of Figure 3.10).
- Test 5: Moving platform with complicated, heart-like motion and optical stabilization in effect.

The stable platform test was not important for error or timing, but was instead used as a control for subjects with poor hand eye co-ordination. This was not an issue, however, as all subjects were able to accurately place dots in the holes.

The motion tests were to qualify the difficulty in suturing to a moving heart. The use of the two motions for the testing was to determine if there would be a significant advantage to one form of motion with the strobe than another. Before the testing began, it was theorized that the simple motion would be easy enough to track by the subjects that not much error would be introduced with or without the strobe. Then, the real benefit of the strobe would not be seen until the motion became significantly complicated that the human subject could not track it. For this reason and for the reason of increased realism, the complicated test was implemented as well. As well, this could be used to determine the necessary level of complication for future tracking mechanisms should they be employed with optical stabilization.

During the tests without strobing, the room lighting was set for normal reading conditions. When strobing was in place, the room lights were turned off.

3.2.1 Test Evaluation

Subjects were allowed to make six marks on the paper during each test (during practice time this was unlimited) - one mark per dot in the corner being worked on. Evaluation of the performance during the test was both objective and subjective. The objective tests were performance based quantitative measures of the subject's tests. The subjects were also required to make a subjective assessment of themselves and how they perceived the tests in the form of a questionnaire administered once testing was completed.

Performance was judged by time to completion, placement accuracy and the neatness with which the marks were made. Placement accuracy was measured by calculating the total distance the marks were made from the centre of the circles for each test. Neatness errors were incurred if the subject did not place a dot, but instead their mark was smeared. This usually occurred when the platform hit the pen when the subject was not expecting it. These errors were quantified by summing the lengths of the smears for the six attempts. Performance measures were considered for the second through fifth
tests in which the platform was moving. Once again, the first test was merely used as a control for determining adequate hand-eye coordination and substantial enough error was not present for its use to be meaningful.

Subject judging was made according to their perceived comfort level (confidence in what they were doing), ease of completion, demands on concentration and fatigue. Judgements were made on a scale from one to ten according to the applicability of the perceived parameter. Using the stable test as a baseline, it would be expected that the subjects would perceive a ten for comfort level and ease of completion, and a one for concentration demands and fatigue.

Copies of the forms that were used to record this information are included in Appendix D.

3.3 Results

Twenty-two subjects completed the tests. They were composed primarily of students, but technicians, secretarial staff and other non-university personnel also took part in the experiments. These were people who would normally not perform tasks similar to this, so, to compensate, training in the form of practice on the rotating platform for each test was allowed until the subject felt ready to do the test. The age and standard deviation of the subjects ranged from 22 to 42 years of age. Nine of the subjects were older than 30 years; thirteen were 30 and under. The average age of the subjects was 28±6.5 years. There were eight female participants and fourteen males.

As can be seen in Table 3.1, as the tests proceeded the subjects perceived them to become more difficult. Comfort level and ease of performance were both seen to decrease, falling from averages of 9.91 and 9.73 to 4.41 and 4.05. Demands on concentration and fatigue were seen to rise from average values of 1.86 and 1.23 to 8.00 and 5.55. All of these results run contrary to the hypothesis that the strobe would benefit the surgery. This data is also charted in Figure 3.14.

Table 3.1: Group means and standard deviations of subject perceptions during the strobing tests (scale of 1 to 10 questionnaire replies)

<table>
<thead>
<tr>
<th>Perception</th>
<th>Test 1</th>
<th>Test 2</th>
<th>Test 3</th>
<th>Test 4</th>
<th>Test 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Comfort Level</td>
<td>9.91 ± .29</td>
<td>7.32 ± 1.62</td>
<td>6.36 ± 1.81</td>
<td>5.41 ± 1.92</td>
<td>4.41 ± 2.32</td>
</tr>
<tr>
<td>Ease of Performance</td>
<td>9.73 ± 1.08</td>
<td>6.82 ± 1.94</td>
<td>5.95 ± 1.94</td>
<td>5.27 ± 1.96</td>
<td>4.05 ± 2.36</td>
</tr>
<tr>
<td>Demands on Concentration</td>
<td>1.86 ± 2.01</td>
<td>5.41 ± 2.58</td>
<td>6.90 ± 2.71</td>
<td>7.00 ± 2.25</td>
<td>8.00 ± 2.43</td>
</tr>
<tr>
<td>Fatigue</td>
<td>1.23 ± 0.69</td>
<td>2.82 ± 1.56</td>
<td>3.77 ± 2.30</td>
<td>4.41 ± 2.38</td>
<td>5.55 ± 2.96</td>
</tr>
</tbody>
</table>
The timekeeping and error measurements backed up the results of the subjective tests, further indicating the strobe would not be an aid to surgery. By analysing the group means and standard deviations for the subject performance for both simple and complicated motion, time required, placement error and neatness error were all seen to increase for the tests using the strobe, as compared to those without. As can be read in Table 3.2 and seen in Figure 3.15, during the simple motion test, average time to completion increased from 17.85 to 25.17 seconds while placement and neatness errors increased from 13.30 to 23.41 mm and from 14.1 to 21.39 mm. During the more heart like motion tests, the same trend was noticed. Time increased from 21.6 to 33.35 seconds, placement error increased from 13.17 to 26.91 mm and neatness errors increased from 11.21 to 14.76 mm.

Table 3.2: Group means and standard deviations for subject performance during the strobing tests

<table>
<thead>
<tr>
<th>Performance Measure</th>
<th>Test 2</th>
<th>Test 3</th>
<th>Test 4</th>
<th>Test 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time (seconds)</td>
<td>17.85 ± 9.61</td>
<td>25.17 ± 15.87</td>
<td>21.60 ± 15.25</td>
<td>33.35 ± 16.66</td>
</tr>
<tr>
<td>Placement Errors (mm)</td>
<td>13.30 ± 8.66</td>
<td>23.41 ± 13.40</td>
<td>13.17 ± 6.10</td>
<td>26.91 ± 9.73</td>
</tr>
</tbody>
</table>

Significance testing, and subsequent power testing, of the objective test data at the five percent level verified the group mean and standard deviation conclusion. It was carried out according to the following hypotheses:

1. $H_0$: the strobe was beneficial in the tests as compared to the non-strobed test conditions: $\mu_{strobed} < \mu_{non-strobed}$ for time required, and placement and neatness errors made.

2. $H_1$: the strobe had no effect on subject performance: $\mu_{strobed} = \mu_{non-strobed}$.

3. $H_2$: the strobe impeded subject performance: $\mu_{strobed} > \mu_{non-strobed}$

The results of the testing can be seen in Table 3.3, where $\mu_x$, refers to the mean for Test $x$. The subjective measures were not evaluated due to their more qualitative nature.

From the table it will be noted that the first hypothesis was rejected immediately for both simple and complicated motions. Therefore, the strobe was proven to not be beneficial to subject performance.

Testing of hypothesis $H_1$ showed some possibility, with acceptance for simple motion time requirements and complicated motion neatness error. However, it was judged that the strobe having no effect was not very likely as the power test for simple motion time requirements revealed that the hypothesis of no effect would be rejected nearly half the time.
Table 3.3: Significance and power testing of the unpaired data

<table>
<thead>
<tr>
<th>Hypothesis</th>
<th>Time</th>
<th>Placement Error</th>
<th>Neatness Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>$H_2 &lt; H_3$</td>
<td>failed</td>
<td>failed</td>
<td>failed</td>
</tr>
<tr>
<td>$H_2 &lt; H_4$</td>
<td>failed</td>
<td>failed</td>
<td>failed</td>
</tr>
<tr>
<td>$H_2 = H_3$</td>
<td>passed</td>
<td>failed</td>
<td>passed</td>
</tr>
<tr>
<td>$H_2 = H_4$</td>
<td>failed</td>
<td>passed</td>
<td>passed</td>
</tr>
<tr>
<td>$H_2 &gt; H_3$</td>
<td>failed</td>
<td>passed</td>
<td>failed</td>
</tr>
<tr>
<td>$H_2 &gt; H_4$</td>
<td>passed</td>
<td>passed</td>
<td>failed</td>
</tr>
</tbody>
</table>

This left hypothesis $H_2$. From Table 3.3, it is seen that four of six conditions were passed at the 5% significance level, although the power test for simple motion neatness errors indicated a good likelihood of future tests rejecting this hypothesis. All in all this revealed the second hypothesis to be very likely.

To better confirm this conclusion, paired testing was also performed. This was used to reduce the variance - and thereby achieve a more accurate assessment - between the tests by using ratios of performance for comparison. For each subject, ratios were calculated for time required, and placement and neatness errors made for each of the platform motions. The new means and standard deviations were then used as before to assess the previous hypotheses for significance levels of five percent and to calculate accepted hypothesis power. The recalculated means and standard deviations appear in Table 3.4 and the resulting significance and power tests are documented in Table 3.5.

Table 3.4: Paired testing means and standard deviations for subject performance during the tests

<table>
<thead>
<tr>
<th>Performance Measure</th>
<th>Test 2 vs. Test 3</th>
<th>Test 4 vs. Test 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time</td>
<td>1.51 ± 0.94</td>
<td>1.73 ± 0.61</td>
</tr>
<tr>
<td>Placement Error</td>
<td>2.42 ± 1.74</td>
<td>2.99 ± 2.88</td>
</tr>
<tr>
<td>Neatness Error</td>
<td>1.89 ± 1.26</td>
<td>2.13 ± 2.28</td>
</tr>
</tbody>
</table>

From Table 3.5, it can be seen that the hypothesis of the strobe being beneficial is outright rejected, the hypothesis of it having no effect is extremely unlikely, and the hypothesis of the strobe having a negative impact is assured. All but the neatness error significance test for the complicated motion passed, and with probability of future data sets rejecting this hypothesis of less than 30%.

Based on all this it can be said that despite the good intentions of the strobe in helping surgery, the device is actually a hindrance. Surgeons can expect to be just as, or even more, quick and accurate by simply tracking the motion of the heart by eye. The reason for the failure of the strobe in aiding the
users seems to be a matter of human reaction times being too slow. A recurring comment during the tests that affirmed this was, “By the time I put the pen down, the light was off”.

3.4 Recommendations

Despite the tests made here showing this implementation of optical stabilization to be faulty, the whole concept of virtual immobilization of the heart still has merit. To pursue this idea further, efforts should be made to ensure that the devices and techniques developed do not rely on human reaction times to accomplish their tasks. A camera system that provided a gradual warning of the impending time to suture for the surgeon (potentially by gradually changing a color indicator on the screen) could allow the surgeon to get their hands in motion and thereby reduce their reaction time. It is possible, but based on the results presented and due to the much more complicated matter of performing actual suturing, this process would be extremely difficult to successfully implement.

A complete relaxing of demands on reaction time as would be the case with the master-slave cardio-stationary robotic solution described in the introductory chapter, would also be effective. However, it has been said ([15]) that the attempt to get the robot to perform suturing may be a mistake as it replaces a task performed well by humans with one performed, at best, poorly by robots. Thus, the task should be a hybrid one. Computer vision based tracking of the heart could be performed and a platform implemented to follow the heart’s motion and keep the heart stable for the operator, but tools attached to this table to allow the surgeon to perform the suturing should still be in place.

As a simple demonstration of a hybrid system like this and a demonstration of the effectiveness of optical stabilization, one needs only to secure a camera (or cameras) and their hands (in a safe manner) to a moving platform (either moved by machine or by a collaborator, and potentially as simple as a
book). Then, a task can be attempted on the platform while receiving visual cues through a monitor or stereoscopic goggles connected to the camera(s). At first, this may feel strange, however, once the user gains their bearings and becomes relaxed, this will quickly change and the task can be completed on the moving target as easily as if being performed on a still one.

In the mean time, while hybrid human-robot optical stabilization systems are being developed, surgeons can rely on physical stabilization measures to allow grafting without the use of cardio-pulmonary bypass equipment. The solution developed for this thesis is discussed in the next chapter.
Figure 3.9: Electronic circuitry for testing platform
Figure 3.10: Heart waveforms: measured (top graph), and artificial: simple (middle) and complicated (bottom)

Figure 3.11: The heart tissue position sensing setup
Figure 3.12: Unmarked target sheet on platform for testing of left handed subject

Figure 3.13: Subject performing a visual stabilization test
Figure 3.14: Subject perceptions from performing the optical stabilization tests
Chapter 3. Optical Stabilization Testing

Figure 3.15: Subject performance during the strobing tests
In addition to optical stabilization, it was desired to have a device for physical stabilization of the heart. This device would contact the heart and hold a section of it still so that the surgeon could suture to it directly. The method of physical stabilization implemented here is a hollow ring shape that attaches rigidly to the heart, creating within its opening, a flat unmoving surface that the surgeon may suture to unimpeded by the beating of the heart. This ring is connected by a rigid link (the surgical arm to be described in Chapter 5) to the patient through the retractor.

The primary design concern was the means of attachment to the heart. The methods of attachment investigated included needles, hooks or retractable teeth sunk into the myocardium, velcro attachment, clamps or grippers, and vacuum suction. Needles and hooks were rejected as it was thought their implementation would be too mechanically complex, their deployment would cause too much damage to the myocardium, and the bleeding resulting from these injuries could obscure the suturing site. Velcro methods would be difficult to implement as the attachment to the heart would present the same problem of accurate and reliable stitching to the heart. Clamps and grippers of the heart itself could be made to hold the heart by enveloping it. However, this was deemed inappropriate as the shock of this can cause fibrillation to occur. This was observed in testing on a pig when its heart was lifted a small distance. Vacuum suction, although it has potential for some bruising, was selected for the attachment method as it was deemed the easiest to construct and apply, and the safest.

4.1 Stabilizer Design Requirements

Once the attachment method was decided upon, the shape and dimensions of the stabilizer were decided according to the constraints that the stabilizer must:

- fit to the heart
- facilitate suturing to take place within it
- permit grafting of both veins and arteries
• maintain a bloodless field so as to provide the best visibility of the anastomosis site as possible
• cause minimal damage to the myocardium
• hold the cardiac tissue beneath it stable.

To form a good fit to the heart, the stabilizer must be relatively small and round so as to follow the topography of the heart and maintain the vacuum seal. This small size will be in conflict with the constraints demanding the stabilizer be sufficiently large to provide room for suturing within it and a strong hold upon the tissue beneath it, due to the proportionality of surface area of vacuum application with holding force.

In order for the surgeon to work with the stabilizer, the inside space must be wide enough for suturing inside and the height of the wall must not interfere. An inner radius of at least 10 mm was decided upon as a lower limit for the workspace. A ratio of this to height of 5:3 was decided on to limit the profile, suggesting a wall height of no more than 6 mm for this inner radius. This limit was to allow the surgeon approximately 60° of approach.

To permit grafting of both arteries and veins, the stabilizer “ring” should actually be horseshoe shaped, as shown in Figure 4.16. Arterial grafts require the artery to remain connected to their blood source, whereas veins, not having access to a supply of oxygenated blood, can be and are completely harvested. Thus, arterial grafts have only one free end. This requires the stabilizer to have a gap to permit its removal once the graft is complete. The width of the gap was required to be 10 mm. This allows mammary arteries to pass through with ease, as they typically are only 4 mm in diameter [31].

![Diagram of Cardiac Stabilizing Ring](image-url)
To prevent blood from interfering with the surgeon’s vision as they suture within the ring, the ring must maintain a bloodless surgical field. To do this, the surface of vacuum application was made continuous and the stabilizer’s vacuum channel of sufficient height to allow partial aspiration of the myocardium, providing a good hold and temporarily pinching off blood flow to the tissue within the ring. In application, the ring would be placed across the coronary artery to be grafted to as shown in Figure 4.16. With the vacuum pressure applied, the portion of the heart beneath the ring is immobilized and the coronary artery is pulled into the vacuum channel. The artery is then pinched at points A and B as seen in Figure 4.16, stopping the flow of blood between the points, resulting in a bloodless field.

The application of a continuous vacuum field also means a more efficient application of vacuum which permits the pressure levels applied to the heart to be as low as possible. This will minimize any damage that the stabilizer might cause to the tissue it fixates. Minimizing the height of the ring also minimizes damage as it reduces the possible amount of strain imposed upon the cardiac tissue.

As the final and most important constraint, the cardiac stabilizer must hold the tissue within it steady so that suturing can take place. As the force being used to hold the heart is provided by vacuum pressure, this constraint is a constraint on the size of the stabilizer - the vacuum surface area is what controls the force given a constant pressure. Thus, the stabilizer must be large enough to immobilize the cardiac tissue in its vicinity without using extreme pressures that would damage the heart.

### 4.1.1 Maintaining a Stable Suturing Area

To quantify the final constraint, the heart force provided by the stabilizer was estimated as a combination of what would be necessary to hold the heart elevated and what would be necessary to resist the forces produced in its regular expansion-contraction cycle. Considering the mass of a large blood-filled heart to be 570 g when in diastole [32], the stabilizer is required to provide a static lifting force of 5.60 N.

To quantify the force necessary to resist the motions produced by the heart’s pumping, the forces and torques from two pig hearts were measured using a JR3 force torque sensor (FTS) and the O-ring (a “o” shaped stabilizing ring to be described fully in Section 4.2.1). These were held in place by an experimental arm composed of stainless steel rods connected by universal snugs, a diagram of which is displayed in Figure 5.23 a). It is further described in Section 5.2). The experimental setup of the FTS positioned above the pig and using the C-ring stabilizer is viewable in Figure 4.17. at the bottom of the picture can be seen the pig with exposed thoracic cavity. Attached to the retractor is the rod and snug arm. The round object towards the middle of the picture is the FTS. Attached to the FTS at the right
side of the page can be seen a bloodied C-ring. The data captured was recorded from the FTS with an National Instruments DAQ700 ADC card, displayed and stored through use of National Instruments' DaqWare software program and then analyzed with Matlab code.

Figure 4.17: The force torque sensor and experimental arm above the pig’s thoracic cavity

Measurements were taken with the ring attached to the pig hearts' base, left and right ventricles. The mean of the data measured was then removed to compensate for the DC offset levels created due to tension and compression in the FTS mounting screws. This also removes any gravity loading present (which will be minor as the heart is not lifted). Finally, the data captured was adjusted for position, orientation and species. The dynamic force requirement determined by this testing was the provision of approximately 20.1 N of hold (see below). Therefore the overall requirement for the ring was that it could provide 25.7 N of force upwards (in the negative $y$ direction of the end effector frame) on the heart.

**Adjusting for Position and Orientation**

Measurements taken by the FTS were for forces at and torques about the FTS centroid. To realize this data at the stabilization device, or end effector (EE), it was first adjusted by subtracting the mean value. This was to remove biasing of the results due to tension and compression of the FTS mounting screws. As well, this removed the gravity loading of the heart on the FTS - hence the need for a static and dynamic force requirement to determine the holding constraint. Then, the data was translated to
Chapter 4. Physical Stabilization of the Heart

the end effector by considering the FTS, its connection plates and the end effector to form a rigid link to allow the problem to be treated statically. The data was reoriented to the stabilizer’s frame of reference by rotations about the \( z_{FT} \) axis by \(-90^\circ\) and about the current \( z \) axis by \( 70^\circ \) (to accommodate a \( 20^\circ \) rotation of the ring relative to the vacuum connection). The ring, the FTS and the end link of the experimental arm can all be seen with their frames of reference, which describe these translations and rotations, in Figure 4.18.

![Diagram of force-torque sensor](image)

**Figure 4.18:** The force torque sensor

Adjusting for Species

Once the data was correctly oriented, it was then scaled to predict equivalent human forces and torques. According to [33], the characteristics of the cardiovascular system of all mammals are functions of their mass. Heart rate, \( \omega \), and cardiac output, \( Q_b \), can be described as:

\[
\omega = 229M^{-\frac{1}{4}} \quad \text{and} \quad Q_b = 224M^{\frac{5}{4}} .
\]  

(4.5)  
(4.6)

As well, [33] claims that interspecies comparisons can be made on the basis of mass. Approximating the length of a blood vessel in a prototype mammal \( (p) \), its length relates to the model mammal \( (m) \) according to:

\[
l_p = \left( \frac{M_p}{M_m} \right)^{\frac{1}{3}} l_m .
\]  

(4.7)

Using the above equations (4.5,4.6,4.7), an inter-species relationship for heart force was constructed
by performing dimensional analysis on the force of contraction:

\[
F = ma = \frac{\text{volume}}{\text{time}} \times \text{density} \times \text{velocity}
\]

\[
= \text{output} \times \text{density} \times \text{dist.} \times \text{freq.}
\]

Then, assuming the blood densities were equivalent, the force translation equation was constructed as:

\[
\frac{F_h}{F_p} = \frac{Q_h \rho_h l_h \omega_h}{Q_p \rho_p l_p \omega_p}
\]

\[
= \frac{224M_h^{3/2} M_h^{1/2} 229M_h^{-1/4}}{224M_p^{3/2} M_p^{1/2} 229M_p^{-1/4}}
\]

\[
= \left( \frac{M_h}{M_p} \right)^{3/8}
\]

Similarly, torque was found as the product of output, density, length, frequency and length again:

\[
\frac{T_h}{T_p} = \left( \frac{M_h}{M_p} \right)^{3/8} \left( \frac{M_h}{M_p} \right)^{1/8} = \left( \frac{M_h}{M_p} \right)^{1/4}
\]

The test subject in the first round of force-torque measurements was a 26 kg pig. The human extrapolation had a mass of 200 kg. This provided a significant safety factor, accommodating application to people ranging in size from small children to large sumo wrestlers. In terms of actual forces and torques, this scaling implies a 200 kg human will produce heart forces and torques 5.5 times and 10.8 times greater than the 26 kg pig.

The adjusted force and torque data at the stabilization device for measurements taken on the pig’s left ventricle are displayed in Figure 4.19. It can be seen that the stabilizer will receive a range of forces and torques between -6 and 5 N, and between -0.6 and 0.4 Nm. The maximum and minimum values for the forces and torques measured on the two pigs and scaled to large human sizes appear in Tables 4.6 and 4.7. These values were used to determine that the dynamic force requirement of the cardiac stabilizer was 20.1 N due to largest force value, the force along the y axis on pig two. Appropriately, this axis is normal to the plane of the ring by definition (see axis $y_{EE}$ in Figure 4.18).

The maximum force and torque values from these tests were also used to calculate the required braking torques at the joints of the surgical arm, as will be discussed in Chapter 5.
4.2 Cardiac Stabilizer Design Alternatives

The cardiac stabilizer went through a number of design iterations as the above requirements were attempted to be met. The design process was structured to first show proof of concept, then pursue design alternatives and in the end, produce the final prototype. The stabilizers built throughout these stages are pictured in Figure 4.20 and a comparison of them is provided in Table 4.8. In this table the forces are calculated assuming a vacuum pressure of 38.1 KPa (approximately 287 mmHg) is being applied. This is for comparison of the first three rings against the final design, which at this pressure is able to produce the necessary force to meet the criteria for resistance force of 25.7 N.

It should be noted that the forces of this table are derived from the area of the opening on the bottom of the rings as a projection onto a flat surface. These calculations would then be for the rings as they approach the heart. Once the ring comes into contact with the heart and a seal can be formed, the vacuum pressure then acts on the surface area of the heart that the ring covers. As the vacuum pulls the heart tissue into the ring, there is more heart muscle tissue surface area to act upon and therefore the holding force of the ring will increase. Therefore, the maximum force that the rings can deliver is proportional to their entire inner surface area (representing the maximum tissue surface area), which will
Figure 4.20: Cardiac stabilizer designs (dimensions in mm)
be the product of the ring’s projected surface area onto a flat plane of contact and the channel height. It must be noted though, that to attain the maximum holding force would depend upon infinitely compliant heart tissue to completely fill the corners of the device.

A problem with compliant tissue is its propensity for cutting off of the vacuum flow around the ring. Should the vacuum flow to tissue be cut off, that tissue will be temporarily freed from the grip of the vacuum, will be pulled away by the contraction of the heart and could eventually lead to total separation.

From these last two paragraphs and from the discussion of stabilizer design constraints, it is noticed that the height of the vacuum channel was a major design concern. With tall rings there is good hold and good visibility in the grafting site due to inhalation, but more potential for damage to the tissue and a greater reduction of the surgeon’s workspace due to the necessarily higher angle of attack for the suturing instruments. With short rings visibility and grip are reduced, but there is less potential for damage to the heart and less interference with the surgeon’s work. To explore this design compromise, after the proof of concept O-ring was constructed and before the final prototype was created, the mesh and grooved rings were created.
Table 4.8: Comparison of dimensions and performance for the stabilizer models (dimensions are in mm for lengths, mm² for areas and N for force)

<table>
<thead>
<tr>
<th>Model</th>
<th>Diameter</th>
<th>Overall Height</th>
<th>Channel Height</th>
<th>Channel Width</th>
<th>Channel Area</th>
<th>Theoretical Hold Force</th>
<th>Performance Assessment</th>
</tr>
</thead>
<tbody>
<tr>
<td>O-Ring</td>
<td>50</td>
<td>6</td>
<td>4.5</td>
<td>5</td>
<td>675</td>
<td>25.7</td>
<td>excellent</td>
</tr>
<tr>
<td>Grooved</td>
<td>42</td>
<td>4</td>
<td>3</td>
<td>10</td>
<td>966</td>
<td>36.8</td>
<td>failed</td>
</tr>
<tr>
<td>Mesh</td>
<td>42</td>
<td>4</td>
<td>3</td>
<td>10</td>
<td>465</td>
<td>17.7</td>
<td>failed</td>
</tr>
<tr>
<td>C-Ring</td>
<td>42</td>
<td>4</td>
<td>3</td>
<td>9</td>
<td>675</td>
<td>25.7</td>
<td>very good</td>
</tr>
</tbody>
</table>

4.2.1 Proof of Concept

The initial ring was designed not for arterial grafting, but merely for proof of concept. As such, it had an unbroken “o” shape. It was made with a diameter of 50 mm and height of 6 mm. Its vacuum channel was 4.5 mm high and 5 mm wide, with an overall area of vacuum application (flat, before inhalation of tissue) of 675 mm². These dimensions were based on an educated guess of what would be a good size for the ring based on a heart model approximately the size two fists placed together [16]. The O-Ring can be seen in Figure 4.20 a).

As can be seen in the figure, the the vacuum delivery is through a handle bent at 20° to the plane of vacuum delivery. This was to remove it from the beating heart - reducing the danger of being struck and dislodged, and to make it easier to attach to the heart in the crowded thoracic cavity.

The channel width was made fairly thin to provide a good deal of room for suturing inside.

4.2.2 Further Experimentation

The first design was made with a relatively tall profile in order to provide substantial hold to learn whether or not the concept of stabilization was effective. Once this ring was seen to work it was desired to have a ring that allowed even more room for the surgeon to work in and allowed for the grafting of arteries. With the stabilized grafting of arteries, there is a greater likelihood of the heart pulling away from the ring as the tissue inhaled can only be approached from the side closest to the vacuum inlet with vacuum pressure. Should vacuum be cut off to some tissue in a complete ring (a solid “o” shape) it could still be gripped by the vacuum on the opposite side.

As an experiment to see if the integrity of the vacuum flow throughout a broken ring and its hold could be maintained with a low profile to limit obtrusiveness, the grooved and mesh ring designs were constructed. These are shown in Figure 4.20 b) and c). These would determine as well, the impact that
the amount of tissue inhaled by the stabilizer would have on the overall hold.

The rings were made with a square shape due to the complexity of the design. Even though this would hamper the rings' ability to grip round objects, this was deemed easier to machine and as these rings were merely intended for in the lab use and not for hospital application, they were constructed in this fashion.

The grooved version is named for the grooves cut into its underside. These create vertical flanges preventing the inhalation of cardiac tissue and thereby the blocking of the vacuum channel. To determine the width and height of the grooves in this design, the previous successful O-ring design was examined. It was noticed with the O-ring that there was no vacuum blockage and its channel dimensions were 4.5 mm in height by 5 mm wide. Taking this ratio, \( \frac{h}{w} \geq \frac{4.5}{5} = 0.9 \) and assuming that the compliance of the heart would be constant with a small change in dimension, it was assumed that a similar ratio could be applied to the new ring. The minimum gap width permitted for easy machining in our workshop was 3 mm thus a height of 2.7 mm should have been adequate. This was rounded up to 3 mm.

The mesh version attempted to solve its tissue inhalation problem by imposing a wire mesh between the heart and the vacuum channel. This effectively reduced the area available according to the gauge of the mesh used. Here the gauge was conservatively estimated to be such that the total available area was halved.

From the calculations presented in Table 4.8, the grooved ring should have been able to easily hold a beating heart. However, both it and the meshed ring (which at the low pressure of 38.1 kPa, did not meet the criteria for hold), due to the limited extent to which the rings allowed tissue to be inhaled, had very weak hold and were unable to maintain a seal. That the weak hold was due to limited tissue inhalation was proved by increasing the interior chamber depth by attaching 1.5 mm thickness rubber to the bottom of the stabilizer. This increased height gave enough hold for gripping to the palm of a hand and other compliant objects.

### 4.2.3 The Final Prototype Stage

With the experimentation complete, the final ring design was constructed. This was the C-ring. This ring was similar to the O-ring in general shape and area, so as to maintain performance and approximate hold power, but it had a "C" shape with a 10 mm gap to permit the passage of completed arterial grafts. To make this ring as small as possible the inner radius was set to 10 mm according to the constraints, resulting in a 60° wedge removed from the ring. The outer radius was then found to achieve the O-ring's
area, as 21 mm. The dimensions and ring shape can be seen in Figure 4.20 d).

By these dimensions, the shape and the function of this ring, the C-ring was seen to be in accord with the design constraints. It fit to the heart; it had enough room within it to permit a graft to take place; it had a broken shape to permit venous and arterial grafts to take place; it could maintain a bloodless field; it could hold the myocardial tissue stable enough to allow for the anastomosis.

### 4.2.4 Testing the Stabilizer

This ability to stabilize the heart and maintain a bloodless field was confirmed in testing. A veinous graft was made to the LAD artery on a pig using the ring. The operation proceeded quickly and efficiently. The thoracic cavity was opened and myocardium exposed. The rod and snug arm (the more rigid and easier to apply pneumatic arm was not developed at this stage (these arms are described in Chapter 5)) and the C-ring were assembled and the ring was applied about the grafting site, as shown in Figure 4.21. There, the C-ring can be seen resting on the pig’s heart, flanked by the blades of the chest retractor.

![Figure 4.21: Applying the C-ring (with experimental arm) for grafting](image)

With the site immobilized the graft was made while the heart continued beating. The final result of this is shown in Figure 4.22 where the newly attached vein can be seen as the dark tubular shape between the retractor blades, being held by a pair of surgical pliers.

With this design, there is still a concern that vacuum flow can be more easily cutoff than in the O-ring as the vacuum flow can only approach tissue from the one direction in the broken ring. To alleviate this
then, the C-ring to be integrated with the surgical arm has been made slightly deeper than this design.

4.3 Future Designs

Future work on the ring could include enhancements to the shape and size of the ring and the possibility of added control to the vacuum pressure provided.

4.3.1 Future Stabilizer Shape

As enhancements to the designs discussed above, other shapes could be pursued to reduce the cardiac stabilization device’s size and improve its performance, to allow it greater application in other surgeries, and to reduce potential for bruising.

Two ideas to reduce the device size and improve performance were to construct the ring in a colosseum shape or as an ellipse. The colosseum shaped ring would have its inner wall (that wall immediately encircling the grafting site) shorter than the outer wall giving it the appearance of a stadium. The lower wall would provide reduced interference with the surgeon’s work, allowing them to approach the site from an angle defined by the slope between the inner and outer walls, and not from directly overhead. The higher outer wall would provide substantial height so as to maintain the flow of vacuum around the ring while creating space for the compliant heart to be pulled into to ensure a solid grip on the
myocardium. This could also be done by constructing a "C"-shaped ring with a vacuum channel cross-
section resembling a thin rectangle (lying on its longer side) with a small square upon it at the corner
of the rectangle at the outer radius of the ring. The square would have dimension small enough and
be positioned such that no tissue could be inhaled into it. Thus this ring would perform essentially the
same as the C-ring does, but with the square to provide assurance that the vacuum flow would not be
cut-off.

As another idea for a broken ring that would allow CABG operations with the mammary arteries,
a spiral ring was considered. For the spiral, the radii (inner and outer radii of the ring walls) would
expand linearly with angle $\theta$ from its closest arm to its endpoint. At $\theta = 2\pi$ the radii would be larger
by the thickness of the mammary artery space, $t$, plus the difference between the two radii, $h$. The
expanding spiral would allow for complete encirclement of the anastomosis site similar to the O-ring
with extra overlap, while permitting mammary artery grafts as the C-ring does. The problem with this
design is that the expanding radii quickly lead to a larger than desired ring size. This makes it much
more difficult to apply to the curved surface of the heart and have hold. Either a much higher vacuum
to force the heart to contort to the ring or a flexible spiral to conform to the surface of the heart would
be required.

In an effort to get around the problem of the increasing size with expanding radius, the spiral could
instead be made as an ellipse. This spiral design is essentially the same as the spiral except that instead
of expanding linearly its expansion would be based on a sinusoid. Thus, a full $2\pi + n$, $n > 0$ radian
expansion could be had in a much more compact shape by concentrating the expansion about the axis
perpendicular to the axis of origin. This is shown in Figure B.31.

To aid application in other surgeries, such as minimally invasive or port access surgeries, the ring im-
plemented in the form of multiple smaller stabilizers connected to the same vacuum source. Application
would be as done as it is now with the ring, but the shape would more likely be sausaged shaped.

Finally, bruising to the tissue could be reduced by rounding the inside edges to the ring. Currently
90° angles are there now which create a significant increase in local pressures on the tissue at the edge.
Should material be added such that a more rounded surface could be presented, the local pressures would
be decreased. However, this may not be an important consideration as future designs may be made out
of a disposable plastic that would be compliant enough that the local pressures would not be a problem.
4.3.2 Control

Future enhancements of the vacuum delivery system could include the addition of control for the vacuum pressure delivered to the ring. Vacuum levels could be made to adapt such that during the systolic period when contraction occurs and the heart forces and torques pulling the heart away from the ring are at their maxima, the vacuum is at its strongest level. Similar to the strobe’s operation, the EKG signal could be used to trigger the rise and fall of suction at the appropriate times. This could help reduce the overall vacuum forces on the myocardial tissue reducing the tissue damage that takes place.

In another similar yet more elaborate implementation, the ring could have several separate vacuum channels that would be turned on and off cyclically. This would provide the necessary force to hold the heart steady while not remaining attached to one spot for a long period of time and thereby reducing damage caused through continuous application. However, this approach would increase the obtrusiveness, complexity and cost as more control circuitry would need be devised and multiple pumps or complex controlled mechanical slides and manifolds and valves used.
Chapter 5

Surgical Arm Development

To hold the stabilizer fixated to the heart and form a rigid link between it and the patient, a surgical arm is required. However, as stated in the introduction, the solutions of the current stabilizers were too time consuming in application and the available arms were judged to be too cumbersome, too bulky and too heavy for the delicate and dexterity demanding work of coronary artery surgery. The goal of this research, then, was to design a surgical arm that could perform better than previously designed surgical arms for application to CABG surgery.

5.1 Surgical Arm Design Constraints

The constraints enabling evaluation of the arms performance were categorized and defined as follows:

Rigidity When in operation, the arm must be able to hold the cardiac stabilizer fixated while being pushed against by a large heart. When not in operation, the arm must be compliant so that the surgeon may easily position it.

Speed of activation The arm must fixate nearly instantaneously to save surgeon time and effort.

Safety The arm must never be a threat to either the staff or patients, even if power fails.

Unobtrusiveness The arm must not be a hindrance to the surgeons’ efforts.

Maneuverability The arm must be sufficiently dextrous to allow positioning in the small and awkward space of the thoracic cavity.

Weight The arm must be light.

Cost The arm must be affordable.

Of note in these constraints and worthy of further explanation is the weight constraint. Whereas traditional surgical arms are generally very heavy objects mounted to operating room tables capable of supporting them, the arm here was to mounted directly to the patient on their retractor, providing a mechanical ground between the arm and the patient. This was to eliminate worries of patient body size affecting the size and positioning of the arm and to remove a major source of vibration and potential
Chapter 5. Surgical Arm Development

harm to the patient, as the patient’s movement would be coupled to the surgical arm. Patient motion is a problem should there be aspiration of material or fluids during the surgery and coughing as a result.

For the arm designs that were judged to perform well enough on paper to merit construction, the maneuverability of the arm was first determined as a check on the arms reachable workspace. Then, calculations were performed to determine the necessary braking torques at the joints to resist the end effector forces and torques rising from the heart’s beating and the gravity loading of the arm’s individual components.

5.2 Design Alternatives

Several designs were postulated, analysed against these constraints and judged according to their potential merits and shortfalls before decision was made as to the design of the final arm. The design of the arms centred around the problems of braking and how to make the arm rigid enough that it could immobilize the heart.

Some of the ideas that were eliminated quickly included prismatic jointed arms, arms using quick release type braking mechanisms, wire driven cantilever braking methods, vacuum braking, and fluid power braking. Prismatic jointed arms would be simple but would demand a great deal of material for joint extension. This would result in a heavy design. Quick release type braking mechanisms, which would demand manual manipulation at each joint, could not be used for lack of instantaneous immobilization. Wire driven cantilever brakes are problematic in that they demand maintenance to counter cable stretch, and due to the potential for catastrophic failure in the form of sudden snapping of the cables. Vacuum braking would demand the use of another separate vacuum line or pump in the OR due to the differing pressure level needs and time of application. As well, the absolute maximum vacuum pressure is 101.3 KPa, which is small when compared to possible pneumatic pressures. In order to get comparable vacuum braking forces the surfaces of contact would have to be greatly enlarged which would be impossible in the thoracic cavity. Fluid power would perform adequately but has the danger of leaking non-sterile fluids into the surgical field.

The surgical arm alternatives that were considered seriously enough to go through a design stage were: the rod and snug experimental arm, the flexible arm, the electromagnetic arm, and the pneumatic and pistonless pneumatic arms. Depictions of various aspects of these designs are displayed in Figures 5.23 and 5.24. A comparison of these arms according to the constraints outlined in Section 5.1 is made in
Chapter 5. Surgical Arm Development

5. Surgical Arm Development

a) Universal Snug Joint for use in Rod and Snug Experimental Arm

b) The Flexible Arm (without end effector)

c) The Electromagnetically Braked Arm (Stepper Motor Arm is similar)

Figure 5.23: Arm design alternatives (page 1)
Chapter 5. Surgical Arm Development

56

braking segment

Figure 5.24: Arm design alternatives (page 2)

56
Table 5.9.

Table 5.9: Qualitative comparison of the arm design alternatives

<table>
<thead>
<tr>
<th>Arm Model</th>
<th>Rigidity</th>
<th>Activation Speed</th>
<th>Safety</th>
<th>Obtrusiveness</th>
<th>Maneuverability</th>
<th>Weight</th>
<th>Cost</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rod and Snug</td>
<td>moderate</td>
<td>poor</td>
<td>excellent</td>
<td>excellent</td>
<td>good</td>
<td>good</td>
<td>good</td>
</tr>
<tr>
<td>Flexible</td>
<td>poor</td>
<td>good</td>
<td>good</td>
<td>very good</td>
<td>good</td>
<td>good</td>
<td>excellent</td>
</tr>
<tr>
<td>Electromagnetic</td>
<td>very good</td>
<td>excellent</td>
<td>poor</td>
<td>poor</td>
<td>good</td>
<td>poor</td>
<td>moderate</td>
</tr>
<tr>
<td>Pneumatic</td>
<td>excellent</td>
<td>very good</td>
<td>very good</td>
<td>very good</td>
<td>very good</td>
<td>very good</td>
<td>very good</td>
</tr>
</tbody>
</table>

For a more quantitative measure of the braking methods being used, joint torques were measured for some of the designs made. Torque testing of the joints was made using the force-torque sensor described in Section 4.1.1. It would be attached to the arm or arm segment and then rotated about each axis. The maximum resistance was found at the point of motion, when friction changes from static to dynamic. At this point there is a reduction in the coefficient of friction and a corresponding decrease in the force of friction and hence, the resistance torque at the joint. The numerical results are presented in the individual arm design sections.

From the comparison of the chart, it can be seen that the arm that would perform best under these constraints is the pneumatic arm. As such, this arm was selected to be developed into a full prototype. Details of its design process and performance against the constraints are discussed in Section 5.2.4. The other arm designs are discussed in the following sections, starting at Section 5.2.1.

5.2.1 The Rod and Snug Experimental Arm

Initially, the arm used for testing the stabilizing concept was composed of four stainless steel rods for links and used four hand-tightening universal snugs for connection and braking between the links. The snugs allowed for translation and rotation for each link at the snug. A diagram of a typical joint and the possible motions is shown in Figure 5.23 a).

The thinness of the rods (9.5 mm diameter) and the redundant nature of the snugs made the arm very maneuverable and non-obtrusive. The small amount of materials and simple construction also made the arm cheap and lightweight. Thus it was good for experimental purposes. However, it was lacking in that application was tenuous: when a rigid arm was needed each joint had to be tightened by hand separately. Thus despite its benefits, it was not suitable to be used as the final arm.
5.2.2 The Flexible Arm

The flexible arm was composed of a plastic machinist's cutting fluid delivery hose fitted to the bottom of a metal link. On the top of this same link, vacuum would be supplied through the arm directly to the cardiac stabilizer. The cutting fluid hose itself is a hose made of several connected and independently adjustable and interlocking components. It is similar in appearance to the cable clamps used to hold the Guidant and Medtronic stabilizers discussed in Section 1.2.2.

The inter-joint braking strength is based on friction from the tight fit between each joint. Each joint allows for three DOF, although due to the degree of attachment to the preceding ball, these DOF are restricted in their motion. An advantage to this arm is that these constraints can easily be overcome by simple insertion of more pieces into the arm. The arm that was tested consisted of eleven ball joints plus a nozzle and an attachment piece. This arm is pictured in Figure 5.23 b).

Because the arm was hollow and airtight, it could deliver vacuum directly through its interior to a stabilizer attached to the nozzle. This made for a very unobtrusive design. Being made out of plastic, the arm was extremely light. It and its mounting link weighed 118.4 g. Its flexibility and redundant DOF made it extremely maneuverable. Finally, being non-powered it did not have any of the safety problems associated with power failure.

Despite all this, the flexible arm was dismissed due to failure in rigidity. When tested with the force-torque sensor, the flexible arm was seen to hold well most of the time, but with small impacts (similar to those it would receive from the beating of a heart) it would gradually be nudged out of position. Once the data captured was adjusted for position and orientation relative to the centroid of the joint about which the forces and torques were being measured, it was found that the joints could at most resist 1.07 Nm about the $x$ and $y$ axes, and 0.428 Nm about the $z$ axes. The average values of the maximum torques recorded on six joints was 0.409 Nm about the $x$ and $y$ axes, and 0.254 Nm about the $z$ axes. As a back of the envelope approximation of the torque requirements of the arm, the $x$ or $y$ axis torque appearing at the base joint due to the force requirements of the stabilizer (25.7 N, see Section 4.1.1) being delivered through a torque arm of 11.7 cm (the distance to the centroid of the model heart, see Section 5.2.4) is 3.01 Nm. This is nearly triple the capabilities of the flexible arm joints.

If it was decided to be worth pursuing, a flexible hose with larger ball joints might be able to handle these torque loads. The larger joints would theoretically provide a larger area of contact and more friction through it. Due to the light weight of the plastic and the thinness of the arm this increase in
size and weight would not be a problem.

5.2.3 The Electromagnetic Arm

As another alternative for braking, electromagnetic braking was considered. Under this design a three-link arm composed of stainless steel links connected by electromagnetic shaft brakes would hold a three-DOF wrist connecting the end effector with the heart. The shaft brakes are meant to stop the rotation of a rotating shaft when applied. In this case, application of the brakes occurs when the electricity to the brakes is cut off; the electric current turns on magnets that pull the braking material away from the shaft. This leads to a safer design in case of power failure. The electromagnetic arm is shown in Figure 5.23 c) in an overall system view and with a view of a typical joint. An electromagnetic wrist for this arm was to be made consisting of a ferromagnetic or otherwise magnetized ball fixated by magnets surrounding it.

From API Deltran, the company making the best brake found in terms of compactness and strength, the three brakes in the arm could be made from 500 g, 72.9 mm outer diameter, 31 mm high servo brakes. These would be capable of resisting 4.52 Nm of torque, which would allow for resistance of the torque requirement calculation of the previous section, Section 5.2.2, 3.01 Nm with some extra capability to handle the mass of the links and the brakes. These brakes have a 40 msec activation time and consume 17.7 W of power [34]. An arm with these brakes would be very quick to activate and very rigid. The arm should also be maneuverable as this configuration allows it a full six DOF.

Although the electromagnetic arm would be free from the danger of collapse during power failures, its overall level of safety would still be questionable. The large amount of power that the servo brakes and an electromagnetically braked spherical joint at the wrist demand has potential for danger, particularly if the arm became wet. This electric power consumption also creates a great deal of heat which will need to be dissipated. Finally, the wires to deliver this power complicate matters by making the arms difficult to sterilize and more obtrusive.

Proper shielding to reduce the danger of electric shock and also for the prevention of arm induced electromagnetic radiation affecting the measurements or operation of the operating room electronics would be necessary. This would increase obtrusiveness and weight for the arm, making it unacceptable for chest mounting. The brakes on the links alone would weigh 1.5 kg. Added to this would be the wrist, end effector, and connecting links, plus the shielding. These could easily push the mass above 3 kg, depending on the dimensions chosen and the exact design of the wrist.
For the reasons of safety, obtrusiveness and weight, it was decided not to pursue this design further.

The Stepper Motor Braking Arm

A similar arm to the electromagnetic arm is the stepper motor arm. It would be constructed in the same manner as the electromagnetic arm except that the electromagnetic brakes would be replaced with stepper motors. Stepper motors are generally used for discrete placement tasks but can be made to operate as brakes by energizing their coils with a constant pattern.

The advantages of this design over the electromagnetic design and some of the others, are cost and controllability. The cost of stepper motors range from US$45 to $100. Control could be easily implemented for automatic positioning or for disturbance recovery: if the arm was bumped and moved, the joints that were altered could be quickly returned to their original position.

However, this design still has the shortcomings of the electromagnetic arm and a few more. Stepper motors move with discrete steps, which will not permit exact positioning. However, this may be inconsequential with several DOF and step sizes as small as 0.025°; stepper motors are also fairly weak in power, generally having typical holding powers from 0.16 to 1.41 Nm. Finally, the stepper motors cannot be implemented into an arm design because their braking is dependent upon electricity. Should there be a power failure, the stepper motor arm would collapse into the patient's chest cavity.

5.2.4 The Pneumatic Arm

The pneumatic arm and its off-shoot, the pistonless pneumatic arm relied on pneumatic pressure to push surfaces together to create the braking force and rigidity required.

The pneumatic arm is a seven link design - the seventh being the end effector and cardiac stabilizer link - using pneumatic power for fixation. The odd links (links one, three and five) are stainless steel rods for connection to the retractor and interconnection between the other three segments. These other segments (links two, four and six) are the links that perform the braking to make the arm rigid. This overall system view can be seen in Figure 5.24 f). Each braking segment is composed of sleeves which thread together and seal to contain ball bearing and piston ball-and-socket joints at each end. The sleeves are made of dissimilar non-freezing metals to avoid locking up without the use of lubricant. The pistons are sliding cylinders that have had their surfaces facing the ball bearings at the end of the segments hollowed so that they can form a ball and socket joint with the ball bearing. A teflon ring and O-ring on the perimeter of the pistons enable sliding while providing for the appropriate seal. A view of
these components and their relation to each other can be seen in Figure 5.24 d).

As stated, the arm locks in position when pneumatic pressure is delivered to the interior of the braking segments. The pressure forces pistons into the ball joints pinning them against the outer sleeves. Pressure control is delivered separately to each braking segment.

The ball bearings at each end of the braking segments act as three DOF joints, giving the arm eighteen-DOF. This redundancy is necessary as the sleeve material holding the ball bearings in place restricts pitch and yaw rotations to ±32.5°. To further increase the maneuverability of the arm, the connecting rods were bent 45°.

Quantifying Maneuverability

To quantify maneuverability for the pneumatic arm designs, the inverse kinematics had to be calculated for all positions on a heart model and shown that it was possible for the arm to orient its end effector so as to grip in these locations. This is to allow full applicability to all the coronary arteries.

The model used was composed of the thoracic cavity, heart and arm.

The thoracic cavity was modeled as a simple open rectangular box whose width and length were estimated from a spread retractor to be sixteen and 30 cm, respectively, and whose depth was estimated during a CABG surgery to be 18 cm.

<table>
<thead>
<tr>
<th>Table 5.10: Typical heart dimensions (from [32])</th>
</tr>
</thead>
<tbody>
<tr>
<td>Female</td>
</tr>
<tr>
<td>----------------</td>
</tr>
<tr>
<td>Length (cm)</td>
</tr>
<tr>
<td>Transverse Diameter (cm)</td>
</tr>
<tr>
<td>Antero-Posterior Diameter (cm)</td>
</tr>
</tbody>
</table>

The heart was modeled as an ellipsoid with axis lengths based on the maximum dimensions [32] shown in Table 5.10, enlarged by 25% as a design safety factor. For calculation purposes, the ellipsoid was discretized into 290 points derived from 20° interval samples on vertical plane sections taken along its length.

These models were then implemented into a computer modeling program so that correct positioning of the ring on the heart model could be visually verified. The program consisted of two main sections of code, the first for visual verification of workspace and the second for calculation of torque requirements for the joints.
The first section was a graphical user interface encoded with the VOGL graphical language and C. This code provided a visual description of the arm and its collisions with the heart and thoracic cavity models. The description was provided as a wireframe view of the arm under test, and the thoracic cavity and heart models. An example of this is the wireframe model constructed for the pistonless pneumatic arm shown in Figure 5.25. Quantifying maneuverability was done here by verifying that the end effector of the arm under test could be placed on each target on the digitized heart without interfering with itself or the thoracic cavity wall. However, perfect tangential placement of the end effector on the heart was not always achieved. Instead the end effector centroid was only required to touch the heart targets, meaning that for some targets there would be perfect contact with the heart and at other targets the forward edge of the end effector could penetrate the heart ellipsoid.

![Wire-frame model of the stabilizer, surgical arm, heart and thoracic cavity](image)

Figure 5.25: Wire-frame model of the stabilizer, surgical arm, heart and thoracic cavity

While graphically iterating through these heart model positions, the code also performed calculations of the torques that would appear at each of the joints due to end effector and gravity forces and torques. This was to provide an estimate of the maximum forces and torques that the arm joints would be subjected to for arm joint brake performance assessment. This last function will be described next in Section 5.2.4.
Quantifying Rigidity for the Pneumatic Arm

To quantify the required holding torques at the joints, the end effector forces and torques and gravity loading of the arm components had to be estimated and then mapped to torques at the arm joint axes.

The end effector loading was derived from maximum values for the scaled heart force and torque measurements obtained in the end effector frame, as described in Section 4.1.1. The maxima were used to provide worst case loading values. Their values are displayed in Tables 4.6 and 4.7.

This was repeated for every point on the heart model for which the inverse kinematics portion of the code found a solution. However, in the calculations for the pneumatic arm this was only a subset of all the possible inverse kinematics solutions. Instead of exploring fully the redundant solutions of the arm, this program instead only found one solution per point. Because of the small subset of solutions, the torque requirements for each joint will not be accurately represented. A much more exhaustive approach was used for determining the brake torques for the pistonless pneumatic arm. It will be described in Section 5.2.5.

The solution for the pneumatic arm was based on a direct approach to the target by the arm as follows:

1. The arm was given a roll rotation at the first joint to align the end effector with the heart target.
2. The endpoint was iteratively lowered, by adjusting the yaw angle for each joint, one joint at a time. This continued until the end point was within a threshold value of the target or until the point when it was realized that the arm could not connect with the target.
3. For the target points that the arm could not connect with, the point location was stored for further reference and the program moved on to attempting to connect with the next target.
4. For all points where the connection to the heart model was made:
   (a) The resultant joint angles of this iterative inverse kinematics were used to determine the Jacobian transformation matrix.
   (b) The Jacobian was then used to map the heart forces and torques, and gravity loads to each joint according to the transformation,

\[ \tau = J^T F \]  \hspace{1cm} (5.11)

where \( J \) is a \( 6 \times 18 \) matrix and \( F \) is the applied end-effector wrench.
Chapter 5. Surgical Arm Development

For this process, maximum values of 1.21 Nm for roll (about the z axis), 2.22 Nm for pitch (about the y axis), and 1.81 Nm for yaw (about the x axis) were predicted for the pneumatic arm. The full results of the brake torque calculations follow in Table 5.11. Maximum and minimum torques calculated while attaching to all targets on the heart model are presented with the joints numbered in ascending order from the base of the arm to its end effector.

Table 5.11: Maximum and minimum braking torques (in Nm) calculated for the pneumatic arm

<table>
<thead>
<tr>
<th>Joint</th>
<th>Roll (z)</th>
<th>Pitch (y)</th>
<th>Yaw (x)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.505, -0.455</td>
<td>2.216, -0.676</td>
<td>1.684, -0.422</td>
</tr>
<tr>
<td>2</td>
<td>1.176, -0.497</td>
<td>2.208, -0.631</td>
<td>1.812, -0.422</td>
</tr>
<tr>
<td>3</td>
<td>0.875, -0.491</td>
<td>1.637, -0.363</td>
<td>1.016, -1.088</td>
</tr>
<tr>
<td>4</td>
<td>1.087, -1.035</td>
<td>1.544, -0.357</td>
<td>0.783, -0.582</td>
</tr>
<tr>
<td>5</td>
<td>1.059, -1.117</td>
<td>1.458, -0.305</td>
<td>0.087, -0.857</td>
</tr>
<tr>
<td>6</td>
<td>0.962, -1.206</td>
<td>0, -0.980</td>
<td>1.458, -0.305</td>
</tr>
</tbody>
</table>

Pneumatic Arm Joint Braking Capabilities

Theoretical values for the capability of the pneumatic arm ball-socket joints were calculated to compare with the demands for joint torques that the end effector and component gravity forces would create.

The amount of resistance torque that can be provided by the piston-ball is based on the pressure derived friction forces on the surface of contact. The torques are calculated by summing the pressure forces acting on infinitessimal areas over the surfaces of contact. These surfaces are shown as highlighted edges in Figure 5.26.

![Figure 5.26: Braking calculations with reduced piston](image)

For the surface of contact between the piston and ball there will be contact over the range of \( \phi = \{0, \pi/2 - \alpha\} \) where \( \alpha \) is the angular range of material removed from the lip of the piston’s socket. This
material is removed to allow contact to take place without the piston jamming against the sleeve. This can be seen in the diagram of Figure 5.26, where a amount of material has been cut away from the edge of the piston socket. Taking this missing section of piston into account, the area of surface contact is reduced by

\[
\alpha = \sin^{-1} \frac{a}{r}
\]  

(5.12)

In this design, 3.1 mm was removed from the cup, and the ball radius was 12.7 mm. Therefore \( \alpha = 14.4^\circ = 0.253 \text{rad} \). As a result, the first surface of contact extends through an angular range of \( \{0, 1.32\} \text{ radians} \).

As seen in Figure 5.26 the material overhang having contact with the ball will be approximately 52.1°. Therefore, the second surface of contact, between the steel ball and the outer shell, will have a range of application of \( \phi = 0.661, \pi/2 \text{ radians} \). The angle 0.661 rad is the sum of the angles for rotation constraints due to the material holding the ball in place (32.5°) and due to the connecting rods (53.8°).

Due to the different orientation of the axes (z axis as roll, y axis as pitch and x axis as yaw), two separate calculations are required for determining the resistance torque due to these contact surfaces: torque about the z axis, and torque about the x and y axes. x and y are lumped together by symmetry.

In summary of the calculations made, the brake torque produced about the z axis is:

\[
\tau_z = \mu P^2 \pi r^3 \left[ \frac{\phi}{2} - \frac{\sin(2\phi)}{4} \right]^b_a
\]  

(5.13)

and the brake torque produced about the x and y axes will be:

\[
\tau = 4\mu P r^3 [-\cos\phi]^b_a
\]  

(5.14)

The details of the calculations are presented in Appendix C.

Then, for a pressure of 610 kPa (a conservative number as hospital nitrogen lines are capable of over 800 Kpa), a coefficient of friction of 0.8 for dry steel on steel [35], the resistance torques are calculated to be 3.01 Nm about the z and y axes, and 2.62 Nm about the z axis for the cup and socket surface of contact. For the same parameters of \( \mu, P \) and \( r \), then the contribution to overall resistance torque of the second surface will be 4.38 Nm about the z axis and 3.16 Nm about the x and y axes. Therefore, the overall calculated resistance torque for the pneumatic joint will be 7.00 Nm about the z axis and 6.17 Nm about the x and y axes.
Pneumatic Arm Assessment

From the above Sections 5.2.4 and 5.2.4, it can be seen that the pneumatic arm design was able to handle the torque loads demanded of its joints. Maximum values produced by the forces and torques applied were found to be 1.21 Nm for roll and 2.22 Nm for pitch and yaw. The overall capabilities of the arm were calculated to be 7.00 Nm in roll and 6.17 Nm in pitch and yaw.

For the constraints of speed of application, safety, obtrusiveness and weight, the arm was also seen to perform well. Speed of application is nearly instantaneous as it only depends upon the pressurization of the small volume supplied by wide piping. The arm is unobtrusive due to its thinness, being only 3.18 cm in diameter over the length of the segments excepting where the air attachments are. This small size and the use of air as the braking force allows for a lightweight arm. In total it has a mass of 1.92 kg.

In matters of safety the pneumatic arm is quite safe. Two options for handling power failure exist according to the configuration of the valves. In the event of a power failure, the required power necessary to switch the valves on and off will not be available and the valves will revert to their non-powered state. Should the arm be configured such that the valves are open to allow pressurization, the arm will collapse into the patient. As the arm is light, small and designed without sharp edges, the potential for this to cause harm is minimal. Should the arm be configured such that the valves must open to depressurize, the arm will remain rigid. The arm will not fall into the patient immediately, but very gradually as the overall pressure in the hospital’s compressor slowly leaks out.

5.2.5 The Pistonless Pneumatic Arm

With the favourable performance of the pneumatic arm design against the surgical arm constraints, a prototype was constructed. However, despite the fact that the design exceeded its criteria, the constructed arm was unable to maintain its full weight. The problem seemed to lie with the assumption of ideal contact between the ball and socket in the calculations. This is a faulty assumption as it ignores the possibility of dirt and other particles coming between the surfaces of contact and the inaccurate tolerances in the machining of the balls and pistons. Instead of the ideal contact, there was only a couple of points or a single ring of contact, as a result. Therefore, the necessary braking force could not be produced.

As the pneumatic powered braking concept was still highly desirable for its low weight, variations on it were pursued to attempt to correct for the lack of reliable braking power. Rubber coating the balls
and sockets was attempted but a solution could not be found that would resist wear. Wear results in high maintenance for the contact surfaces as well as the possibility of catastrophic failure due to pieces of material coming free and jamming between the ball and the sleeve, permanently freezing the joint. Roughening the sockets was also attempted through sandblasting but this was found to produce only a limited improvement, as the surfaces would soon be polished smooth again with use.

The next concepts were for replacement of the steel pistons with rubber in the form of rubber pistons and internal air bladders. However, difficulty in rubber shaping prevented their construction.

Instead, what was implemented was a completely pistonless design. The arm would remain essentially the same, except now there would be only the one contact surface, between the balls and the shells. To maintain hold, then, the coefficient of friction between the surfaces was increased by replacing the steel ball bearings with half stainless steel, half rubber balls. The stainless steel half provided a solid anchoring point for the connecting rod and the rubber provided a high friction contact surface to the outer sleeve. A cutaway view of this arm showing this can be seen in Figure 5.24 g). Its system view is the same as the pneumatic arm's and can be seen in Figure 5.24 h).

After experimentation with various durometer rubbers, urethane rubber was selected to be used. It is a vulcanized, thermoplastic polymer with a Rockwell hardness of 70 (to prevent yielding under the pressure this had to be high) and 100% strain value of 7.0 MPa - indicating approximately 9% strain under normal usage, and has the convenience of being able to be poured and cured at room temperature. It is also easily sterilized, being autoclavable.

By removing the pistons, the connection rods could be made hollow and air could be piped throughout the arm. Thus, the arm became a “stainless steel balloon”. Since only one connection for air was required, bulk and weight could be reduced by removing the extra air fittings. Furthermore, weight and obtrusiveness decreased again as the segments could now be shortened. This reduction in weight justified a solid stainless steel construction, creating a more robust arm and one with no chance of corrosion and joint lockup due to junctions of dissimilar metals. The pneumatic arm has the risk of lockup due to cross-threading between the stainless steel interior and aluminum outer shell.

The pistonless pneumatic arm braking segments were now only 6.1 cm long, shrinking overall length to less than 50 cm, including the end effector and connecting rods. As well, the connecting rods were shortened such that ball centroid to ball centroid they have a length of 7.5 cm. The maximum arm segment diameter was maintained at 3.2 cm as the balls were the same size, 2.54 cm in diameter. The new total weight of the arm is 940 g.
To verify that the arm could reach the required locations on the heart, a graphical model was constructed as was done for the pneumatic arm in Section 5.2.4. The arm design as it appeared is seen in Figure 5.25.

Quantifying Rigidity for the Pistonless Arm

The rigidity criteria for the pistonless arm was also determined in the same manner as for the pneumatic arm, by using a model for the heart, thoracic cavity and arm and by calculating the joint torques according to end effector and arm loads. However, the inverse kinematics solution set was expanded to handle the redundancy of the joints, and violation of the thoracic cavity constraints was handled through calculation and not through visual inspection. All feasible arm configurations were tested in order to obtain the highest braking torques required.

The same model for the thoracic cavity and heart were used as done in Section 5.2.4. However, for improved accuracy, the number of heart targets was increased by increasing the number of elliptical sections to make up the heart and the amount of samples per each of these elliptical sections.

The model used was a simplified version of the actual arm. To reduce the redundancy, the arm was made into a four link design by amalgamating the first braking segment with the first connecting rod after the base, and the second braking segment with the second connecting rod - in essence, this was solved by giving the third and fifth links to zero lengths. This reduced the DOF to twelve. New rotation constraints were approximated for these segments to be 82°. This included the extra maneuverability due to the bending of the connection rod.

Referring to Figure 5.27, to determine required joint torques, first the locus of the end effector joint was determined by placing the end effector tangent to the ellipsoidal heart model at discrete points. Points that violated the thoracic cavity constraints by entering the patient or by being positioned above the patient such that the target location could not be reached by the end effector link without penetrating the patient, were discarded. This resulted in a discretized surface representing an expanded heart model workspace.

Then, from the base of the arm, a discretized surface of all the possible locations for the endpoint of Link 1 was constructed. This surface was the intersection of a sphere with a vertically opening cone with frustum determined by the rotation limits.

Next, position and orientation for Link 2 and Link 3 were determined. Spheres scribed by the motion of Link 2 from each of the points on the base conic surface and scribed by the motion of Link 3 around
Figure 5.27: Joint torque determination for the pistonless pneumatic arm

Chapter 5. Surgical Arm Development

the EE joint locus were intersected. Discretized points on this intersection were tested for violation of the joint rotation limits. Valid points determined the remainder of the arm configuration.

The configuration angles were determined in order to construct the Jacobian transformation matrix, which again was used as it was for the pneumatic arm in Section 5.2.4 to translate the heart forces and torques and gravity loads to each joint through use of equation 5.11. Due to the reduced DOF, the Jacobian used here was a $6 \times 12$ matrix.

The full range of end effector forces and torques were applied for each configuration. Heart cycles of data for the scaled LV, RV and heart base force torque sensor measurements (Section 4.1.1) were used to estimate the forces throughout normal heart activity. The arm mass gravity loading was kept constant.

For the tests made using forces and torques gathered throughout the heart cycle at the heart base, LV and RV, the joint torques ranged from -4.610 and 4.052 Nm for the pitch and yaw axes and from -3.885 to 4.874 Nm for the roll. Considering the magnitudes only, the joints must be able to resist 4.87 Nm of roll, and 4.61 Nm of pitch and yaw. The breakdown of this for each ball joint appears in Table 5.12.

Braking Capabilities The braking torque of the joints for this arm were calculated in the same manner as for the pneumatic arm, as described in Section 5.2.4. Here however, there is only one surface of contact, between the ball and the outer shell, and its range of friction application is for $\phi = 0.531, \pi/2$ radians. The angle 0.531 rad is the sum of the angles for rotation constraint due to the connecting rod (5.38°) and due to the shell (25°) material holding the ball in place. Then, inserting these ranges into the
Table 5.12: Maximum and minimum braking torques (in Nm) calculated for the pistonless pneumatic arm

<table>
<thead>
<tr>
<th>Joint</th>
<th>Roll (z)</th>
<th>Pitch (y)</th>
<th>Yaw (x)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>4.638, 0.000</td>
<td>4.052, 0.000</td>
<td>3.440, 0.000</td>
</tr>
<tr>
<td>4</td>
<td>-3.008, 0.000</td>
<td>-2.142, 0.000</td>
<td>-3.249, 0.000</td>
</tr>
</tbody>
</table>

equations 5.13 and 5.14, for 12.7 mm radius balls, 610 KPa pressure, and a coefficient of friction of 1.5 (a low value for rubber on clean stainless steel [35] to provide yet another safety factor), the calculated holding torque was 8.70 Nm about the z axis, and 6.47 Nm about the x and y axes.

Comparing these values to the requirements calculated in Section 5.2.5 (absolute magnitudes of 4.87 Nm about z, and 4.61 Nm about x and y), the designed joints were seen to perform well. The resistance torques are calculated to be stronger by 3.83 Nm for roll and 1.86 Nm for pitch and yaw.

5.3 Future Surgical Arm Development

With the advantages of the pneumatic arm, plus reduced weight and obtrusiveness, construction of the arm was proceeded with. A photograph of the pistonless arm integrated with the stabilizer is shown in Figure 5.28.
From this point, the new arm design needs to be verified. Force torque testing should be performed to ensure that the arm will perform as desired. Then, once it is assured to work in the operating room, clinical trials should proceed.
The research conducted for this thesis was to improve cardiac surgery by providing methods of stabilization of the heart so as to remove the heart bypass machine from procedures. The removal of the heart bypass machine is desirable because, although it performs its required tasks of blood filtration and oxygenation well, it is known to cause considerable damage to the patient due to destruction of RBC and platelets, due to the immune response created by the contact of blood with the unfamiliar surfaces of the blood pump, and due to its non-pulsatile pressure effects on the patient's organs. To facilitate the removal of the blood pump, this thesis developed and evaluated optical and physical methods of heart stabilization.

6.1 Contributions

The contributions made by this thesis towards the solution of this problem include:

- **Optical stabilization investigation** The optical stabilization devices (Chapter 2) offered a non-contact method of heart stabilization, whereby a strobe held above the heart would flash the heart at the same point of each heart cycle. This was to give the surgeon a frozen view of the heart to allow them to suture without motion distraction.

- **Physical stabilization development** A method of stabilization, whereby the grafting location and tissue around it are held immobilized with a mechanical stabilizer, was developed. The stabilizer (Chapter 4) developed was particularly novel for its continuous application of vacuum. This gives it the advantages of being able to maintain a bloodless field, and having a strong grip at low pressures.

- **Surgical arm development** To hold the stabilizer when attached to the heart, a low weight, highly dextrous surgical arm was required. In Chapter 5, several novel designs were investigated and the pistonless pneumatic arm was shown to be the best. It is of small size, unobtrusive in the surgical field, extremely light, deployable directly to the patient's chest retractor, lockable and
unlockable at the flick of a switch, and safe.

- **Measuring and interspecies scaling of heart forces and torques** Also of note and presented in this thesis were measured forces and torques from a pig's heart, and a method for scaling these forces and torques between species.

### 6.2 Overall Results

In the optical stabilization testing chapter (Chapter 3), it was seen that the strobe was unable to improve the accuracy of the “surgeons” using it. In the human factors testing of the twenty-two subjects, it was revealed that optical stabilization is not effective due to slow reaction times; using the strobe was less effective than simply following the heart motion with your own eyes. Perhaps with a much more complex task, which would involve a reduced ability of the human subject to track the moving target, the strobe might prove effective.

The physical stabilizer had much better results. In testing, it was used to perform an end-to-side anastomosis (vein to artery attachment such that blood could be delivered) on a pig. Through application of the C-ring, a surgeon was able to operate directly on the moving heart and make a patent graft in a bloodless field. The above testing was done before the construction of the pistonless pneumatic arm, with the rod-and-snap experiment arm.

### 6.3 Future Directions

The optical stabilization technique by itself may not be applicable to non-CPB, CABG surgery, however, it could still be useful in concert with other techniques. It is proposed that computer vision based tracking of the heart could still be done with a platform implemented to follow heart's motion to provide optical stabilization of the heart and instruments for the operating surgeon. Then, the surgeon could more easily perform the suturing, as if suturing to a stable surface.

Further development will be required of the physical stabilization devices as well. The ring needs to be tested on animal subjects with the new pistonless arm in place. Before this can happen though, the pistonless pneumatic arm needs further testing in the lab for its braking capabilities and general performance. With the arm verified in the lab, testing of the complete cardiac stabilizer system will follow in animal and human trials to verify that it is a safe, effective method of providing cardiac immobilization. For future enhancement of the stabilizer, ring designs such as the colosseum and elliptical ring designs
could be implemented.

To demonstrate the effectiveness of this stabilizer system, a comparative study should be made of it with the stabilizers of CTS, Guidant and Medtronic. This study could measure objectively the effectiveness in controlling motion of the surface, the time and effort in application, and other important parameters.

The arms developed in this thesis should be further explored to determine their potential for useful application in non-CABG related problems, both surgical and non-surgical. With regards to the pistonless pneumatic arm, although the original intention for this arm was for cardiac stabilization in CABG procedures, this arm could be effectively applied to any other situation demanding a small, light, dextrous, quickly deployed and rigid arm.

Finally, more force-torque testing and data collection on multiple pig hearts should be performed to better verify and quantify the results obtained. As well, this should be performed on human subjects to verify the inter-species translation.
Appendix A

Cardiac Function

This appendix is to provide more in-depth details of heart function, cardiopulmonary bypass pump
effects, and the procedures involved in coronary artery bypass grafting.

A.1 The Heart and Circulatory System

The heart provides the primary means for circulation of blood in the human body. It is made up of
myocardial muscle tissue - special in that it contracts, not with our conscious control, but involuntarily.
It is instead regulated by baroreceptors and other sensors acting through the sympathetic and parasym­
pathetic nervous systems, the vagus nerve being the principal of these. This unconscious control allows
for continuous beating even when the person is unconscious, as would be the case during surgery. Thus,
problems arise when surgery must take place on the beating heart.

The myocardial muscle is formed into an organ composed of four chambers: the right and left atria
and ventricles. The atria serve as a place for collection of blood before it is pushed into the ventricles.
The left atrium collects oxygenated blood from the lungs. The right atrium collects de-oxygenated blood
to be sent to the lungs. From the ventricles, the blood is pumped to the rest of the body: the right
ventricle pumps blood through the pulmonary artery back to the lungs for filtration of carbon dioxide
and other wastes and for re-oxygenation; the left ventricle pumps oxygenated blood to the body through
the aorta. Because it pushes the blood a cumulative greater distance, the left ventricle tends to be
larger due to the greater required muscle mass. A diagram of blood flow through the heart appears in
Figure A.29.

The average heart has a mass of 330 g when unfilled. The largest human hearts when filled with
blood during diastole have masses as large as 570 g, and typically 425 g during systole. Thus, a large
heart’s typical ejection is about 145 g of blood per heartbeat [32]. Typical values for the displacement
of the chambers can be found in Table A.13 and typical heart dimensions can be found in the text in
Table 5.10.
Figure A.29: Heart anatomy and blood flow (modified from [3])
Table A.13: Heart chamber displacements (from [32])

<table>
<thead>
<tr>
<th>Chamber</th>
<th>Displacement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right ventricle</td>
<td>160 to 230 mL</td>
</tr>
<tr>
<td>Left ventricle</td>
<td>143 to 212 mL</td>
</tr>
<tr>
<td>Right Atrium</td>
<td>100 to 185 mL</td>
</tr>
<tr>
<td>Left Atrium</td>
<td>100 to 135 mL</td>
</tr>
</tbody>
</table>

A.1.1 The Coronary Arteries

Among the first offshoots of the aorta are the coronary arteries. Their position on the heart can be seen in Figure A.30. These bring oxygenated blood to the muscle of the heart itself. Due to the relative mass of the heart tissue and its oxygen demands compared to the rest of the body, these arteries are narrow and as a result tend to collect calcium and other materials as deposits on their walls. This leads to narrowing - ischemia - which may result in a complete cutoff of the artery from the myocardial tissue that it serves. This can result in necrosis (tissue death) and cardiac arrest as a result.

Coronary artery bypass grafting surgery, as to be described in Section A.3, is one method of preventing this and the form of cardiac surgery this thesis directly hopes to improve.
Appendix A. Cardiac Function

A.2 The Artificial Heart: The Cardio-Pulmonary Bypass Machine

The cardio pulmonary bypass system is used in surgery to replace the normal function of the heart and lungs. It functions similarly in that it filters the blood, removes carbon dioxide and adds oxygen to it. Flow rate and gaseous removal and addition are all controlled by the OR perfusionist based on the patient’s age, weight, height and base surface area (BSA). This control can allow for very close approximation of the patient’s natural circulatory system.

Despite the close approximation, there are still major differences in construction and operation compared to the natural heart and lungs, as would be expected. First, the pumping function is performed by roller pumps producing a constant pressure output instead of the pulsatile pressure normally produced and to which the aorta, arteries and arterioles are accustomed\(^1\). Secondly, the walls of the machine are not endothelially lined as blood vessels are.

A.2.1 The Negative Impact of CPB

A great deal of damaging effects arise because of the mentioned differences between the bypass machine and the cardiovascular system. Many of these have been known since 1966. Then, at a meeting of the American National Academy of Science’s Division of Medical Science’s Committee on Trauma, the following physiological problems in prolonged pumping were discussed: hemolysis (disrupted red blood cells, actually due to the trapping of RBCs between the pump’s rollers), anemia, alteration of RBC electrical characteristics and blood viscosity, potential damage to plasma proteins, changing clotting factors, and oxygenation causing fat embolization [37].

In more recent times, this list has been added to. Borst concluded in [22] that the problems of CPB were a result of the blood traveling through non-endothelially lined channels, to receiving gaseous and particulate emboli, and to experiencing non-physiologic shear stresses.

Due to the running of the blood through the artificial surfaces of the heart-lung machine’s tubes and filters and not through endothelium lined tissues, the immune response of the blood is triggered. Neutrophils and virtually all other humoral and cellular components of the inflammatory response are activated. The non-physiologic shear stresses delivered by the pump produce hemolysis as mentioned above. As well, there is a resultant 40% drop in circulating platelets and depression of the platelet function with a strong bleeding tendency. This bleeding tendency arises due to the destroyed blood cells.\(^1\)

\(^1\)By the time the pressure wave reaches the capillaries, venules, veins and venae cavae the blood pressure has lost its pulsatory motion [36].
causing the activation of kallikrein, which in turn causes the release of bradykinin and other vasodilators, and the activation of fibrinolytic cascade which may contribute to postoperative bleeding. Finally, there is also the enhanced production of interleukin-1 (an inducer of fever conditions and promoter of protein catabolism).

To avoid clogging of the heart-lung machine filters and pipes, anti-coagulants are added to the patient’s bloodstream at the beginning of a CPB operation. Heparin is typically used. To counter this, at the end of the operation, protamine (a coagulant) is added as well to the bloodstream. However, defective coagulation may occur as protamine does not completely restore the coagulation cascade inhibited by heparin prior to CPB.

In terms of man hours, CPB also has a negative impact. In order to establish CPB, upwards of eighty minutes ([22]) are added to the operating room time for all surgeons and staff involved. As well, an extra person - a perfusionist - is required throughout the procedure.

The extra equipment necessary is also a negative in the form of monetary costs. The heart-lung machine, itself, is a large expenditure. As well, each operation requires a number of expensive disposables (filters, oxygenator, ...) that for sanitary reasons need to be replaced with each operation.

There is also the risk of thrombosis due to the clamping of the aorta. Prior to putting the patient on full CPB, the aorta is clamped proximal to the insertion site of arterial CP cannula. This clamp can be dangerous should there be present arteriosclerosis as there would be for patients undergoing CABG. The danger here is the tendency of the clamping to produce arteriosclerotic emboli to the brain. There has been assessed a 0.5% chance of embolic complications for every operation [22].

As well, the replacement of the heart’s natural pulsative flow with the constant pressure of the heart-lung machine is believed to cause constriction and ischemia of the major arteries. It is also noted to cause the red blood cells to aggregate, renal function to be impaired, renin to be released, and cellular hypoxia leading to metabolic acidosis [2].

Without the use of CPB there would be a number of benefits, some have been partly outlined above. To following list summarizes these:

• Advantages of lesser costs:

  1. There is no need for a perfusionist.

  2. There is no need for the disposables associated with CPB.

  3. Hospital stay is reduced (from seven to ten days down to four).
Appendix A. Cardiac Function

• Advantages of improved health:

1. Reduced mortality of CABG.
2. Reduced morbidity.
3. Reduced need for intensive treatment of haemostatic disorders.
4. Reduced cardiac failure postoperatively.
5. Reduced reconvalence period and an earlier return to the patient's normal activities and lifestyle.

A.3 Coronary Artery Bypass Grafting Surgery

To bypass the artery stenoses, there are many techniques currently in practice and under review of which CABG surgery is the most reliable. Its basic strategy is the complete revascularization of the blood starved areas by bypassing all severe stenoses (greater than 50% diameter reduction) in all coronary branches and arterial trunks greater than 1 mm in diameter [2]. The revascularization is had by either harvesting a section of redundant vein or artery and reattaching it so that it will pass oxygenated blood to the previously cutoff area. For venous grafting, redundant veins (typically the saphenous veins of the lower leg) are removed and attached proximally to the aorta and distally to the grafting site on the other side of the coronary artery blockage. For arterial grafting, other arteries (usually the left or right internal mammary artery (LIMA and RIMA), but this could also be the abdominal arteries) are isolated and have their distal ends anastamosed (attached so as to pass blood) to the coronary artery beyond the block. The LIMA is suitable for grafts on the left anterior descending (LAD) coronary artery (CA), the diagonal CA, the circumflex artery/obtuse marginal artery and the ramus intermedius CA; the RIMA is suitable for grafting to these arteries as well as the RCA and the posterior descending artery [38].

The typical procedure is performed by two or more surgeons with the aid of nurses, an anaesthesiologist and a perfusionist.

Once the patient is unconscious and the necessary preparations made (EKG and pressure lines connected, intra-venous fluid supply, respiration, and urine removal lines attached), the saphenous vein is harvested (depending on the number and location of the the stenoses to be bypassed and the grafts to be made) for grafts to the aorta that the mammary artery cannot reach. Simultaneously, the chest is opened with a sternotomy and the rib cage spread to provide better access to the chest cavity. Again depending on the nature of the clots and their location, the mammary artery is mobilized for use as a
bypass. The pericardial sack is opened for access to the heart for grafting and for arterial and venous connection to the bypass machine. Then, the heart and lungs are stopped and circulation switched over to the heart-lung machine to perform the processes of carbon dioxide removal, oxygen addition, perfusion pressure maintenance and blood filtration. With the heart bypassed, the grafting sites are prepared and the grafts are made. On average one arterial and three venous grafts are made per case [22]. The heart is restarted, the chest and leg closed and the patient taken off to recover. The process usually requires between three and four hours.

A.3.1 CABG without Cardio-Pulmonary Bypass

CABG without CPB is not a common operation but has been performed in many research studies. It remains uncommon due to the ubiquitousness of the pump, whose reliability in the short term of the OR has so far precluded its long term performance in the recovery room.

Typically, the operation proceeds as it would with bypass. The patient is prepped (although heparin is not administered), the required veins (if any) are harvested, and the chest is opened by either a median sternotomy or by making portals in the interstitial tissues for minimally invasive surgery trocars. The pericardial sack is then opened to gain access to the heart and pinned so that it provides the greatest exposure to the myocardial tissue and to prevent the lungs from interfering. Then, with the heart exposed stitching can begin immediately or, as often is the case, the surgeon will attach sutures into the myocardial tissue and tie them to either the pericardial sack, the retractor or another structure (anatomical or otherwise) so that the area of interest on the heart will be stabilized. The work of the anaesthesiologist is different in that they refrain from heparin and protamine, unless there is an emergency and the pump is required. As well, they need to keep an anti-arrhythmic in readiness as the technique often requires cardiac manipulation. The perfusionist's job is also changed as they are made idle until required in an emergency.

This off-pump procedure is particularly useful for cases involving [8]:

1. Patients with heavily calcified aortas. These present a great risk of particulate embolization when clamping is performed and cannula attachment can cause this, dissection and rupture as well.
2. Patients with previous cardiac operations.
3. Patients with prior cerebrovascular accidents.
4. Patients on dialysis or with otherwise impaired renal function. The pump's non-pulsatile flow is known to present problems to this group, as discussed above.
5. Patients of religious background denying them blood transfusions, such as the Jehovah's witnesses.

Research in the Field

Various surgeons and research teams have attempted this surgery, generally with positive results. Work has been performed from as early as 1972 up to the present.

In research performed by Kluge [6], aorto-coronary bypass was performed without the use of artificial extracorporeal circulation in a minimally invasive manner. The heart was exposed through the fourth and fifth interspaces, the pericardium incised and retracted away to keep the lung’s motion from interfering with the surgery. A shunt was able to be sewn from the aorta to a proximal portion of the LAD and circumflex arteries. These were at the base of the heart where the myocardial contractions produced only a rocking motion that did not prohibit the procedures. However, the vigorous motion of the myocardium prevented more distal anastomoses.

Atkins [7] performed CABG without CPB on twenty-two patients. Median sternotomies were performed and then grafted to while the heart beat. This study was to determine the effects of CPB on septal function. It was found that there was an improvement in septal function as noted by improvement in the overall patient ejection fraction as compared to an on-pump control group.

Pfister et al. performed a comparison study between 220 patients in off-pump operations and a control group of 220 patients that went through the same procedures on-pump [8]. They found that the circumflex grafts require too great a torquing of the heart for the necessary exposure to be able to do them without bypass. Generally the number of grafts were small, only one or two per patient, and averaging 1.32 grafts per patient, and were always on the RCA, LAD or diagonal arteries. The results of their work was that the experimental group was seen to spend less time in ICU, had statistically less infection, less mediastinitis (infection of the sternum), a marked decrease in the need for postoperative transfusions, and a decrease in low cardiac output syndrome. Patients that benefited from this especially included the elderly, patients requiring rework, hypertensives, women and those with more impaired left ventricles.

In addition to the methods of suturing to the heart in order to stabilize it, mechanical devices have also been constructed. These include devices made by Cardio-Thoracic systems, Guidant, Medtronic and the device of this thesis. These either use suction or pressure to hold the suturing site still while suturing is taking place. Advantages of using these devices include [39]:

1. No ischemic arrhythmia.
Appendix A. Cardiac Function

2. No loss of regional cardiac pump function or hemodynamic deterioration.
3. No risk of acute myocardial infarction.
4. No time restraint on anastomosis suturing.
5. No need for regional preconditioning or cardioplegia.

The works of other surgeons and research teams in this area can be found in [9], [10] and [11].

A.3.2 Open-Heart vs. Minimally Invasive Surgery

As opposed to the standard sternotomy, where the chest is opened by cutting through the sternum and then retracting, minimally invasive surgery proposes that the operation be performed laparoscopically through trocars between the patient's ribs or through a portal made by removing a section of rib. This difference in approach to the same procedure would lead to much smaller wounds than the open-heart sternotomy permits but offers other difficulties in surgical procedure.

Due to the small workspace available, there will be patients whose required grafts make them more suitable to the minimally invasive technique and those that cannot be operated on with it at all. Patients requiring grafts to the RCA or LAD are particularly amenable to this approach. These areas of the heart are easily accessible to the minimally invasive entry points and offer less motion. The back of the heart would be completely inaccessible however, which is generally the most common location for anastomosis sites, on the circumflex coronary artery. Surgery could also not take place minimal invasively on opposite sites of the heart. This is in accordance with the patient being positioned on their side during the operation so that the portals may be opened up between their ribs. It is also likely that infants and children may not be able to be practiced on with minimally invasive techniques strictly due to the small dimensions of their ribs and intercostal spacing.

Another disadvantage is the lack of vision of minimally invasive surgery. Minimally invasive requires performing operation through portals, viewed with fiber optic cameras. This causes the loss of stereo vision for the surgeon making the operation more difficult. However, this problem is known to disappear as the surgeon gains experience.

Because of the disadvantage of unreachable locations on adults, there is a movement amongst those surgeons practicing MIDCAB to move back towards the heart-lung machine, but in a more limited manner, being applied without a sternotomy. Sternman et al, in their patent “Methods for Performing Thoracoscopic Coronary Artery Bypass” [38], proposed the performing of the CABG surgeries with minimally invasive techniques, and with the use of cardiopulmonary bypass and cardioplegia. This CPB
would be applied to the femoral artery and inferior vena cavae and the ascending aorta would be isolated with a balloon occlusion. As well, a lung would be collapsed and the patient put on their side to make room for the viewing cameras. This technique avoids the larger wounds of the sternotomy, but ignores the damaging effects of CPB.
Appendix B

Spiral Ring Calculations

One of the possible future forms for the cardiac stabilization device would be a spiral. This spiral ring would in essence be a C-ring with one of its edges stretched to encircle the other. Although the construction means were not readily available (casting or CNC milling would most likely be required), calculations were pursued to determine its viability.

For the spiral, the inner and outer radii of the ring walls would expand linearly with angle $\theta$. At $\theta = 2\pi$ the radii would both be larger by the thickness of the mammary artery space, $t$, plus the difference between the two radii, $h$. Mathematically, the radii, inner and outer respectively, would appear as:

$$
\begin{align*}
r &= r_0 + \frac{h + t}{2\pi} \theta \\
R &= R_0 + \frac{h + t}{2\pi} \theta
\end{align*}
$$ (B.15)

The expanding radii quickly lead to an undesirably large sized ring, which is much more difficult to apply to the curved surface of the heart and have hold. Either a much higher vacuum to force the heart to contort to the ring or a flexible spiral to conform to the surface of the heart would be required. The size of the ring can be characterized by its maximum diameter:

$$
\begin{align*}
d_{max} &= R_{\theta_{max}} + R_{\theta_{max} - \pi} \\
&= R_{2\pi+n} + R_{\pi+n} \\
&= R_0 + \frac{h + t}{2\pi}(2\pi + n) + R_0 + \frac{h + t}{2\pi}(\pi + n) \\
&= 2R_0 + \frac{h + t}{2\pi}(3\pi + 2n)
\end{align*}
$$ (B.16)

For a spiral with overlap of $n = \frac{3}{4}$, starting radii $r_0 = 19\,mm$ and $R_0 = 24\,mm$ (18$mm$ and 25$mm$ when taking the wall thickness into account), and mammary space $t = 10\,mm$, the maximum diameter will be 78.3 mm.

The force that would be provided by the ring was calculated as an integration of the pressure over infinitesimal segments of the vacuum surface area. These areas were approximated as trapezoids with
parallel sides \(a\) and \(b\), height \(h\) and area \(\frac{1}{2}(a+b)h\). Substituting the infinitessimal interior circumference, \(dc\), for \(a\), and the infinitessimal exterior circumference, \(dC\), for \(b\), the trapezoid area will be: \(dA = \frac{1}{2}(dc + dC)h\). The total area is found by summing over the respective circumferences:

\[
A = \int_0^c \frac{1}{2} h \, dc + \int_0^C \frac{1}{2} h \, dC
= \frac{h}{2} \left[ \int_0^{2\pi + n} (r_0 + \frac{h + t}{2\pi} \theta) d\theta + \int_0^{2\pi + n} (R_0 + \frac{h + t}{2\pi} \theta) d\theta \right]
= \frac{h}{2} \left[ (2\pi + n)(R_0 + r_0) + \frac{h + t}{2\pi} (2\pi + n)^2 \right]
\]

(B.17)

where \(n\) is the amount of overlap to the spiral, in radians.

Applying this calculation to a spiral with no overlap and starting radii of 19 and 24 mm, and a mammary artery space of 10 mm, the area will be \(9.11 \times 10^{-4}m^2\). For the same dimensions but for an overlap of 30°, this becomes \(10.1 \times 10^{-4}m^2\). In contrast to this an O-ring with similar radii would have area \(6.75 \times 10^{-4}m^2\).

### B.1 The Elliptical Spiral

In an effort to get around the problem of the increasing size with expanding radius, the elliptical spiral was designed. This spiral design was similar to the first spiral except that instead of expanding linearly its expansion would be based on a sinusoid. Thus, a full \(2\pi + n\), \(n > 0\) radian expansion could be had in a much more compact shape by concentrating the expansion about the axis perpendicular to the axis of origin. This is shown in Figure B.31.

The effective area and vacuum force for this shape are calculated as before:

\[
A = \int_0^c \frac{1}{2} h (dc + dC)
= \frac{h}{2} \left[ \int_0^{2\pi + n} \left( r_0 + R_0 \right) d\theta \right]
\]

(B.18)

Now, \(r\) and \(R\) change with distance along the circumference such that they will be described by a linear expansion and a sinusoidal expansion. To maintain the vertically stretched shape (as the axis of origin is in the horizontal in this case), the sinusoid will be at its positive maximum along the vertical axis. To do this the sine term used is at double the frequency of rotation and shifted:

\[
r = a_0 + (a_x - a_0) \frac{\theta}{\pi} + \left( b - a_0 - (a_x - a_0) \frac{\pi/2}{\pi} \right) \sin \left( 2\theta - \frac{\pi}{2} \right)
= a_0 + (a_x - a_0) \frac{\theta}{\pi} + \left( b - \frac{(a_x + a_0)}{2} \right) \sin \left( 2\theta - \frac{\pi}{2} \right)
\]
NonLinear Spiral: \( a_0 + (a_\pi - a_0) \frac{\theta}{\pi} + (b - (a_\pi + a_0)/2) \sin(\theta(t)) \)

The last line of the equation was arrived at through use of the double angle theorem for \( \sin(a \pm b) = \sin(a)\cos(b) \pm \sin(b)\cos(a) \).

The area beneath this shape would then be:

\[ A = \frac{h}{2} \left[ \int_0^{2\pi+n} (r_0 d\theta + Rd\theta) \right] \]  

for:

\[ \int_0^{2\pi+n} r d\theta = \int_0^{2\pi+n} a_0 + (a_\pi - a_0) \frac{\theta}{\pi} + (b - (a_\pi + a_0)/2)(-\cos(2\theta)) d\theta \]
\[ = a_0(2\pi + n) + \frac{1}{2\pi} (a_\pi - a_0)(2\pi + n)^2 + (b - a_\pi + a_0)/2 (-\frac{1}{2} \sin 2\theta) \]
\[ = a_0(2\pi + n) + \frac{1}{2\pi} (a_\pi - a_0)(2\pi + n)^2 - \frac{1}{2} (b - a_\pi + a_0) \sin(4\pi + 2n) \] (B.21)

\[ \int_0^{2\pi+n} Rd\theta = \int_0^{2\pi+n} (r + h) d\theta \]
\[ = (a_0 + h)(2\pi + n) + \frac{1}{2\pi} (a_\pi - a_0)(2\pi + n)^2 - \frac{1}{2} (b - a_\pi + a_0) \sin(4\pi + 2n) \] (B.22)
Again, $R$ is larger than $r$ by the thickness of the ring, $h$. The $h$ term appears only in the first term of $R$ as it cancels out through subtraction in the other terms of the equation.

Combining the equations would result in an area of:

$$A = \frac{h}{2}(2a_0 + h)(2\pi + n) + \frac{h}{2\pi}(a_r - a_0)(2\pi + n)^2 - \frac{h}{2}(b - \frac{a_r + a_0}{2})\sin(4\pi + 2n)$$  \hspace{1cm} (B.23)

Due to the presence of the negative phase of the sinusoid there is, a substantial amount of waviness in the function. This can be alleviated through the replacement of the double sine function with a squared sine function. Then, the inner radius becomes:

$$r = a_0 + (a_r - a_0)\frac{\theta}{\pi} + (b - \frac{a_0 + a_r}{2})\sin^2\theta$$ \hspace{1cm} (B.24)

The new integration of the radius becomes:

$$\int_0^{2\pi+n} r\,d\theta = a_0(2\pi + n) + \frac{a_r - a_0}{2\pi}(2\pi + n)^2 + (b - \frac{a_r + a_0}{2})\int_0^{2\pi+n} \sin^2\theta\,d\theta$$ \hspace{1cm} (B.25)

Since $\int \sin^2 au\,du = \frac{\theta}{2} - \frac{1}{4a}\sin 2au + C$:

$$\int_0^{2\pi+n} \sin^2\theta\,d\theta = \frac{\theta}{2} - \frac{1}{4}\sin 2\theta\big|_0^{2\pi+n} = \frac{2\pi + n}{2} - \frac{1}{4}\sin(4\pi + 2n)$$ \hspace{1cm} (B.26)

Continuing the previous integration:

$$\int_0^{2\pi+n} r\,d\theta = a_0(2\pi + n) + \frac{a_r - a_0}{2\pi}(2\pi + n)^2 + (b - \frac{a_r + a_0}{2})\left[\frac{2\pi + n}{2} - \frac{1}{4}\sin(4\pi + 2n)\right]$$ \hspace{1cm} (B.27)

and the outside radius ($R = r + h$) integrates as:

$$\int_0^{2\pi+n} R\,d\theta = (a_0 + h)(2\pi + n) + \frac{a_r - a_0}{2\pi}(2\pi + n)^2 + (b - \frac{a_r + a_0}{2})\left[\frac{2\pi + n}{2} - \frac{1}{4}\sin(4\pi + 2n)\right]$$ \hspace{1cm} (B.28)

The area will then be:

$$A = ha_0(2\pi + n) + \frac{h^2}{2}(2\pi + n) + \frac{h}{2\pi}(a_r - a_0)(2\pi + n)^2 + h(b - \frac{a_r + a_0}{2})\left[\frac{2\pi + n}{2} - \frac{1}{4}\sin(4\pi + 2n)\right]$$ \hspace{1cm} (B.29)

The maximum diameter for these rings will either be across the vertical axis defined by $\theta = \pi/2$ and $\theta = 3\pi/2$ or along the diameter defined by the endpoint and one half rotation before it ($\theta_{end} = \pi$).

For an elliptical ring with parameters $a_0 = 5$, $a_r = 12$, $h = 5$, and $n = \pi/6$, the area will be $6.2 \times 10^{-4} m^2$, almost that of the C- and O-rings. This particular spiral is shown in Figure B.31.
To calculate the magnitude of the pneumatic arm (both with pistons and without) ball and socket joint frictional torques and thereby the maximum torques these joints could resist, frictional-force-derived torques on infinitesimal areas were summed over the contact surfaces. For the pneumatic arm these surfaces were between the piston and the ball, and between the ball and the outer shell. For the stainless steel arm there was only the one surface of contact, between the rubber hemisphere and the outer shell. The friction forces on these surfaces act in opposition to the imposed motion thus creating the braking action desired.

In Figure C.32 a), the infinitesimal area (delineated by \( rd\theta \) and \( rd\phi \)) is shown, along with the acting forces and their relation with the centroid of the ball. Acting on the infinitesimal segment is the pressure derived force acting in the direction the air is pushing the ball. The friction derived from it is then found by multiplying its projection onto the surface normal by the coefficient of static friction for the materials concerned. The pressure force is derived from the air pressure multiplied by the surface area over which it acted. The surfaces used were the infinitesimal bands which can be seen in Figure C.32 b).

The braking torque calculations were performed by first calculating the force of friction acting on an infinitesimal area of the socket. Then, the torque arms to each surface were calculated and the infinitesimal torque found by multiplying them together. The general case for any braking surface follows.

The force of friction generated due to the pressure will equal the pressure multiplied by the area it acts upon, scaled by the coefficient of static friction. Then, assuming that the pressure acts equally on all areas, the infinitesimal section, \( dA \), will be acted on by force of friction:

\[
\begin{align*}
  dF_{\text{pressure}} &= \mu P dA = \mu Prd\phi \cdot rd\theta \\
  &= \mu Pr^2 d\phi d\theta
\end{align*}
\]  

for coefficient of friction \( \mu \), pressure \( P \), and ball radius \( r \). This is assuming that the pressure acts equally on all areas.
Appendix C. Ball Socket Joint Brake Torque Calculations

Then, the braking torque generated by the brake will be found by allowing this friction force to act through the torque arm, the distance between the surface of contact and the axis of rotation. Due to the alignment of the joint, different resistance torques are generated at the $z$ axis (the roll axis), as compared to the $y$ (pitch) and $x$ (yaw) axes. Pitch and yaw are identical in this application as the socket is symmetrical.

C.1 $z$-Axis Brake Torque Calculations

For $\tau_z$, the torque is calculated over infinitessimal cylindrical bands of radius $rsin\phi$ and height $rd\phi$. Thus, the area for integration is:

$$dA = 2\pi r^2 sin\phi d\phi$$  \hspace{1cm} (C.32)

The friction arising on this band of contact will then be:

$$dF = \mu P 2\pi r^2 sin\phi d\phi$$  \hspace{1cm} (C.33)

for coefficient of friction $\mu$, pressure $P$ and ball radius $r$. $\phi$ is the angle between the $z$ axis and the surface.

For these bands the perpendicular distance to the $z$ axis is $D = rsin\phi$. The magnitude of the
resultant torque can then be found as the product of the force and this distance, integrated over the surface of contact:

$$\tau_z = \int_a^b \mu P2\pi r^3 \sin^2 \phi d\phi$$  \hspace{1cm} (C.34)

Since, from [40],

$$\int \sin^2 \alpha \, d\alpha = \frac{a}{2} - \frac{\sin 2\alpha}{4a} + C$$  \hspace{1cm} (C.35)

therefore, the resistance torque about the z axis will be:

$$\tau_z = \mu P2\pi r^3 \left[ \frac{\phi}{2} - \frac{\sin(2\phi)}{4} \right]_a^b$$  \hspace{1cm} (C.36)

C.2 z and y-Axis Brake Torque Calculations

Calculations for the x and y axes proceed differently due to the different orientation. To handle this, the infinitessimal surfaces are not bands but instead segments similar to the one in Figure C.32 a). The area for this will then be

$$dA = r \, d\phi \cdot \sin \phi r d\theta = r^2 \sin \phi d\phi d\theta$$  \hspace{1cm} (C.37)

where, again, $\phi$ is the angle between the z axis and the segment, and $\theta$ is the angle of rotation about the z axis. The force of friction upon it will then be:

$$dF = \mu Pr^2 \sin \phi d\phi d\theta$$  \hspace{1cm} (C.38)

![Diagram](image)

**Figure C.33**: Braking calculations about the x axis

The torque arm, the distance to the area segment from the axis of rotation, is $D = r |\sin \theta|$. The absolute value bars are to prevent cancellation of the torques as integration occurs around the band. The sign of the torque from the individual areas is not important, only the magnitude of the torque.
generated. A picture of the torque arm and how it relates to the segment can be seen in Figure C.33. Then the torque about these axes would be:

\[
\tau = \int_{0}^{2\pi} \int_{a}^{b} \mu P r^2 \sin \phi d\phi d\theta |\sin \theta|
\]

\[
= 2\mu P r^3 \int_{0}^{\pi} \int_{a}^{b} \sin \phi \sin \theta d\phi d\theta
\]

\[
= 2\mu P r^3 [-\cos \phi]_a^b \int_{0}^{\pi} \sin \theta d\theta
\]

\[
= 2\mu P r^3 [-\cos \theta]_a^b [-\cos \theta]_0^\pi
\]

\[
= 4\mu P r^3 [-\cos \phi]_a^b
\]

(C.39)
Appendix D

Testing Procedure Forms

This appendix contains the form that was used for collecting personal details of the subjects involved and their perceptions and impressions for and of the optical stabilization tests. As well, following this is the form for timing evaluation of subject performance during the tests. These tests are detailed in Chapter 3.
Appendix D. Testing Procedure Forms

Strobe-Assisted Suturing Testing/Human Factors Study

Participants Details

Name: 
Age: 
Sex: 
Height: 
Occupation: 

I agree that both this investigation and my part in it have been defined and fully explained to me, and that I understand the explanations. I have been provided with a description of the procedures to be used in this investigation, as well as of any anticipated risks or discomforts and I have discussed them in detail with the investigator. I have been given an opportunity to ask whatever questions I might wish.

___________________________   _________________________
Subject's signature                Date

Subject's Evaluation of Strobing

Score on a scale from one (low applicability) to ten (high).

Comfort in performing task: 1: _____ 2: _____ 3: _____ 4: _____ 5: _____
Demands on concentration: 1: _____ 2: _____ 3: _____ 4: _____ 5: _____
Fatigue from performing the task: 1: _____ 2: _____ 3: _____ 4: _____ 5: _____

Comments
Appendix D. Testing Procedure Forms

Test Evaluation

Test One: Static Surface

Accuracy in point placement: ______________________
Neatness in point placement: ______________________
Time: ______________________

Test Two: Moving Surface

Accuracy in point placement: ______________________
Neatness in point placement: ______________________
Time: ______________________

Test Three: Moving Surface with Strobing

Accuracy in point placement: ______________________
Neatness in point placement: ______________________
Time: ______________________

Test Four: Moving Surface with Complicated Motion

Accuracy in point placement: ______________________
Neatness in point placement: ______________________
Time: ______________________

Test Five: Moving Surface with Complicated Motion with Strobing

Accuracy in point placement: ______________________
Neatness in point placement: ______________________
Time: ______________________
Bibliography


[15] Dr. Samuel Lichtenstein, St. Paul’s Hospital, 1081 Burrard St., Vancouver, British Columbia, Canada. Personal communication.

[16] Dr. Kassam Ashe, St. Paul’s Hospital, 1081 Burrard St., Vancouver, British Columbia, Canada. Personal communication.


