MRI Compatible Optical Force Sensing Technique for Robotically Assisted Biopsies

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Abstract

Conducting accurate biopsies of tissue is critical for early cancer detection. The goal of this project was to develop a force feedback system, to be used in automated MRI biopsy procedures. The design utilizes interferometry techniques, paired with a custom fixture and processing circuitry to translate strain experienced by a needle into force. A multi-DOF fixture was machined to allow for detecting axial and lateral forces on the needle. An optics table was built to ensure stable and precise positioning of the components. A PCB was tailor-made to integrate the mechanics and electronics to provide a platform to analyze the optical feedback. Testing was completed using phantoms, calibration weights, simulation and formal design verification. The fixture was evaluated for multi-DOF sensing using strain gauges and validated the FPI sensing principle with axial forces. The device begins the process of collecting accurate needle force profiles for future robots that utilize needle steering. Eventually, this data may allow for in vivo characterization of cancerous tissue.
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# Contents

Abstract  
Acknowledgements  
List of Figures  
List of Tables  
Abbreviations

1 Executive Summary  
2 Introduction  
3 Literature Review  
   3.1 Basics of MRI  
   3.2 Robot Assisted Biopsy Procedure  
   3.3 Needle Force Profile  
   3.4 Force Progression Through Secondary Media  
   3.5 Previous Projects  
4 Project Strategy  
5 Alternative Designs  
   5.1 Needle Driver Force  
   5.2 Friction Force Needle  
   5.3 Radioactive Tracking  
   5.4 Fiber Bragg Grating  
   5.5 Fabry Perot Interferometer - Single Sensor  
   5.6 Fabry-Perot Interferometer - Multiple Sensors  
   5.7 Pressure Sensing  
   5.8 Design Option Analysis  
   5.9 Other Options Not Considered  
6 Design Verification and Validation  
   6.1 Basic Principle Overview  
   6.2 Fixture Design
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.2.1</td>
<td>Physical Constraints</td>
<td>27</td>
</tr>
<tr>
<td>6.2.2</td>
<td>Design Process</td>
<td>28</td>
</tr>
<tr>
<td>6.2.3</td>
<td>Analysis of Fixture Design 1</td>
<td>29</td>
</tr>
<tr>
<td>6.2.4</td>
<td>Analysis of Fixture Design 2</td>
<td>32</td>
</tr>
<tr>
<td>6.3</td>
<td>Fixture Design Testing</td>
<td>37</td>
</tr>
<tr>
<td>6.3.1</td>
<td>Equipment</td>
<td>37</td>
</tr>
<tr>
<td>6.3.2</td>
<td>Axial Forces</td>
<td>39</td>
</tr>
<tr>
<td>6.4</td>
<td>Fiber Optic Design</td>
<td>42</td>
</tr>
<tr>
<td>6.4.1</td>
<td>Physical Constraints</td>
<td>42</td>
</tr>
<tr>
<td>6.4.2</td>
<td>FPI Sensor Specifications</td>
<td>43</td>
</tr>
<tr>
<td>6.4.3</td>
<td>Lens Setup</td>
<td>44</td>
</tr>
<tr>
<td>6.4.4</td>
<td>Interference Change</td>
<td>44</td>
</tr>
<tr>
<td>6.5</td>
<td>Circuit and PCB Design</td>
<td>45</td>
</tr>
<tr>
<td>6.5.1</td>
<td>Purpose and Constraints</td>
<td>46</td>
</tr>
<tr>
<td>6.5.2</td>
<td>Light Sample Collection</td>
<td>46</td>
</tr>
<tr>
<td>6.5.3</td>
<td>TSL1402R Photodiode Array Functionality</td>
<td>47</td>
</tr>
<tr>
<td>6.5.4</td>
<td>Analog to Digital Conversion (ADC)</td>
<td>48</td>
</tr>
<tr>
<td>6.5.5</td>
<td>Analog Filter to ADC</td>
<td>50</td>
</tr>
<tr>
<td>6.5.6</td>
<td>Computation and Control</td>
<td>55</td>
</tr>
<tr>
<td>6.5.7</td>
<td>Printed Circuit Board</td>
<td>57</td>
</tr>
<tr>
<td>6.6</td>
<td>Circuit and PCB Testing</td>
<td>63</td>
</tr>
<tr>
<td>6.7</td>
<td>Digital Circuitry</td>
<td>65</td>
</tr>
<tr>
<td>6.7.1</td>
<td>Overall Design</td>
<td>66</td>
</tr>
<tr>
<td>6.7.2</td>
<td>Clocks</td>
<td>67</td>
</tr>
<tr>
<td>6.7.3</td>
<td>Sample Control</td>
<td>67</td>
</tr>
<tr>
<td>6.7.4</td>
<td>ADC Control</td>
<td>68</td>
</tr>
<tr>
<td>6.7.5</td>
<td>SPI Communication</td>
<td>69</td>
</tr>
<tr>
<td>6.8</td>
<td>Digital Circuitry Testing and Simulation</td>
<td>70</td>
</tr>
<tr>
<td>6.9</td>
<td>Algorithm</td>
<td>75</td>
</tr>
<tr>
<td>6.9.1</td>
<td>Development</td>
<td>75</td>
</tr>
<tr>
<td>6.9.2</td>
<td>Background</td>
<td>78</td>
</tr>
<tr>
<td>6.9.3</td>
<td>Principal of Operation</td>
<td>79</td>
</tr>
<tr>
<td>6.9.4</td>
<td>Implementation</td>
<td>79</td>
</tr>
<tr>
<td>6.10</td>
<td>Algorithm Testing</td>
<td>82</td>
</tr>
</tbody>
</table>

7 Discussion  

7.1 Fixture Testing Results  
7.1.1 Axial Force Testing  
7.1.2 Non-Axial Force Testing  
7.1.3 Simulation Results of Non-Axial Forces  
7.1.4 Experiment Results of Non-Axial Forces  
7.1.5 Fixture Testing Discussions  

7.2 Algorithm Testing  

7.3 Economics  

7.4 Environmental Impact  

7.5 Societal Impact  

7.6 Manufacturability  

7.1 Fixture Testing Results  
7.1.1 Axial Force Testing  
7.1.2 Non-Axial Force Testing  
7.1.3 Simulation Results of Non-Axial Forces  
7.1.4 Experiment Results of Non-Axial Forces  
7.1.5 Fixture Testing Discussions  

7.2 Algorithm Testing  

7.3 Economics  

7.4 Environmental Impact  

7.5 Societal Impact  

7.6 Manufacturability
8 Recommendations and Conclusion 98
  8.1 Recommendations ........................................... 98
  8.2 Conclusion .................................................. 100

A Design Option Analysis Scoring 101

B Experiments 103
  B.1 Fixture Design Testing ...................................... 103
    B.1.1 Equipment ................................................. 103
    B.1.2 Methodology ............................................. 104
    B.1.2.1 Axial Forces .......................................... 104
    B.1.2.2 Non-Axial Forces ..................................... 105
  B.2 Analog Rise Delay Testing .................................. 106
    B.2.1 Equipment ................................................. 106
    B.2.2 Methodology ............................................. 106
    B.2.3 Data ...................................................... 107
    B.2.4 Conclusions .............................................. 107

C Engineering Drawings 108
  C.1 Design 1 .................................................... 108
  C.2 Design 2 .................................................... 109
  C.3 PCB Schematics .............................................. 110

Bibliography 126
### List of Figures

1.1 Complete system: including FPI, fixture, laser, coupler, linear array, and PCB . 1  
1.2 Fabry Perot interferometer in fixture using a green laser. ............... 3

3.1 A typical needle insertion force curve from Okaumura [1] ................. 9  
3.2 calibrated and theoretical Strain Results from previous project design[2] . 15

5.1 Experimental setup of the Pressure sensor. .................................. 22

6.1 Overview of the entire system. .................................................. 26  
6.2 Fixture Location in the Needle Driver Module of the BRP robot. .......... 27  
6.3 Initial strain testing for a circular shaped fixture .......................... 29  
6.4 Designs Considered ............................................................... 29

6.5 Behavior of the spokes in design 1 .......................................... 30

6.6 Design 1 ............................................................................. 30

6.7 Ideal force-strain relationship of design 1 from 0N-15N. .................. 32

6.8 Behavior of the spokes in design 2 .......................................... 33

6.9 Design 2 ............................................................................. 34

6.10 Ideal force-strain relationship of design 2 from 0N-15N. ................. 35

6.11 BRP Robot along with design 2 of the fixture ............................... 36

6.12 Strain gauges placed on the fixture in order to measure strain due to force. 37

6.13 1 of 3 identical circuits used to measure measure strain. ................. 38

6.14 Wheatstone bridge ................................................................... 39

6.15 Set up to test axial forces ............................................................ 40

6.16 Set up to test non-axial forces ...................................................... 41

6.17 Configurations used to experiment non-axial forces .......................... 41

6.18 Robot Controller. The red box shows where the FPI board will go ...... 42

6.19 The board and lenses in a PCIe Slot ............................................ 43

6.20 FISO Force Sensor ..................................................................... 43

6.21 Optical Setup Diagram ............................................................... 44

6.22 Actual layout of lenses ............................................................... 45

6.23 Circuit Diagram ......................................................................... 46

6.24 Balloon popped at 1500 frames per second .................................... 48

6.25 Scanning across the array with white light. The last pixels are blocked by the edge of electrical tape .................................................. 49

6.26 Circuit that was used for 6.25 and other tests. ................................. 49

6.27 General Sallen Key Topology ...................................................... 51

6.28 Cascaded Filter ....................................................................... 54

6.29 Cascaded Filter Values .............................................................. 54
6.30 Analysis of filter with exact values ......................................................... 55
6.31 Analysis of filter using chose capacitors ................................................. 55
6.32 GPIO and Power for microcontroller ....................................................... 56
6.33 Fuses for power rails to external MCU and Laser ..................................... 56
6.34 TXB0108PWR Bidirectional logic level converter ....................................... 56
6.35 FTDI and USB Type B Connector ............................................................ 57
6.36 JTAG and LED ......................................................................................... 57
6.37 PCIe Style Connectors ............................................................................. 58
6.38 Trapezoid on Ground Plane between Analog and Digital Sections .............. 59
6.39 Capacitors in Parallel ............................................................................. 59
6.40 A set of decoupling capacitors ............................................................... 60
6.41 USB Shield .............................................................................................. 60
6.42 Zero ohm resistors .................................................................................. 60
6.43 The 3v3 Power Rail .................................................................................. 61
6.44 Drill hole locations ................................................................................... 61
6.45 Front of PCB ............................................................................................ 62
6.46 Back of PCB ............................................................................................ 62
6.47 Laser Holder ............................................................................................ 62
6.48 The PCB .................................................................................................. 63
6.49 JTAG chain, PROM programming and FPGA debug success ..................... 63
6.50 First Full Power Test of Board .................................................................. 64
6.51 Testing Setup for Filter ........................................................................... 64
6.52 Filter Response from 0Hz to 2MHz. ......................................................... 64
6.53 Filter Cutoff Frequency ........................................................................... 65
6.54 200KHz Response .................................................................................. 65
6.55 Digital Circuitry Block Diagram ............................................................... 67
6.56 sensor clock timing simulation ................................................................. 70
6.57 scope of sensor clock as compared to 4MHz clock .................................... 71
6.58 sample control simulation results ............................................................... 71
6.59 sample control oscilloscope results ........................................................... 72
6.60 analog to digital converter control timing .................................................. 73
6.61 scope of adc input timings ........................................................................ 73
6.62 top module simulation results ................................................................. 74
6.63 Close up of the beam splitter on the light table ......................................... 75
6.64 Fringes after filtering. Two consecutive frames are shown ......................... 76
6.65 Light table setup after processing in Excel. The red lines correspond to fringe movement over time. ................................................................. 77
6.66 Linear Array exposed to the pattern from the FPI. Green laser light is used because red does not show up well with a camera ........................................ 78
6.67 Algorithm Diagram ................................................................................. 80
6.68 Screenshot of the Waveform Utility ......................................................... 81
6.69 Noise from input with moving average ..................................................... 81
6.70 Adding and removing 10 sets of 20 gram weights ...................................... 82
6.71 Pushing down with finger then gradually releasing ..................................... 83
7.1 Data acquired for 0g to 680g weights. ......................................................... 85
7.2 Results used to linearly extrapolate. .......................................................... 86
7.3 Simulated results for the tested fixture. ........................................ 87
7.4 Simulation results - Configuration 1. ........................................ 89
7.5 Simulation results - Configuration 2. ........................................ 90
7.6 Simulation results - Configuration 3. ........................................ 91
7.7 Configuration 1 experiment results ............................................ 92
7.8 Configuration 2 experiment results ............................................ 93
7.9 Configuration 3 experiment results ............................................ 94
7.10 Force Output calculated using 0.766N per cycle .......................... 96

8.1 Red and Green Lasers .......................................................... 99
8.2 Resulting Yellow FPI ............................................................ 99

C.1 Engineering Drawing of Design 1 ............................................. 108
C.2 Engineering Drawing of Design 2 ............................................. 109
C.3 Engineering Drawing of FPGA ............................................... 110
C.4 Engineering Drawing of ADC ............................................... 111
C.5 Engineering Drawing of Filter ............................................... 111
C.6 Engineering Drawing of Power Connector ................................ 112
C.7 Engineering Drawing of USB ............................................... 112
C.8 Engineering Drawing of Programming Circuit. From Paulo Carvalho 113
C.9 Engineering Drawing of Power Circuit. From Paulo Carvalho ....... 114
C.10 Engineering Drawing of Output Connector. From Paulo Carvalho ... 115
C.11 Engineering Drawing of Logic Connector. From Paulo Carvalho .... 116
C.12 PCB Layout - Top Layer ...................................................... 117
C.13 PCB Layout - Bottom Layer .................................................. 118
C.14 PCB Layout - Silkscreen Layer .............................................. 119
C.15 PCB Layout - Bottom Solder ................................................ 120
C.16 PCB Layout - Top Solder ....................................................... 121
C.17 PCB Engineering Drawing for holes 1. This is the same file as the laser holder cutout ................................................................. 122
C.18 PCB Engineering Drawing for holes 2. This is the same file as the laser holder cutout ................................................................. 123
C.19 3d Render of PCB ............................................................ 123
C.20 PCB Bill of Materials .......................................................... 124
C.21 Logo .............................................................................. 125
List of Tables

5.1 Design option analysis scoring .............................................................. 23
6.1 QPSK Like States for Direction. T: Tension and C: Compression .............. 79
B.1 Delay Testing Data ............................................................................... 107
B.2 Delay Testing Averages ........................................................................ 107
## Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Full Form</th>
</tr>
</thead>
<tbody>
<tr>
<td>FPI</td>
<td>Fabry Perot Interferometry</td>
</tr>
<tr>
<td>BRP</td>
<td>Bioengineering Research Partnership</td>
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<tr>
<td>AIM</td>
<td>Automation and Interventional Medicine</td>
</tr>
<tr>
<td>FPGA</td>
<td>Field Programmable Gate Array</td>
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<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
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<tr>
<td>EMI</td>
<td>ElectroMagnetic Interference</td>
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<tr>
<td>CNC</td>
<td>Computer Numerical Control</td>
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<tr>
<td>OD</td>
<td>Outer Diameter</td>
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<tr>
<td>DPI</td>
<td>Dots Per Inch</td>
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<tr>
<td>ADC</td>
<td>Analog to Digital Converter</td>
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<tr>
<td>RTL</td>
<td>Register Transfer Level</td>
</tr>
<tr>
<td>FPS</td>
<td>Frames Per Second</td>
</tr>
</tbody>
</table>
Chapter 1

Executive Summary

The early detection of cancer is vital to reducing the mortality rate of the disease. One method of detection, known as a biopsy, samples suspicious tissue through the use of a needle to test for cancerous cells. Currently, the Bioengineering Research Partnership (BRP) robot, developed by WPI’s Automation and Interventional Medicine (AIM) Robotics Research Laboratory in collaboration with Harvard’s Surgical Planning Laboratory and Johns Hopkins University, helps
to guide the biopsy needle accurately to the area to be examined. However, there is still room for improvement in this process.

Throughout the course of this project, the team was tasked with constructing a sensor to allow the robot to accurately determine the forces present on the needle during a biopsy, while keeping the entire device MRI compatible.

The team began by characterizing all of the forces present on a needle during a biopsy procedure. These forces include the stiffness force, the friction force and finally, the cutting force. Respectively, these forces are the force pressing against the needle tip prior to puncture, the force acting upon the length of the needle after puncture, and the force pushing against the needle tip after puncture. With this knowledge, the team chose to focus on measuring the stiffness and cutting force acting upon the needle during a biopsy.

Examining potential options for measuring these forces in an MRI environment, as well as other design constraints, led the team to develop several conceptual designs for the sensor. Ultimately, the team chose to design a custom fixture to be placed behind the needle on the robot. The fixture holds up to three Fabry-Perot interferometry strain sensors, which couple to a laser. These strain sensors transmit light of varying intensity, depending on the strain placed upon the sensor. By connecting the output of the sensor to a photodiode array, the force placed on the fixture can be calculated.

Ultimately, the team designed a system that was able to measure axial forces. The device could fit inside the current version of the BRP robot and the components inside of the control box. The basic principle behind our design is that the force experienced by the needle strains the previously mentioned custom fixture, which holds an optical sensor, the FISO FOS-N strain sensor. A laser of a specified wavelength is used to send light to the FPI sensor. Based on interferometric principles, the FPI sensor outputs a pattern that changes depending on the forces experienced by the fixture. The pattern is focused onto a linear photodiode array which consists of two sub arrays of 128 pixels allowing us to record a section of the output pattern in terms of intensity. Analog data from the array goes through a filter after which the data is converted to digital signals using an ADC. The data is processed on a custom PCB which uses an FPGA and microcontroller to run an algorithm to analyze and calibrate the data.
Throughout the course of this project, the team followed an axiomatic design process. Each customer need presented in the project strategy portion of the report was translated into design parameters to be tested within the design verification section of this report.

Although the team has made major progress toward the goal of developing a sensor suitable for the Bioengineering Research Partnership robot, there is room for future projects to continue the work presented in this report. The current fixture designed by the team allows for more Fabry-Perot interferometry strain sensors to be placed, to begin examining needle deflection during biopsy procedures. Furthermore, several improvements could also be made to the current design, to allow for more accurate readings, and to improve system stability.
Chapter 2

Introduction

Cancer poses a great medical challenge to scientists and doctors today due to the complex nature in detection and treatment. Over 1.6 million people were diagnosed with cancer in 2015 in the United States alone [3]. Early detection through sampling suspicious tissue (biopsy) stands as a major facet in reducing the overall mortality rate of these diseases. Intensive research presently, and over the last ten years, focuses on increasing the accuracy of detection during biopsy.

The team observed the need for advancement in the biopsy field during a visit to Brigham and Women’s Hospital in Boston. During the visit, the team witnessed the Bioengineering Research Partnership (BRP) robot, developed by WPI’s Automation and Interventional Medicine (AIM) Robotics Research Laboratory in collaboration with Harvard’s Surgical Planning Laboratory under a National Institutes of Health grant, help perform a biopsy inside of an MRI machine.

This recent development attempts to eliminate errors in recognizing and sampling cancerous tissue by assisting the surgeon during the biopsy process. The robot helps guide the needle accurately to the planned area. The process is broken into multiple steps. A primary scan is first obtained for planning and consultation purposes. On the day of the operation, the MRI takes a second scan. The data from this scan is used to plan the positioning information for the robot. After the robot is in position, the doctor manually pushes the needle into the patient, with the robot helping as a guide. A final scan is then taken with the needle in the patient to verify the position.

During the operation, the team witnessed the need for improving timeliness and invasiveness of biopsy procedures. Both of these problems are inter-related, as the entire process took three hours. This was due to needle deflection and the resulting reset time for the robot.

The biopsy needle continuously deflected from the tissue being examined. Each time the needle was set into place, an MRI was conducted to confirm the location of the needle in the tissue. Afterwards, if the needle had missed the area being examined, it was removed, and the process
was re-done. The multiple needle insertions, as well as MRI scans caused the procedure to not only be longer, but also more invasive for the patient.

The team examined how the robot could be modified to bring in data that could be used to help detect needle deflection which will enable future research teams to solve the problem. Furthermore, future versions of this robot have the potential to be fully automated, but currently helps the surgeon in placing and guiding the needle through the tissue. To enable a feedback control system on the robot, a sensing technique is required to measure forces in experienced by the needle.

In needle-based procedures, the surgeon must maintain “tactile force feedback”, monitoring the forces present to adjust the needle accordingly. Using a hands-on approach, a surgeon can adjust the needle based on their observations. With the use of robotics, the surgeon has no physical contact with the needle, and the system must provide this feedback through the use of force measurements. Currently, there is no compact and accurate method to provide tactile force feedback within an MRI environment where these biopsies occur.

Accurate force measurements are necessary to begin to move away from a hands-on approach, and fully realize the potential of robotics in these needle based procedures. Additionally, a robotic system must be able to detect and account for needle deflection, to prevent the needle from missing the targeted area.

The literature review process led us to research done at WPI by a group of undergraduate and graduate students. These papers developed and tested a proof of concept that used optical force sensors to measure forces acting upon a needle. They examined interferometry techniques - the use of light to detect physical changes in a material through interference patterns. As a result, the team decided to also focus on interferometry techniques, to expand and improve upon their groundwork. However, the team also investigated alternate methods that may have been overlooked by the previous research. Regardless of the method used, our team developed three main goals to solve the issues examined during our research. These are:

- Expand on the previous team’s range and accuracy.
- Set the ground work for detecting needle deflection.
- Build a PCB board that can be used in the next version of the robot.

The team chose to use an iterative approach throughout the course of this project, testing and designing a single aspect of the project, and expanding on that design based on the results. This allowed the team to reach each of the previous goals individually, and combine the results into the final product. By reaching each of the previously mentioned goals, the new system
will have potential to be further developed to detect needle deflection, and integrated into an automated robot, to conduct more accurate, and more effective biopsies.
Chapter 3

Literature Review

The team first began by examining literature, as well as previous projects, relating to MRI, biopsies and possible sensing techniques in MRI environments. Throughout the following section, the team presents its findings, building a foundation for the topics discussed throughout the entire report.

3.1 Basics of MRI

To begin to understand the motivation behind this project, we must first examine a few key concepts relating to magnetic resonance imaging, or MRI. MRI is a technique that uses a magnetic field and radio waves to create detailed images of organs and tissue within the body. The first magnetic resonance scanning machine was invented by Raymond Vahan Damadian, and tested in 1977 [4].

However, the use of a magnetic field and radio waves limits the materials that can be used within and in the vicinity of MRI equipment. The biggest limitation is that ferrous materials which consist of iron, cannot be present around the machine, as the MRI induces a strong magnetic field during operation. As a result, only non-ferrous materials may be used to construct any equipment to be used during an MRI. Additionally, the equipment must not affect the MRI image quality in any way. Finally, conductive, non-ferrous materials must also be avoided, as the magnetic field will cause electrical noise and distortion within the system.

MRI guided biopsies are an advancement over the traditional biopsy. During a traditional biopsy, the tissue to be examined is removed via a biopsy needle that is hand guided by a surgeon. However, without an imaging system, the surgeon may remove tissue from the incorrect area. An MRI guided biopsy expands on this concept, while adding the previously mentioned
imaging system. By taking an MRI scan after the needle has been positioned, the surgeon is able to examine the position of the needle and make adjustments accordingly [5].

One major drawback to this method is the time needed to conduct an MRI each time the surgeon misses the area to be examined, and must remove the needle and attempt again. However, this drawback is also in the process of being mitigated through the use of MRI compatible robots. While there are limitations to MRI compatible robotics, there has been progress in the field. For example, a robot developed in collaboration with WPI and Boston’s Brigham and Women’s Hospital [6] has allowed doctors to look at real-time MRI images, while simultaneously controlling the robot to perform a biopsy procedure. This allows for a less invasive procedure which is far more precise than ultra sound imaging or a blind biopsy.

3.2 Robot Assisted Biopsy Procedure

The previously stated goals of this project fall in-line with the current advancements of MRI compatible robotics. By developing an MRI compatible force sensor, the forces acting upon the needle during an autonomous biopsy procedure can be measured, allowing for detection of skin and tissue barriers, as well as needle deflection and bending. Using this information, a robot performing a biopsy procedure could compensate for needle deflection, deflect a needle to reach a target area, and precisely determine when the target area has been reached by the needle.

3.3 Needle Force Profile

Before examining the types of sensing techniques applicable in an MRI setting, it is important to examine the forces present during a biopsy needle insertion. The total force acting on a needle during insertion can be characterized as the total of three forces: the stiffness force, the friction force and finally, the cutting force. The stiffness force occurs prior to the needle puncturing the capsule, while the friction and cutting force occur after the needle has punctured. These three forces are to be discussed in the following paragraphs.

According to Allison M. Okamura et al, biological tissue is linearly elastic for small deformations. However, a second-order polynomial form must be used to model larger deformations that occur during a needle puncture. During their experiment, in which they punctured two bovine livers, they found that the maximum stiffness force before puncture averaged 2.3040N as needle depth 16.65mm. After puncture, the average stiffness force dropped to 0.65N. It is important to note that each of these measurements had large standard deviations of 0.82N and 0.65N respectively. Okamura et al. attribute these large standard deviations to the variability between material properties when dealing with organs[1]. It is important to note that the stiffness force needed
to puncture through soft tissue varies depending on a variety of factors. For example, in the study conducted by Okamura et al. while the average force before puncture was 2.3040N, in some tests, a bovine liver was punctured with as little as 1.5N [1]. In a separate test, conducted by the Department of Radiation Oncology at Thomas Jefferson University by Yan K. et al., the stiffness force required to penetrate a prostate varied from approximately from 5 Newtons to slightly under 12 Newtons. This was the highest example of stiffness force acting upon a needle during a prostate biopsy our team managed to find while conducting research[7].

Additionally, the team examined the book “Medicine Meets Virtual Reality 2001: Outer Space, Inner Space, Virtual Space” by James D. Westwood. In the book, Westwood states that the max force needed to penetrate through muscle is approximately 12 Newtons, as well [8].

Max stiffness force can also vary depending on the location in which the needle is inserted. In the previously mentioned example test, conducted at Thomas Jefferson University, varying where the needle was inserted also altered the max stiffness force. For example, when inserting the needle into a “peripheral zone” on the prostate, one prostate experienced a max stiffness force of 5.5N. However, when the needle was inserted into a “central zone”, the max stiffness more than doubled, having a value of just under 12N [7].

Due to the large variance in the stiffness force exerted on the needle, extra care must be taken to determine the maximum value our sensor can detect. Despite this large variance, the axial forces measured during needle insertion typically produce similar curve shapes, with the primary difference being the max stiffness value of the curve. The following figure displays several examples of a typical needle insertion force curve, created by Okaumura et. al. during their tests [1].

![Figure 3.1: A typical needle insertion force curve from Okaumura [1]](image)

The second force examined by Okamura et. al. was the friction force along the length of the needle [1]. They attribute this force to three primary factors: Coulomb friction, tissue adhesion,
and damping. Coulomb friction can be defined as the friction which occurs as the needle slides against tissue. Tissue adhesion occurs when the punctured tissue grips the needle due to friction forces. Finally, damping, or viscous friction, are the friction forces between the needle and fluids within the punctured specimen.

By using a robot, Okamura et. al. moved a needle in a sinusoidal motion, with a set frequency and speed. The needle was cycled through three different locations on a 25g liver with a known thickness [1]. The drawback to this method is that, although a friction model was developed using this information, it relied heavily on controlling the density of the tissue, as well as the amount of tissue coming into contact with the needle. In a surgical environment, the density of the tissue may be unknown, or may not be as consistent as in this model.

The final force examined by Okamura et al. was the cutting force. The cutting force is defined as the force necessary for the needle tip to slice through the tissue, after the needle has punctured. In Okamura et al. experiments, they found that the average cutting force was 0.94N, which was calculated by deducting the friction force from the total measured force after puncture. It is important to note that this average did not include the additional stiffness force created due to internal collisions with vessels. According to Okamura et. al, as a result of these internal collisions, the measured cutting forces were not constant, and increased as the insertion depth of the needle increased [1].

The research done by Okamura et. al. also pointed to two other factors that affect needle insertion forces: the diameter of the needle, as well as the type of tip on the needle. Needles with a larger diameter had significantly larger force slope, regardless of the tip type. Triangular tips had the lowest force, bevel had intermediate forces, and cone tips had the highest amount of force acting on them. Okaumura et al. attribute this difference to the number of sharp edges present on the needle. The number of sharp edges on the needle correspond directly to the cutting forces applied to the needle [1].

The forces studied in the Okaumura et. al. experiments along with needle diameter and type play a significant role in causing needle deflections during a biopsy procedure. There is an inverse relationship between the diameter of the needle and deflection; as the diameter of the needle increases, the needle deflection increases, and the forces acting on the needle increases. However, bevel tip needles deflect much more, due to their asymmetric structure. Okamura et. al. raise the question as to why the other needles also deflect, despite their symmetric structure. They attribute these deflections to random density variations in the silicone used to test the needles, rather than the needles themselves [1].

It is important to mention that the model created by Okamura et. al. was tested using ex vivo testing. As a result, the model cannot take into consideration several factors that may alter the forces acted upon the needle including: temperature, bleeding and vascular pressure. According
to Okamura et al., additional in vivo testing would be necessary to improve the applicability of the model [1].

3.4 Force Progression Through Secondary Media

Another important aspect of the project was examining available sensors which would not only accurately collect data, but would be safe to use in an MRI environment. Traditional sensors change an electric property, such as resistance or capacitance to cause a measurable change in electromagnetic properties, such as current or potential. MRI compatible sensors, however, need a secondary medium to change before affecting an electric or magnetic field. For example, light or radiation could be used as a medium in an MRI compatible sensor. This medium transmits data about the measured quantity by altering its properties. These altered properties are then measured by traditional means using electronics, such as a photodiode. With this idea, the team investigated previous research in light, pressure, and radiation as potential secondary media.

However, before the team examined fiber optic sensors themselves, the team first needed to learn more about fiber optics in general. A fiber optic cable is a cable that is capable of propagating light. Fiber optic cable can be split into two types: single-mode fiber and multi-mode fiber. In single-mode fiber, only a single-mode, or ray of light can propagate down the fiber. In multi-mode, the light travels via multiple propagation paths at once, due to the larger core. However, this also has the potential to create a noisy signal, depending on the length of the fiber [9].

Giru Rajan stated in his 2014 book, Optical Fiber Sensors: Advanced Techniques and Applications, that optical fiber sensors offer many advantages which are of interest to the team [10]. Optical fiber sensors are resistant to Electro Magnetic Interference (EMI) and are environmentally rugged while allowing for high resolution and bandwidth. However, optical fiber sensors are more costly [10, p.3].

According to Rajan, Optical fiber sensors can be classified into three main types, depending upon how light is modified within the sensor, or the sensor’s “working principle”. The three main types are light intensity, phase, and spectrum, also known as polarity. [10, p.4]. Each of these types are discussed below with their associated sensing technology.

Intensity sensors use changes in received brightness of light to a photo sensor and are further split into four subcategories: reflected, evanescent and macrobending/microbending types [10, p.5].

The reflected types use changes in the intensity of reflected light from a surface to detect changes either in distance or as an indirect measurement of another quantity such as pressure [10, p.5]. While intensity based setups are simple, this simplicity comes at a cost. Rajan wrote, however,
optical fiber bending, coupling misalignment, source power fluctuation, etc., can cause signal attenuation and signal intensity instability [10, p.5].

The second type of intensity sensor uses evanescent fields to function. In this method a part of a fiber is usually changed so that the core is able to come into contact with the measured material. Changes in the material’s index of refraction change the amount of light passed by the fiber [11].

The final two types of intensity based sensors are macro-bending and micro-bending sensors. The two types function in a similar fashion, with each altering the intensity of light depending on the bending of the sensor/fiber. The primary difference between the two types is the scale at which bending occurs. On macrobending sensors, the fiber is bent an amount that can be seen by the human eye, while microbending sensors measure microscopic bending[12].

Phase modulated fiber optic sensors use interference of separate beams of coherent light. The use of interferometers within these sensors allows for extremely small measurements to be made. Rajan stated that “Mach-Zehnder, Michelson, Fabry-Perot and Sagnac are among the most popularly used devices.[10, p.6] These designs work in very similar modes, but change how and where an interference pattern is generated. Of interest to this project are the Fabry-Perot and Mach-Zehnder Interferometers.

A Fabry-Perot interferometer uses two mirrors. The first mirror is semi reflective, while the second is totally reflective. Changes in the distance between these two mirrors induces a change in the reflected phase. There are two types of Fabry-Perot sensors: Intrinsic and extrinsic. In an intrinsic FPI sensor, the sensing cavity, the space between the two mirrors, is a single mode optical fiber. In an extrinsic cavity, this single mode fiber is replaced with an air gap or another material. For the sake of this project, the team focused on extrinsic FPI sensors, as all of the companies examined by the team, such as FISO, sell extrinsic FPI sensors. In addition, if extrinsic FPI sensors are used, the team would also need to select a material to place in between the cavity [13].

In a strain application, the length of the cavity is changed proportionally to the strain applied to the sensor. As a result, the following equation can be derived to relate the strain to the intensity of light at a single given point:

\[ I = I_1 + I_2 + 2I_1I_2\cos\Delta\phi \]

In this equation, I is the intensity at a given point, while \( I_1 \) and \( I_2 \) are the intensities of the two individual waves, each being reflected by one of the two FPI mirrors [14].

A Michelson interferometer functions with a similar principle, where a single beam of light is split and reflected towards two mirrors at fixed locations. These two beams are both reflected
back through the beam splitter, towards a detector, where their amplitudes are combined to create an interference pattern. If one of the mirror’s angle is changed relative to the beam-splitter, the interference pattern produced will change. This is due to the "path difference", or difference between the length traveled, of the two beams of light.

The path difference between the two beams of light can be represented in the following equation:

$$OPD = 2d \cos(\theta) + \frac{\lambda}{2}$$

(3.2)

Where $d$ is the distance between the two mirrors, theta is the angle at which the observer looks at the incoming light, or the angle at which one of the mirrors is turned, and lambda is the wavelength of the light source[15]. Using this equation, one can find the optical path difference with a given angle, and thus, the phase difference between the two waves of light, using the following equation:

$$\Delta \phi = 2\pi \lambda \Delta x = k \Delta x$$

(3.3)

where $\Delta x$ is the phase OPD between the beams of light. With the phase difference, one can predict the intensity of the light in a given area. As a result, a sensor can make use of Michelson interferometry techniques by altering the angle of one of the mirrors, while leaving the other fixed, producing a change in light intensity as a result[15].

A Fizeau interferometer functions by splitting a laser via the use of a beam splitter, where half of the laser is directed towards the detector directly, and the other half is directed towards a reference material. The laser is then reflected from the reference material back towards the detector. By comparing the laser to the reference material, an interference pattern can be made. Fizeau interferometry can be combined with other interferometry techniques to create interference patterns which are easier to analyze [16, p.289].

Changes in the spectral properties of light within a fiber can also be used to measure external changes. The wavelength of the light is modified by an outside factor. An example of one of these sensors are Fiber Bragg Grating [10, p.6]. Fiber Bragg Grating sensors use special fiber that has different sections of fiber with varying indexes of refraction along the length. Changes in the index of refraction result in a corresponding change in wavelength reflected. The index of refraction can be influenced by a number of physical properties of the sensor, including strain and temperature.

However, the cost of Fiber Bragg Grating can be upwards of 20,000USD, specifically for the spectral analysis equipment, as well as the optical source needed [2]. As a result, the team chose not to look into this method further due to the limited budget of the project.
Su et al. found that fiber optic sensors, specifically sensors that utilize Fabry Perot Interferometry, are superior to foil strain gauges when used inside of an MRI machine. Significant advantages were found in terms of resolution and interference from electromagnetic interference. However, the authors note that additional evaluations were needed within an MRI environment to examine performance and durability. Furthermore, the lack of optical shielding on the device led to departures from the expected theoretical model and cycles need to be counted for higher resolutions[17].

### 3.5 Previous Projects

The concept and much of the work for this project are based on a proof of concept project explored by WPI students previously. The project aimed to prove that Fabry Perot interferometry could be used as a viable, robust, and accurate way to measure forces. In the project, the authors explored the basis behind Fabry Perot Interferometry and used a Fabry Perot strain sensor to create a force sensing application [2].

The project team was able to measure the light intensity at a single point on the pattern output by the FPI strain sensor using a photodetector. As the sensing element was strained, the intensity of light would change over a period as predicted. They attached the strain sensor to a cantilever beam, and strained the beam using differing amounts of force. Using known forces, they were able to show the relationship between force applied and the output light intensity. Thus, they were able to prove that the FPI strain sensing element could be viably used as part of a force sensor.

One drawback to the project team’s solution of measuring the light intensity at a single point is identical intensity outputs for multiple strain inputs. Due to the cyclical pattern of the FPI’s output, the light intensity output by the sensor repeats as the FPI sensor is strained. As a result, multiple strain values result in identical voltage outputs. As a result, the device has a much lower dynamic range. This can be seen in the following figure, in which the group maps the strain of the previous design to the light intensity seen by the photodetector [2].

Another project, conducted by Shang et al., expands on this concept. Here, a Fabry Perot interferometer strain sensor was placed onto a flexure, in order to detect needle insertion forces when conducting biopsy procedures. While the design expands on the previous project by integrating the sensor into an MRI-compatible needle placement robot, it also carries over the flaw of having multiple strains map to a single output voltage, similar to the design mentioned above [14].
Despite the drawbacks in the previously mentioned designs, the FPI strain sensing element is immune to electromagnetic fields, and its usefulness as a force sensor had already been proved, it is an ideal candidate for use in MRI compatible force sensing.
Chapter 4

Project Strategy

Before the team could begin developing project designs, the team first needed to determine the goals and limitations for the project. Aside from the sources presented in the literature review, the team also conducted interviews with the project’s advisors, as well as graduate students from WPI’s AIM Lab, to help specify concrete goals for the project. Additionally, the team traveled to Boston, in order to examine and learn more about a biopsy procedure. While there, the team spoke to the lead engineer at the Surgical Planning Laboratory, and gained additional insight to the project.

These people served the role of customers for the project, to determine primary deliverables. From initial interviews with these customers, the team developed the following client statement: "The goal of the MRI Compatible Optical Force Sensing project is to develop a force sensor that is both MRI compatible, and can detect different layers of tissue depending on inputs."

Through continued interviews, the team established four main goals for the project: establish a platform to apply the proof of concept from the previous MQP, improve on both the accuracy and range of the previous design, make the system MRI compatible and finally, develop a system that is functional with the current biopsy robot.

Additionally, constraints were also developed throughout the course of the project. First, the device needed to be compatible with all types of surgical needles, otherwise the price of the project, and any products to come out of the project, would be significantly more expensive. Second, the device needed to be minimally invasive, for patient safety reasons. Finally, the team needed to ensure that the device was easy to use for doctors. If the device took a significant amount of time to set-up, doctors would most likely avoid using it, and continue to use previous biopsy methods.

As the team worked on medical equipment, the team determined that doctor and patient safety was also of vital importance to the project. Finally, individual abilities within the group, as well
as the monetary cost of parts and lead time were also considered, in order to meet the project’s strict deadline.

With the previous mentioned constraints and goals in mind, the team began to refine the initial client statement, developing a new client statement to reflect the goals and constraints of the project. The final client statement is as follows: "The main purpose of the MRI Compatible Optical Force Sensing Project is to develop a force sensor that primarily expands on the previous MQP’s work, improving the accuracy and range of the device, miniaturizing the device, and making the system MRI Compatible. To implement the device in a medical setting, the device must also be safe for both patient and doctor, easy to use, and compatible with all surgical needles, as well as the biopsy robot”.

The team then began to develop a plan of action to accomplish the established objectives, while considering the constraints. The team determined that the most efficient way to examine possible solutions was to divide them into two primary categories: Interferometry techniques and other techniques. Possible solutions were divided in this manner as the previous MQP group focused primarily on interferometry techniques and did extensive research in the area. As a result, the team chose to also focus on interferometry techniques, but did not want to eliminate the possibility of a superior solution to the problem.
Chapter 5

Alternative Designs

In order to ensure that the best solution was found, other conceptual designs had to be considered and evaluated. The team examined the different types of fiber optic force sensors, as well as other techniques available to develop the most optimal solution to the problem. In the following section of the paper, a brief summary of some of the designs will be presented, concluding with an analysis comparing all of the developed conceptual designs.

5.1 Needle Driver Force

The first design the team developed was a device that would apply a constant force to the base of the needle. From here, the speed of the needle would be measured, to determine when the needle has passed through different layers of tissue. While the design was conceptually simple, the team decided not to pursue it as it would most likely not function properly during a real trial. This is primarily due to varying densities within a single tissue layer, as well as needle deflection, which may vary the speed of the needle.

Furthermore, the team would need to design a device to move the needle at a perfect constant velocity, which would be costly and impractical to the scope of the project. Additionally, the device would most likely restrict needle movement for the surgeon. For example, if the surgeon needed to rotate the needle, the devices readings may be affected. As a result, the team chose not to pursue this idea.

5.2 Friction Force Needle

The second design developed by the team was measuring the friction forces present on the needle, in order to determine whether or not the needle has deflected. As a deflected needle
Alternative Designs

would theoretically have more points of contact with the tissue compared to a non-deflected needle, needle deflection is theoretically possible with this method. However, the team chose not to pursue this design for two major reasons. First, as tissue is not uniformly dense, the detection method may not work in practice. Second, this would require a custom made needle, going against one of the original goals of the project, and driving the cost of the project up significantly. Finally, this design is not feasible because it requires the force sensor that is the main focus of this project to have already been developed.

5.3 Radioactive Tracking

Drawing inspiration from a branch of nuclear medicine, the team discussed a design concept that could use radiation produced by nuclear decay of an element to track the position of the needle. Radioactive isotopes are used in applications such as imaging and destroying cancerous cells. For example, gamma rays from Iodine-123 is used to image the thyroid gland and study any irregularities in the organ. The ability of gamma ray to pass through the human body allows it to be tracked from the outside of the body [18]. Using a similar principle, the team developed a design which involved embedding a radioactive component inside the tip and along the length of the needle. This would allow the needle to be tracked from outside using similar equipment to that of thyroid imaging.

This design concept would allow real time tracking of the needle position and form. However, the design requires a device to image the needle by measuring the radiation from the radioactive element. This separate device would also need to be MRI compatible and be able to process data in real time. Moreover, while choosing the radioactive element, it would be necessary to investigate its impact on human tissue. The biggest drawback of this design was that a custom needle would have to be designed and manufactured which was neither feasible nor possible in the time-line of the project. Nevertheless, this design has potential if further developed.

5.4 Fiber Bragg Grating

The team also considered the use of Fiber Bragg Grating to measure the forces acting upon the needle. As stated previously, in the Literature Review section of this report, Fiber Bragg Grating functions by changing the index of refraction within a fiber by influencing a physical property of the fiber, such as temperature and strain.

In the team’s design, a fiber Bragg Grating strain sensor would be attached to a fixture, with known strain properties. The fixture would subsequently be attached to the base of the biopsy gun and the fixture and sensor would strain depending on the force applied to the needle. The
strain sensor would then output different wavelengths of light, depending on the force applied to the needle. A spectrometer would be used to detect this change in wavelength, and calculate the corresponding force.

However, there were also several negatives in regards to a Fiber Bragg Grating design solution. As stated in the Literature Review chapter, Fiber Bragg Grating sensors are expensive, and not within the budget of this project without additional funding. Furthermore, Fiber Bragg Grating sensors are temperature sensitive, and as a result, temperature would need to be taken into consideration when using a Fiber Bragg Grating strain sensor. Finally, extra care must be taken when handling Fiber Bragg Grating, or any other type of fiber optic sensor, as they are extremely fragile.

5.5 Fabry Perot Interferometer - Single Sensor

Similarly to the proposed Fiber Bragg Grating design, the team also considered using a Fabry Perot Interferometer strain sensor. Conceptually, the design is identical to the Fiber Bragg Grating design, where the strain sensor is attached to a fixture with known strain properties. The fixture itself is again, placed at the base of the biopsy gun, and the force acting upon the needle can be determined based on the strain of the fixture, and thus, the sensor.

Aside from the technical differences between the two sensors, examined during the literature review, there are also some differences. A Fabry Perot Interferometer’s output is independent of the temperature of the sensor. Furthermore, the cost of a single Fabry Perot Interferometer is significantly less as compared to the Fiber Bragg Grating sensor.

While the device improves on some design flaws present in Fiber Bragg Grating, it also has several drawbacks. While less expensive than Fiber Bragg Grating, the cost of an FPI strain sensor is still fairly high, at 250USD (FISO). Furthermore, similar to most fiber optic solutions, the sensors are not extremely durable, and care must be taken to not damage the sensor.

5.6 Fabry-Perot Interferometer - Multiple Sensors

To further improve on the previously discussed FPI design, the team also proposed to further improve on the design by making two major modifications to the fixture. First, three FPI strain sensors would be attached to the fixture, allowing the strain to be taken at three separate points, spaced evenly around the fixture. Second, the fixture itself would be moved onto the needle directly. The combination of these two points would theoretically allow for needle deflection to be detectable. As the needle deflects in one direction, the FPI sensor closest to the direction the
needle is deflecting should exhibit more strain. Using this, it should be possible to examine the output of each individual FPI sensor, compare each, and determine if the needle is deflecting, and in what direction.

The addition of two more FPI sensors presents additional challenges and drawbacks. First, the cost of the project would increase due to the additional sensors. Second, the team would need to attach the designed fixture directly to the needle in order to examine needle deflection forces. While not challenging in a lab environment, attaching the fixture directly to the needle means the team would also need to consider the sterilization of the device, if it were used in an actual medical procedure. Finally, determining deflection based on each sensor’s output would be more computationally complex compared to determining force in a single direction. While not impossible, this means the team would need to dedicate more time to examining the forces acting upon the needle, and how the forces are translated onto the fixture.

5.7 Pressure Sensing

Pressure sensors were also evaluated to measure the force applied to the needle at a distance, in a similar setup to the FPI. The needle would be secured to a plate which would press against the tube. The tube itself is sealed by a cap on one end and a pressure sensor on the other. NXP’s MPXV5004GC7U pressure sensor reads small (max 0.57 PSI) changes in ambient pressure, so the tube itself did not need to be pressurized. Figure 5.1 shows the bench set up for testing the pressure sensor. This sensor was powered using a lab bench supply and the analog output was read by an oscilloscope.
Figure 5.1: Experimental setup of the Pressure sensor.

However, leakage over time and the stiff nature of pneumatic tubing caused concern. The idea was abandoned. Development of a more flexible leak proof tube make this an interesting area for research.

5.8 Design Option Analysis

The following chart is a design option analysis for all of the potential solutions to the project brainstormed by the team. Each design option was compared based on the criteria established from the design goals and restrictions: monetary cost, invasiveness, ease of use, part lead time, functionality, required knowledge to construct, durability, responsiveness, and finally time to build. Each of these criteria were weighted differently, depending on their importance. For example: the team determined that a solutions functionality was more important than the part lead time; as a result, functionality was weighted more heavily.

Each presented design received a score of 1 to 3 in the previously mentioned criteria, with a 1 being the lowest score possible in a respective category, and a 3 being the highest. A breakdown for the scoring system for each individual criteria can be viewed in appendix A. The results of the comparison can be seen in table 5.1.
The criteria upon which each design was scored is listed below. Further details on each criteria can be found in Appendix A.

1. Cost
2. Invasiveness
3. Ease of Use
4. Part Lead Time
5. Functionality
6. Required Knowledge
7. Durability
8. Responsiveness
9. Complexity

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<th>3</th>
<th>4</th>
<th>5</th>
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<th>8</th>
<th>9</th>
<th>Total</th>
<th>Weighted Total</th>
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<td>35</td>
<td>57</td>
<td>71</td>
<td>100</td>
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<td>3</td>
<td>3</td>
<td>3</td>
<td>3</td>
<td>1</td>
<td>2</td>
<td>22</td>
<td>1243</td>
<td></td>
</tr>
<tr>
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<td>2</td>
<td>3</td>
<td>2</td>
<td>1</td>
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<td>3</td>
<td>1</td>
<td>3</td>
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<tr>
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<td>2</td>
<td>1</td>
<td>2</td>
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<td>3</td>
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<td>3</td>
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</table>

Using the results, the team determined that the most effective course of action would be to pursue a design with photodiode arrays, as well as multiple FPI sensors. However, due to the budget of the project, the team chose to purchase a single FPI sensor, and develop a fixture that would allow additional FPI sensors to be attached in the future.

5.9 Other Options Not Considered

Several novel options were found during the literature review that would require significant research and development, namely optical coherence tomography and slow light. Optical coherence tomography allows for deformations in tissue to be imaged at small depths [10, p.475]. Proof of concept technology that slows the speed of light down also exists, which would theoretically allow for more accurate resonance in fiber optic sensors [10, p.525]. Rajan believes that
slow light has little to offer outside of research, stating, “[...] in the sensing context, slow light is something of a distraction though of course it may be useful in packaging research proposals” [10, p.525].
Chapter 6

Design Verification and Validation

6.1 Basic Principle Overview

In order to meet our primary requirements, we designed a system that was able to measure axial forces that could fit inside the current version of the BRP robot and the components inside the control box. The basic principle behind our design is that the force experienced by the needle strains a custom fixture which holds the optical sensor, a FISO FOS-N strain sensor. A laser of a specified wavelength is used to send light to the FPI sensor. Based on interferometric principles, the FPI sensor outputs a pattern that changes depending on the forces experienced by the fixture. The pattern is focused onto a linear photodiode array which consists of two sub arrays of 128 pixels allowing us to record a section of the output pattern in terms of intensity. Analog data from the array goes through a filter after which the data is converted to digital signals using an ADC. The data is processed on a custom PCB which uses an FPGA and microcontroller to run an algorithm to analyze and calibrate the data. The details of every part of the design will be further elaborated in this section. Figure 6.1 shows an overview of the entire system.

Throughout the course of this project, the team followed an axiomatic design process. Each customer need presented in the project strategy portion of the report was translated into design parameters to be tested within the design verification section of this report.
6.2 Fixture Design

The most significant mechanical component of this project is the fixture onto which the primary optical sensor is attached. The fixture will be placed behind the needle in the needle driver module of the BRP robot as seen in figure 6.2. The FPI sensor by itself, is extremely fragile and requires attachment onto a flexure to function as a strain sensor. Therefore, the fixture was
design. During the process, several constraints had to be acknowledged in order to achieve a practical blueprint.

6.2.1 Physical Constraints

- **FPI Strain Range:** The FPI sensor that was chosen for the design has a specific strain range, from -1000\(\mu\text{e}\) to 1000\(\mu\text{e}\). As a result, the flexure portion of the fixture also needed to have a strain from -1000\(\mu\text{e}\) to 1000\(\mu\text{e}\) within the maximum range of forces encountered. The force range that was required for the application was determined through research and study of various force profiles which resulted in a range of 0N to 12N with 3N overhead. Therefore, the fixture had to be designed in a material that could produce the defined strain range with a linear relationship with the chosen force range.

- **Thickness and Size:** The size and thickness of the fixture was constrained by the next design of the BRP robot. A maximum of 60mm diameter and a maximum thickness of 4mm was available in the robot to place the fixture. This led to two designs of the fixtures shows in figure 6.4a and 6.4b, 58mm OD and 51mm OD respectively. The thickness of
the fixture was experimentally changed to achieve the desired strain-force relationship. Simulations were done in Solidworks to study this relationship which resulted in a fixture thickness of 2.43mm.

- **Material**: The material of the fixture was defined by the sensor’s functioning environment. Since one of the goals for the project was MRI compatibility, the fixture had to be made out of a non-ferrous material. After considering common plastics, it was determined that Delrin 2700 NC010 was both strong, flexible and easily machined.

- **Modularity**: The design of the fixture had to be simple enough so as to allow for future modifications if necessary. Since the one of the reach goals of the project was to be able to measure the deflection of the needle, the fixture had to be designed in such a way that the addition lateral detection could be done with the same or similar fixture without compromising the primary goal of measuring axial forces.

### 6.2.2 Design Process

The main goals of the project that were addressed with the fixture were, (1) miniaturization of the sensor, (2) ability to measure axial forces, (3) consideration of deflection forces. Through research the team found that while the needle is being inserted into the tissue, it has the potential of deflection in any direction depending on the forces it experiences. Due to this, a general circular shape was chosen for the fixture. A finite element analysis was performed on a simple flexible disc to observe the strain experienced by the disc due to forces. In order to see the effect of an axial force, a force vector of 5N, normal to the surface of the disc was applied, the resulting strain on the disc is shown in figure 6.3a. The effect of Non-Axial force was studied by applying a force at an angle of 45deg from the surface of the disc, the resulting strain is seen in figure 6.3b. Through this initial study, the team concluded that a circular shape for the fixture would be the best as it would produce strains that are equal in a given radius when the force is axial. Moreover, it also proved that a circular shape would help detect deflections in the needle as it would apply a force at an angle on the fixture which creates a strain pattern that is close to an ellipse with one of the foci on the center of the fixture. Since the strains observed in the 4 quadrants of the circle were different, the direction of the non-axial force could theoretically be determined by examining each quadrant.

From this hypothesis, the team decided to have multiple sensors distributed in the disc to be able to measure axial forces accurately and allow for deflection forces to be detected. The team conducted simulations with 4 designated places for sensors and then 3 designated places for sensors. It was observed that direction of needle deflection could be optimally determined by using 3 sensors. Moreover, using 4 sensors did not provide more information than 3 sensors
which is redundant. This decision resulted in 2 designs, design 1 6.4a and design 2 6.4b seen in figure 6.4.

In order to evaluate between the two fixture designs, extensive studies and simulations were conducted. Mainly, with respect to strain resolution, range and size.

6.2.3 Analysis of Fixture Design 1

This fixture has three spokes of width 5.93mm and length 23.5mm. The outer diameter of the fixture is 58mm. Additional dimensions can be seen in the engineering drawing shown in Appendix C.1. The length of the spoke was determined by the minimum length required to place the FPI sensor and the outer diameter was restricted by the space available in the robot to place the fixture (60mm). The fixture was machined out of Delrin on the CNC. Each spoke individually act as a beam fixed on one end. As seen in figure 6.5b, the maximum displacement
of 3mm occurs where the force is applied - denoted by red. Therefore, the simple beam formula can be used to approximate the strain and deflection. The FPI sensors will be placed inside groves on the spokes with the glass rod closest to the center of the fixture. This set up will align the sensing cavity to the region where maximum deflection of the needle will occur allowing the strain from the fixture to be transfered linearly onto the FPI sensor.

![Beam representation of a single spoke and how it responds to force](image)

![Displacement visualization](image)

**Figure 6.5:** Behavior of the spokes in design 1

When a force (F) is applied, it causes the spoke to deflect. This deflection creates a combination of tension (top) and compression (bottom). The force (F) also causes the length of the spoke to change ($\Delta L$), which is used to calculate the strain ($\epsilon$). The maximum deflection caused by the force is denoted by $\delta$. Material and geometrical properties like the Young’s modulus (E) and moment of inertia (I) of Delrin play an important role in calculating $\delta$.

\[ \epsilon = \frac{\Delta L}{L} \]  
\[ \delta = \frac{F L^3}{3EI} \]

This fixture is designed to be placed on the front part of the needle. The needle is attached onto the fixture with a 3D printed collar-like part shown in figure 6.6a which is tightened on the needle and inserted into the defined holes in the fixture as seen in figure 6.6.

![Collar to hold the needle](image)

![Needle, Collar and Fixture setup.](image)

**Figure 6.6:** Design 1
The principle behind this design is that the collar holding the needle moves with the needle and uses the force experienced by the tip of the needle to push against the fixture and bend the three spokes causing the spokes to strain and deflect as seen in figure 6.5b. The strain force relationship for 0-15N for this fixture is shown in 6.7. This data was obtained from simulations conducted in Solidworks.
From the data we can see that the fixture has a lower range with higher resolutions. The maximum displacement of the spokes was close to 3\text{mm} when a force of 15\text{N} was applied. Even though the relationship was linear, the strain exceeded the maximum allowable (1000\text{\mu e}) at 9\text{N} of force. Moreover, from a practicality standpoint, this fixture would have to be placed on the sterile part of the needle due to which it would have to be sterilized every time it was used. Since the FPI sensors are fragile, this would not be ideal. The bending radius of the FPI sensors coming out of the spokes would also have to be very tight to be compact inside the robot. The OD of the fixture is only 2\text{mm} smaller than the space available making it extremely tight and no room to work with. Therefore, this design was not preferred.

6.2.4 Analysis of Fixture Design 2

In an attempt to resolve the issues encountered in design 1, an improved design was made. This second design also allows 3 FPI sensors to be placed on the fixture. However, the configuration and arrangement of the 3 FPI sensors differ to that in design 1. Mechanically, the behavior of the spokes in design 2 can be approximated to a beam fixed on both ends and the force being applied on the center of the beam (spoke in this case). As seen in 6.8a, the contact point on the spoke is the center of the spoke, causing the maximum strain in this region. This aspect of the design gives more stability to the fixture with a maximum deflection ($\delta$) of 1.2\text{mm} when a force of 15\text{N} is applied. The thickness of each spoke is 4.5\text{mm} and the length is 31.5\text{mm}. Additional dimensions are elaborated in Appendix C.2. The FPI sensors will be placed inside grooves on the
spokes with the glass rod closest to the center and the fiber flows away from the fixture trailing off from the groove. This set up will align the sensing cavity to the region where maximum deflection of the needle will occur allowing the strain from the fixture to be transferred linearly onto the FPI sensor.

![Beam representation of a single spoke and how it responds to force](image1)

![Displacement visualization](image2)

**Figure 6.8: Behavior of the spokes in design 2**

When a force (F) is applied at the center of the beam, it causes the spoke to deflect. Figure 6.9d shows a visualization of how each spoke behaves, maximum deflection is denoted by red. It is seen that in the simulation each spokes act as a beam fixed on two ends with the maximum deflection of 1.5mm in the center. The center of the beam can be defined as half of the total length of the beam \( \frac{L}{2} \). This deflection creates a combination of tension (top) and compression (bottom). The combination of tension and compression result in a change in the total length (\( \Delta L \)) of the beam. The ratio of \( \Delta L \) to original length (L) is defined as the strain. Material and geometrical properties such as the Young’s modulus (E) and moment of inertia (I) of Delrin play an important role in calculating \( \Delta L \).

\[
\epsilon = \frac{\Delta L}{L}
\]

In order to find the deflection (\( \Delta y \)) along any point of the beam (x) the following equation can be used (x-axis is along the beam and Y axis is perpendicular to one edge of the original beam as seen in figure 6.8a).

\[
\Delta y = \frac{Fx}{48EI} (3l^2 - 4x^2)
\]

The Maximum deflect at the point of load can be calculated as follows.

\[
\delta = \frac{F l^3}{48EI}
\]

Another important feature of this design is how it interacts with the needle. The design consists of an intermediate part seen in figure 6.9a which gets inserted into the designated contact points on the fixture. This part ensures that the fixture can be placed on the front section of the needle,
similar to design 1. Additionally, the fixture can also be placed behind the needle with the handle part of the needle pressing against an intermediate structure as seen in figure 6.9c.

![Intermediate Part](image)

(A) Intermediate Part

![Intermediate Part inserted into the fixture.](image)

(b) Intermediate Part inserted into the fixture.

![Position of fixture in the BRP Robot](image)

(c) Position of fixture in the BRP Robot

![Displacement Visualization in Design 2](image)

(d) Displacement Visualization in Design 2

**Figure 6.9: Design 2**

The intermediate part of the fixture can act as a collar which can be fastened onto the needle upon which the part moves with the needle and uses the forces experienced by the needle to push against the fixture causing a deflection visualized in 6.9d. However, as mentioned in the previous section, this collar-like set up would require the fixture to be sterilized after every use. In order to avoid this, the team decided to change the configuration of the sensors to allow for a bigger diameter of the center. This increase in diameter allowed the intermediate part to be of OD 15mm which in turn increases the surface area onto with the biopsy needle handle can be pressed onto. This set up is seen in 6.9c. Although, the contact points on the fixture remain the same, the deflection caused by the needle pushing onto the intermediate part will be the same as in figure 6.9d. This allowed the team to study the strains similarly for both mounting cases. The 3 holes on the outside allow the fixture to be mounted onto a fixed part in the BRP robot.
To ensure that design 2 met the requirements of producing strains between -1000µε and 1000µε, a finite element analysis was done for forces from 0N to 15N. The ideal relationship from Solidworks can be seen in 6.10.

![Simulation of Design 2 - Strain (µε) Caused by Force (N) (Figure 6.10)](image)

**Figure 6.10:** Ideal force-strain relationship of design 2 from 0N-15N.

From the initial simulation data, it was concluded that design 2 provided a better strain-force relationship than design 1. As seen in 6.10 design 2 can withhold forces up to 13N without crossing the allowable strain of 1000µε. The -1000µε would be almost identical with the graph reflected along the x-axis. In addition, design 2 can be placed on the front of the needle or behind the handle of the needle to measure axial forces. However, by placing the fixture behind the handle, the needle would not be able to transfer the deflection forces due to additional fixtures supporting the needle. Therefore, the team decided to test the deflection forces by placing this fixture on the front of the needle. Since the OD of design 2 is only 51mm which is 9mm less than the space available in the robot, this gives room for the fibers coming out of the FPI sensor to be safely managed without forcing a small bending radius to hold them as this could damage the fiber.
Comparing designs 1 and 2, it was concluded that design 2 had more pros than design 1, with the only drawback being that with the current placement inside the robot, non-axial forces will not be detected. Design 2 allows the deflection forces to be measured when placed at the needle. Considering these options, the team conclusively picked design 2 over design 1 to be tested and implemented into the BRP robot. Figure 6.11 shows the entire robot design by AIM Lab doctoral candidate, Marek Waternberg along with the possible placement of design 2 of the fixture seen in white. From a practicality standpoint, placement of design 2 in the robot would not require it to be sterilized after every use as it will be under the sterile blanket and not directly interacting with the needle.

Figure 6.11: BRP Robot along with design 2 of the fixture
6.3 Fixture Design Testing

As the FPI sensors are extremely fragile and expensive, the team first wanted to test our design concept using strain gages. The strain gages were placed on individual spokes of the needle as it would allow the strains experienced by each needle to be monitored separately as seen in figure 6.12. The strain gages were each part of a separate Wheatstone bridge, each connected to an instrumentation amplifier. One of the three identical circuits is shown in figure 6.13. The experiment serves as a proof of concept, measuring both axial and non-axial forces using strain gages attached to the fixture, in place of the FPI sensors. This allowed the team to measure the forces at each spoke of the fixture, without potentially damaging the FPI sensor.

![Strain gauges placed on the fixture in order to measure strain due to force.](image)

Figure 6.12: Strain gauges placed on the fixture in order to measure strain due to force.

6.3.1 Equipment

The following equipment was used throughout this experiment:

- (3) Strain gauges
- (9) LM348N Operational Amplifiers
- (21) 1K ohm resistors
- (3) 120 ohm resistors
- (3) 23 ohm resistors
- (1) Power Supply
- (3) Multimeters

From here, the team developed three separate Wheatstone bridge circuits, one for each strain gauge on the fixture. This allowed the output voltage of each circuit to be proportional to the strain applied to each spoke. Each bridge consisted of three fixed resistor values, two 1K ohm resistors and a 120 ohm resistor, as well as one strain gauge. The 120 ohm resistor was chosen as the strain gauge had a value of 120 ohms when under no strain.

In addition, each Wheatstone bridge was connected to an instrumentation amplifier circuit, to amplify the output and allow for comparison between each circuits output. Each amplifier circuit consisted of six 1K ohm resistors, a 23 ohm resistor and three LM348N operational amplifiers. The 23 ohm resistor is the gain resistor in the circuit, referred to as $R_{gain}$.

One of the circuits, including the Wheatstone bridge as well as instrumentation amplifier, can be seen in figure 6.13, modeled in Multisim.

A multimeter was used to measure the voltage output. The change in voltage is caused by the change in resistance of the strain gage. The change in voltage can be converted to a strain value by using the properties of the strain gage. The strain gages used, were from Vishay Precision Group of value 120ohms and a gage factor of 2.075 as stated in the spec sheet.
The wheatstone bridge used in the circuit is a quarter bridge as displayed in 6.14. A balanced wheatstone bridge is that in which $R_1R_3 = R_2R_4$ resulting in a $V_o$ to be equal to 0. However, in this case, since one of the 4 sensors is a strain gage, the resistance changes, changing $V_o$. Under these conditions, the bridge would be considered to be unbalanced and the resulting $V_o$ can be calculated using equation 6.8. $R_{3o}$ refers to the unstrained resistance of the strain gage. $V_s$ refers to the supply voltage which was 9.86V.

$$V_o = V_s \frac{\delta R \times R_1}{(R_2 + R_{3o} + \delta R)(R_1 + R_4)}$$

(6.6)

The denominator is simplified by recognizing that:

$$\frac{\delta R_3}{R_{3o}} << 1$$

(6.7)

resulting in:

$$V_o = V_s \frac{\delta R \times R_1}{(R_2 + R_{3o})(R_1 + R_4)}$$

(6.8)

Using the derivation of the relationship between strain and voltage from [19, p. 1] the following equation can be used to calculate the strain ($\epsilon$), where $G$ is the gage factor of the strain gage.

$$\epsilon = \frac{V_o(R_2 + R_{3o})^2}{V_sGR_2R_{3o}}$$

(6.9)

In this case $R_2 = R_{3o}$, therefore:

$$\epsilon = \frac{4V_o}{V_sG}$$

(6.10)

### 6.3.2 Axial Forces

To begin, the team first attached three strain gauges to the fixture. The strain gauges are micro resistors, with the resistance value of each gauge changing slightly depending on the strain...
applied. While attaching the strain gauges to the fixture, extra care was taken to ensure the placement of the gages was identical to the placement of the FPI sensors. This would allow us to measure the force on each spoke identically between the FPI sensor and strain gauge.

Next, we developed a stand to hold both the fixture and the needle. The stand allowed a hook to be attached to the needle to hold the weights, and ensure that all spokes on the fixture would have an identical amount of force placed on them. A picture of the set up can be seen in 6.15.

Weights were then added to the hook in increments of 20g, up to 680g, with the output voltage of each circuit recorded as additional weight was added. Using the equation 6.8 and equation 6.9, these output voltages were then converted into strain measurements placed on. Once 680g of weight was placed onto the needle, the data points were then plotted on a graph displaying the weight placed onto the needle Vs. the strain of each spoke of the fixture. The results of this experiment can be found in the Discussion chapter of this paper, in section.

The team also conducted a second experiment to examine the effects of forces placed on the needle at varying angles. To do this, the stand was modified to hold the needle horizontally, with the fixture still attached to the base of the needle. From here, 100g of weight were added to the needle, in increments of 20g. As the needle bent, the force placed on strain gauges varied, allowing the team to observe how needle deflection affected the fixture by observing the output voltage of each circuit.
Once 100g and all data was recorded, the weights were removed and the fixture was rotated 90 degrees. By rotating the fixture, the needle deflection forces were placed on a different portion of the fixture, allowing the team to compare how different angles of deflection affect each spoke of the needle. The experiment was then repeated with the fixture at 90 degrees from the starting point, and again at 180 degrees from the starting point. The strain of each spoke, calculated using 6.8 and 6.9, was then graphed compared to the weight applied to the needle. The results of this experiment can be found in the Discussion chapter of this paper, in section 7.1.
6.4 Fiber Optic Design

6.4.1 Physical Constraints

The optical design is constrained by the physical characteristics of the fixture, controller box and by monetary considerations. The fixture was designed and fabricated to fit the strain conditions typically seen during prostate biopsy, while utilizing the 1000με sensor. FISO states in the data sheet of the FOS-N strain sensor that the resolution of the FPI is 0.01% of the full range. Therefore, the 1000με strain sensor provides the highest resolution compared to other strains available (3000με and 5000με).

The controller box limited the size of optics used. The project was budgeted two PCI Express slots within the control box. Two PCI Express slots approximately allows for 70mm by 210mm by 130mm in usable space. However, a TP-Link fiber router is located in the corner which takes up some space. After measuring the box physically, a usable space of 40mm by 210mm by 130mm was determined. Lasers can be incredibly expensive to purchase, especially the scientific grade. A red pigtailed laser diode LPS-635FC from Thorlabs was available but was determined to be malfunctioning due to its significantly lower than expected and unstable power output. The laser was determined to be broken via testing. The output hovered around 300μW, which is significantly lower than the expected 2.5mW [20]. The decision was thus made to purchase a 100mW 650nm red laser pointer to see if a cheaper alternative was viable. Upon inspection, the laser provided a stable and strong source of light, and thus was sufficient for the project.
6.4.2 FPI Sensor Specifications

The FISO FPI sensor uses interferometry techniques to measure the strain of a material. This strain can then be used to calculate the force applied to the system. The Fabry Perot sensor consists of a single-mode optical fiber of OD, 210 micrometers, through which light propagates into a glass tube consisting of two semi-reflective mirrors. The distance between the two mirrors is referred to as the sensing cavity ($L_{cavity}$) which changes depending on how much the sensor is strained. As the strain range of the sensor is $-1000\mu e$ to $1000\mu e$, the sensing cavity changes between 8000nm to 23000nm respectively. The mirrors are fixed onto two fused glass elements which along with the sensing cavity make up the gage length of the sensor which can be seen in fig 6.20 mentioned in the spec-sheet provided by FISO. The sensing cavity length when the strain is equal to 0 is 15500nm. The mathematical relationship between the change in phase of the interference pattern and the strain can be expressed with the following equation:

$$\delta\phi = 2\pi \frac{\epsilon_{xx}L_{cavity}}{L_{gage}}$$ (6.11)

Figure 6.20: FISO Force Sensor.
This equation was defined by Lemay et. al [2]. Here, $\Delta \phi$ represents the change in phase, while $\varepsilon_x x$ is the strain of the sensing cavity. These values are related by the initial, unstrained cavity length, $L_{cavity}$, and the length of the gage, $L_{Gage}$.

### 6.4.3 Lens Setup

The smallest commercially available option to split the laser and FPI pattern was to utilize a 2 by 1 optical coupler. Compared to a typical beam splitter, a fiber coupler is a much smaller size, due to the lack of a splitting cube and supporting collimation optics. The fiber cores are fused together which allows modes of light to transfer between themselves. In this use case, a 2x1 coupler was chosen. Laser light is coupled to the input of the coupler via a 10x 0.25 numerical aperture lens and fiber holder from Thor Labs. The entire piece is supported by the board. The laser light exits the single ended port of the coupler and passes through an FC to ST connector adapter to reach the FPI. The reflected, phase changed light passes back through the adapter connector and coupler in the opposite direction. One half of the reflected light is passed to the laser port and the other half is projected onto the linear array.

![Optical Setup Diagram](image)

**Figure 6.21: Optical Setup Diagram**

### 6.4.4 Interference Change

Another aspect of the fiber optic design is the overall possible change in the output interference pattern. The interference pattern output is directly related to the distance traveled by the light rays and the wavelength of the light. Since the design of the FPI strain sensor is to change the distance traveled by the light rays as it is strained, this causes the output interference pattern to change. Each time the change in distance, $\Delta D$, is even divisible by the wavelength, the output interference pattern will be identical. If this is considered one cycle, or $2\pi$ radians, it is possible to determine the number of pattern cycles between two FPI output strains.

$$ R = \frac{2\pi \Delta D}{\lambda} $$  \hspace{1cm} (6.12)
Since the zero strain cavity length of the FPI is 15500nm, and the size at maximum strain is 23,000nm the maximum $\Delta D$ is 7500nm. Using the 650nm wavelength, the maximum possible change in radians can be calculated using the following equation.

$$R = 2\pi \frac{7500 \text{(nm)}}{650 \text{(nm)}} = 72.498 \text{rad}$$  \hspace{1cm} (6.13)

When this is combined with the designed force range of the fixture of 0-15N, the ratio of force to interference pattern change, $FR$, can be determined.

$$FR = \frac{15 - 0 \text{N}}{72.498 \text{rad}} = 0.207 \text{N/rd}$$  \hspace{1cm} (6.14)

If multiplied by $2\pi$, this can also be displayed as 1.3N/cycle. This calculation has important implications for development of the algorithm and certain aspects of the timing circuitry.

### 6.5 Circuit and PCB Design

Analysis by hand of FPI data collected by instruments provides an opportunity to gather data on force profiles and needle bending in the laboratory environment. In the clinical setting, however, force feedback to the robot needs to run in real time without the need for human intervention. After a calibration stage, the force feedback becomes a key part in achieving...
autonomy for the BRP Robot. This goal necessitates circuitry and computation in order to automate reading, analyzing, and sending the FPI light output.

6.5.1 Purpose and Constraints

Three main tasks were identified to be implemented by the circuit. The primary task of digitizing the FPI light pattern resulted in the general topology and dictated the rest of the design. Second, the pattern needed to be analyzed by an algorithm to convert the digitized readings into force. Finally, the BRP robot needed to interface with the circuitry in the control box.

The team designed and constructed a circuit that allows for the light data to be collected, filtered, processed and transmitted in real time. When considering the design for the circuit, several constraints were identified. The circuit needed to fit the physical dimensions of the robot controller box and interface with it. Safety and accuracy are the top concerns for the project, and present several constraints. A circuit with unstable and thus unpredictable results cannot be used in medical procedures. In order for the robot to effectively use the FPI, the circuit and computation required results to be obtained quickly. For development purposes, the circuit needed to have testable and thus verifiable sections. The team decided to build a custom circuit instead of off the shelf components to fit these constraints. Figure 6.23 shows a diagram of the circuit. Each part is discussed in detail below.

![Circuit Diagram](image)

**Figure 6.23: Circuit Diagram**

6.5.2 Light Sample Collection

Several technologies exist to collect light data. The team reviewed several options, such as photodiodes, commercial and medical cameras. Photodiodes were immediately identified as the best solution, as the other options do not fit the characteristics of the pattern obtained from
the FPI. Cameras provide excellent two dimensional color captures of light. The FPI uses a monochromatic laser, thus negating the need for color. In addition, the pattern repeats in a sinusoidal fashion, so only one dimension of data provides sufficient information.

A myriad of options exist for photodiode implementations. The previous research used a single photodiode to measure intensity [14]. These diodes are bulky and provide only a single point of data, neglecting secondary changes in the pattern from the interfering wavefronts.

A series of diodes enables for all parts of the pattern to be obtained. In order to achieve the miniaturization goal, the series of individual diodes needed to be placed in close proximity. Very small photodiodes exist, but the solder pad size induces a limit on the density that they can be place on a printed circuit board. Thus, the team focused on single chip solutions that provide the diodes in an array on a single integrated circuit. The TSL1402R from AMS was selected for its high density, large number of photodiode pixels and good response characteristic to red light. The sensor provides a 400 DPI resolution, which nears the limits of human eyes [21]. A higher resolution or smaller sensor is undesirable, as the optics are aligned to the array by hand.

6.5.3 TSL1402R Photodiode Array Functionality

The TSL1402R works on the premise of integrating current (op-amp capacitor configuration) from each photodiode and then storing that charge in a secondary capacitor. The array uses a unique control scheme that dictated how the rest of the components were chosen. The sensor uses a shift register to multiplex the analog output, creating a serial stream of voltages that correspond to each pixel. A main clock input provides timing for output and sample collection. To read the values, a pulse is sent to the start integration (SI) pin. At this time, the integration capacitors for every pixel are connected to their corresponding storage capacitor. With every clock cycle, the multiplexer connects the storage capacitor to an output buffer so that a voltage can be read on the analog output. On the next cycle, the SI pulse shifts so that the multiplexer connects the subsequent pixel. The previous pixels storage capacitor then drains to ground to prepare for the next sample. The integrating capacitor takes some time to dissipate its charge. The dissipation takes place while reading the previous samples values which takes 17 clock cycles. Upon the 19th clock cycle, a new integration period begins and stops when the next SI pulse is encountered. A more indepth discussion about the control of the chip and timing considerations is in section 6.7

The linear array can operate up to 8MHz, but sampling at this speed is unnecessary. A sample rate of 200kHz was chosen, which results in a frame rate of 1550 frames per second (200000/129). Since data on tissue deformation and tearing speeds are not available, the team inferred the needed rate from a video of a slow motion water balloon popping(6.24). Although bizarre,
balloons are made of elastic material that deforms and then rips upon puncture, similar to the behavior of skin. A frame-rate of 1500 fps was determined to be enough to see the material rip and thus the forces changes along with it[22].

![Figure 6.24: Balloon popped at 1500 frames per second](image)

The authors encourage watching Bryce Fritzel’s video which can be found on the following Youtube page[22]. The area of interest begins at 11 seconds. [https://www.youtube.com/watch?v=RpyW3yHn5nU](https://www.youtube.com/watch?v=RpyW3yHn5nU)

### 6.5.4 Analog to Digital Conversion (ADC)

Analog to digital converters have a limit for measurements in changes in an electrical signal. Since the measurements regarding the actual voltage values of the diodes are compared in a relative manner, the accuracy of the converter does not have a significant impact. The ADC chosen provides a resolution of 12 bits, which provides 4096 levels of measurable intensity. Since the optics can be easily adjusted to brighten or reduce the intensity of the fringe pattern, the resolution also has negligible affect on sampling ability.

The sampling speed of the ADC sets the limit of how fast data can be collected and transmitted to the FPGA. However, the signal from the linear photodiode array can be very different from pixel to pixel due to the pattern induced by the FPI. The 128 values are fed into the ADC at a rate of 200kHz. Neighboring photodiode pixels have similar voltage values and thus realistically cause the acquired signal to have a low frequency spectrum. Each clock cycle (blue) in figure 6.25 shows a switch from the previous pixel output (yellow) to the next. The consecutive pixels do not have large changes in voltage. This property is important as it allows for the analog output to be at a distance from the filter and ADC. If this was not the case, a second PCB board would be needed to filter and digitize the readings to be sent via differential pair, or the array would need to be mounted on the main PCB itself. This key property is verified in Appendix B.2. However, the pattern itself from the FPI can change significantly from frame to frame or
within a frame. The design attempted to mitigate this possibility by having a maximum of one change occur during the course of reading frame.

The output of the actual optics setup is significantly more complex than a typical ring pattern, due to the choice of using a 2x1 coupler instead of a beam-splitter or the spacing distance that
exists between the FPI fiber end and the control box connector. This decision was made in order to meet the miniaturization goal from the experimental setup. The team believes that mode changes in the coupler and refraction from the connector air gap cause the increase in complexity.

The BRP Robot is being developed iteratively and concurrently with the project, so the laser and optics may need to be modified in the near future to interface with the robot or fit in a future version of the controller box. Changes in the optics will inevitably cause changes in the FPI pattern. The analysis circuitry needs to be able to handle such a change, so that only changes in software and logic are necessary. Furthermore, a secondary non-optical sensing board was in concurrent development. From a design perspective utilizing the same ADC brings several advantages in terms of code reuse, abstraction and debugging.

6.5.5 Analog Filter to ADC

The Nyquist-Shannon Sampling Theorem states that analog to digital converters need to sample at twice the maximum frequency than the highest needed frequency. Any higher frequencies can not be reconstructed and will show up as lower frequencies during analysis. This phenomenon is called aliasing and causes artifacts of the higher frequency signal to change the actual desired signal. These higher frequencies need to be filtered out. The design utilizes a low pass Butterworth filter in the four pole Sallen Key topology. The Sallen Key topology has a very linear passband area with a steep drop off at the cutoff point [23, p.691]. Cascading two of these filters provides a good balance of performance vs cost and number of needed components. The values are discussed in detail in figure 6.27. The expected maximum frequency is 200kHz which dictates the need for a cutoff frequency of 400kHz. In order to accommodate future changes in the optics, this was expanded to a cutoff frequency of 500kHz. This cutoff is also suitable for the non-optical sensing board.

The four pole filter has two stages that results in a fourth order polynomial [24]. The two stages were made independently using a Q factor of 0.5412 for the first stage and a Q factor of 1.3065 for the second [25, p.9]
Filter transfer functions should be in the from of \([26, p.7]\)

\[
\frac{V_{\text{out}}}{V_{\text{in}}} = \frac{1}{s^2 \frac{2}{\omega_0} + \frac{1}{Q} s \frac{2}{\omega_0} + 1} \quad (6.15)
\]

Where:

\[
s = j\omega \quad (6.16)
\]

Nodal analysis shows that the following is true:

\[
\frac{V_a - V_{\text{in}}}{R_1} + \frac{V_a - V_b}{R_2} + \frac{V_a - V_{\text{out}}}{\frac{1}{sC_1}} = 0 \quad (6.17)
\]

Since the operational amplifier can be treated as ideal, the voltage at the non-inverting input is equal to the output.

\[
V_b = V_{\text{out}} \quad (6.18)
\]

The voltage at point Va is

\[
V_a = V_{\text{out}} \frac{\frac{1}{sC_2} + R_2}{sC_2} \quad (6.19)
\]

\[
V_a = V_{\text{out}} \left[ \frac{\frac{1}{sC_2} + R_2}{sC_2} \right] sC_2 \quad (6.20)
\]

\[
V_b = V_a = V_{\text{out}} \left[ 1 + sC_2 R_2 \right] \quad (6.21)
\]
Since enough terms are present, the nodal analysis equation can be solved via substitution.

\[
\frac{V_{\text{out}}(1 + sR_2C_2)}{R_1} - \frac{V_{\text{in}}}{R_1} + \frac{V_{\text{out}}(1 + sR_2C_2)}{R_2} - \frac{V_{\text{out}}}{R_2} + \frac{V_{\text{out}}(1 + sR_2C_2)}{sC_1} - \frac{V_{\text{out}}}{sC_1} = 0 \tag{6.22}
\]

which expands to:

\[
\frac{V_{\text{out}}}{R_1} + \frac{sV_{\text{out}}R_2C_2}{R_1} + \frac{V_{\text{out}}}{R_2} + \frac{sV_{\text{out}}R_2C_2}{R_2} - \frac{V_{\text{out}}}{R_2} + sV_{\text{out}}C_1 + s^2V_{\text{out}}C_1C_2R_2 - sV_{\text{out}}C_1 = \frac{V_{\text{in}}}{R_1} \tag{6.23}
\]

\(V_{\text{out}}\) is common to all terms on the left side of the equation, \(sV_{\text{out}}C_1\) cancels and \(R_2\) drops from the fourth term.

\[
v_{\text{out}}\left[\frac{1}{R_1} + \frac{sR_2C_2}{R_1} + sC_2 + s^2C_1C_2R_2\right] = \frac{V_{\text{in}}}{R_1} \tag{6.24}
\]

The equation can further be manipulated into the standard transfer function form.

\[
|H(f)| = \frac{V_{\text{out}}}{V_{\text{in}}} = \frac{1}{R_1} \cdot \left[\frac{1}{\frac{1}{R_1} + \frac{sR_2C_2}{R_1} + sC_2 + s^2C_1C_2R_2}\right] \tag{6.25}
\]

Factoring \(\frac{1}{R_1}\) from the denominator, allows the term to cancel outside of the fraction in order to reach the simplified transfer function.

\[
|H(f)| = \frac{V_{\text{out}}}{V_{\text{in}}} = \frac{1}{1 + sR_2C_2 + sR_1C_2 + s^2C_1C_2R_1R_2} \tag{6.26}
\]

The \(s^2C_1C_2R_1R_2\) term converts to the standard \((\frac{s}{\omega_0})^2\) form.

\[
s^2C_1C_2R_1R_2 = \left[\frac{s}{\sqrt{C_1C_2R_1R_2}}\right]^2 \tag{6.27}
\]

\[
\omega_0 = \frac{1}{\sqrt{C_1C_2R_1R_2}} \tag{6.28}
\]

Factoring \(s\) from \(sR_2C_2 + sR_1C_2\) brings out the \(Q\) characteristic in terms of \(\omega_0\).

\[
sR_2C_2 + sR_1C_2 = s\left[R_2C_2 + R_1C_2\right] = \frac{s}{Q\omega_0} \tag{6.29}
\]

\[
s\left[R_2 + R_1\right]C_2 = \frac{s}{Q\omega_0} \tag{6.30}
\]

The final analysis form is thus obtained.

\[
|H(f)| = \frac{V_{\text{out}}}{V_{\text{in}}} = \frac{1}{\left[\frac{s}{\sqrt{C_1C_2R_1R_2}}\right]^2 + s\left[R_2C_2 + R_1C_2\right] + 1} \tag{6.31}
\]
To simplify the design, resistors $R_1$ and $R_2$ are chosen to be the same. $\omega_0$ can now be substituted to find $Q$, and the R values are equal, changing equation 6.29 in order to find the capacitor values.

$$2RC_2 = \frac{1}{Q} \frac{1}{\sqrt{C_1C_2R_1R_2}}$$  \hspace{1cm} (6.32)

$$Q = \sqrt{\frac{C_1C_2R^2}{2RC_2}} = \sqrt{\frac{C_1C_2}{2C_2}}$$  \hspace{1cm} (6.33)

$R_1$ and $R_2$ were set to 1000Ω. As stated above, the cutoff frequency desired is 500kHz and the first stage Q factor is 0.5412 while a Q factor of 1.3065 is used for the second. This creates a system of equations to solve for each capacitor.

$$\omega_0 = 2 \times \pi \times F_c$$  \hspace{1cm} (6.34)

$$\left\{ \begin{array}{l} 2 \times \pi \times F_c = \frac{1}{\sqrt{C_1 \times C_2 \times 1000 \times 1000}} \\ Q = \frac{\sqrt{C_1C_2R^2}}{2RC_2} \end{array} \right.$$  \hspace{1cm} (6.35)

Substituting for stage 1:

$$\left\{ \begin{array}{l} 2 \times \pi \times 500000 = \frac{1}{\sqrt{C_1 \times C_2 \times 1000 \times 1000}} \\ 0.5412 = \frac{\sqrt{C_1C_2 \times 100^2}}{2 \times 1000 \times C_2} \end{array} \right.$$  \hspace{1cm} (6.36)

$$C_1 = 344.539pF$$  \hspace{1cm} (6.37)

$$C_2 = 294.078pF$$  \hspace{1cm} (6.38)

Substituting for stage 2:

$$\left\{ \begin{array}{l} 2 \times \pi \times 500000 = \frac{1}{\sqrt{C_1 \times C_2 \times 1000 \times 1000}} \\ 1.3065 = \frac{\sqrt{C_1C_2 \times 100^2}}{2 \times 1000 \times C_2} \end{array} \right.$$  \hspace{1cm} (6.39)

$$C_1 = 831.744pF$$  \hspace{1cm} (6.40)

$$C_2 = 121.818pF$$  \hspace{1cm} (6.41)

The values of 344pF and 294pF did not have close standard parts. To address this, two capacitors were used in parallel to reach the needed value. Figure 6.28 shows the final layout.
Capacitor $C_1$ was set to 15pF in parallel to $C_5$ with a value of 330pF. Capacitor $C_3$ was set to 820pF and capacitor $C_4$ was set to 120pF.

The theoretical filter (Equation 6.40, Equation 6.41) and filter circuit with available capacitors 6.29 were analyzed using NI Multisim.
Both simulations show that the filter behaves as desired. The exact values simulation 6.30 shows a $f_{3\,\text{dB}}$ frequency of 493kHz. Interestingly, the parts chosen provide a better simulated response with a $f_{3\,\text{dB}}$ frequency of 499kHz (figure 6.31).

6.5.6 Computation and Control

A device is needed to run the linear array and collect the data from the ADC. Multiple solutions to collecting this data exist, including a microcontroller, field programmable gate array (FPGA) and implementation of a soft microcontroller on a FPGA. Due to the strict timing constraints, high speed data and the need to maintain synchronization for good data, the decision was made to use an FPGA. The FPGA runs in real time, with data and control flowing in a non procedural manner. A microcontroller introduces issues with parallelizing the three linear array inputs. The three arrays need to be read at exactly the same time. Relying on interrupt routines to finish servicing could cause the arrays to run at slightly different times.
Since the circuit was designed at the same time as research into the algorithm, a precaution was taken to incorporate general purpose pins to the FPGA so that a secondary processor could be attached. This turned out to be incredibly useful because the FPGA has limited memory and the final algorithm is memory intensive. Future work on the algorithm could decrease the needed memory, allowing for it to run on a soft core processor embedded in the FPGA.

![GPIO and Power for microcontroller](image1)

**Figure 6.32:** GPIO and Power for microcontroller

A series of power rails (-12V, 12V, 5V, 3.3V and Ground) are used to power the laser and external MCU. An external buck converter (DROK) is used to step down the 12V rail to 3.7 volts for the laser. The power rails are protected by 500mA fuses (1812L050PR) from Littelfuse Inc. The laser typically draws 240mA to 270mA of current.

![Fuses for power rails to external MCU and Laser](image2)

**Figure 6.33:** Fuses for power rails to external MCU and Laser

The linear arrays data sheet states that its typical behavior is at 5V. The ADC and filter handle 5V, but the maximum input voltage of the FPGA is 3.95V as specified in the Xilinx Spartan 6 data sheet. A bidirectional logic level converter steps down the SPI MOSI signal to 3.3V, along with stepping up the SI, CS, and clock because the FPGAs max voltage on any pin is 3.3V.

![TXB0108PWR Bidirectional logic level converter](image3)

**Figure 6.34:** TXB0108PWR Bidirectional logic level converter
In addition, an FTDI UART to USB chip was added. This chip allows for the circuit to eventually communicate with a mobile computer and be calibrated while inside the robot controller before the biopsy takes place.

![Figure 6.35: FTDI and USB Type B Connector](image)

Other cards in the robot controller also have FPGAs. The Nexsys 3 board has a very similar chip, so the team chose to use the xc6sl9 as used in the other cards. This allows for code to be shared between the projects with ease. The FPGA needed to maintain code in a non volatile state so that it could be programmed upon startup without the need of a computer. Again, the team used the same PROM, the Xilinx platform flash 04 (XCF04SVOG20C) to accomplish this. The FPGA and PROM are in the same JTAG chain, making programming and verification of integrity easy. This also has the added benefit of purchasing parts in bulk for lower prices. Finally, a tricolor LED was added to maintain consistency with other boards and to provide visual feedback during debugging.

![Figure 6.36: JTAG and LED](image)

### 6.5.7 Printed Circuit Board

The circuit was breadboarded for testing. Upon satisfaction with the performance, the schematic was captured in Altium. The team started with a copy of the BRP Robot’s Motor Driver Card as a reference and template for the location of the connectors and board outline. The initial plan was to keep the components in the same location that are being re-used, but this was abandoned when it became clear that the layout and pins would need to be different. The board had to be manufactured on an accelerated time line, so the original four layer design was changed to a two layer. OSH Park offered a fast turn around while still having gold plating, which is needed as
the pads on the bottom of the board are connectors to the robot controller. These connectors communicate via SPI and implement the interface to the controller. The controller provides several voltage rails to power the components on the board and the laser. The 1.2V FPGA power for the FPGA is a proprietary design from Paulo Caravalho.

Since the board is only two layers, the original approach to shielding the card from EMI by stitching the front and back ground layers together was removed. In order to avoid EMI in this prototype version of the board, the ground plane was split up according to practices identified by e2v corporation [27].
Components were chosen to match the components of existing cards to maintain similarities and reduce costs. In addition, capacitors were placed in parallel whenever possible to increase the number of similar capacitors on the board. This technique allows for parts to be purchased in bulk.
Every power input has decoupling capacitors to shield the power from fluctuations in the power supply and noise. The USB connector is shielded via the shield pins in order to isolate the connector.

![Figure 6.40: A set of decoupling capacitors](image)

A design technique of adding 0 ohm resistors allowed for sub sections to work independently for testing. By adding and removing these resistors, parts can be put on line and taken off with ease.

![Figure 6.41: USB Shield](image)

The power rails were laid out after placing the signals. This was done in an attempt to match the signal path lengths as much as possible. After, the power rails were thickened for heat dissipation in an iterative process until they got close with components and signals.

![Figure 6.42: Zero ohm resistors](image)
Several holes were added to the board to allow for a plastic cover to be added. The holes on the shield fit standard 6-32 standoffs. The laser holder is then glued onto the plastic piece for the final assembly.

The silk screen was made to call out the values and locations of components, along with each section. The gerbers are located in the engineering drawings section, Appendix C.1. The final result is presented in Figures 6.45 and 6.46.
The plastic holder was laser cut using the board outline gerbers and drill files.
6.6 Circuit and PCB Testing

After receiving the board from OSH Park and the components from Digikey, the team began the process of testing and adding parts. The power and ground points were tested using a digital multimeter to check for shorts and the presence of the correct voltage as parts were added.

![The PCB](image1.jpg)

**Figure 6.48:** The PCB

![JTAG chain, PROM programming and FPGA debug success](image2.jpg)

**Figure 6.49:** JTAG chain, PROM programming and FPGA debug success

The PROM, FPGA and JTAG chain were evaluated via a JTAG chain integrity check and boundary scan using Xilinx’s iMPACT software. The PROM was successfully programmed and the FPGA is able to load logic upon startup. The LED currently is set to traverse the red-green-blue spectrum at 50MHz.
The filter was evaluated using an oscilloscope and function generator. The function generator performed a sine wave sweep from 0Hz to 2MHz. The yellow trace is the filter output response and the blue is the function generator input.
The input voltage from the function generator was 2.18V which indicates a f3dB voltage of 1.54V (2.18 * .707). The point was found around 524KHz, verifying the satisfactory performance of the filter.

![Figure 6.53: Filter Cutoff Frequency](image1)

More importantly, minimal attenuation occurred at lower frequencies. The filter works very well at the chosen 200KHz linear array clock.

![Figure 6.54: 200KHz Response](image2)

### 6.7 Digital Circuitry

The project utilized digital circuitry implemented on a field programmable gate array (FPGA). The decision to use an FPGA was based upon the need to accurately control several clock lines and to keep data in sync. The flow of data is easily configurable and changes can be made easily without having to change wires on the printed circuit board. In addition, several of the other cards in the BRP Robot use the same FPGA, allowing for logic to be shared across projects.
The team chose to implement the logic in Verilog as the description language is familiar and Xilinx, the designer of the Spartan 6 FPGA, fully supports the language in their development environment.

6.7.1 Overall Design

The logic is structured with modules that each represent a part of the hardware on the PCB. Connections between these modules is provided in a top module, with the corresponding inputs and outputs plugging into each other. No logic other than routing connections is present within the top module. By keeping modifications of data outside of the top module, the submodules are able to be changed easily and are in control of their interfaces.

The flow of data between the modules was decided early on in the project which provided a road map for development. This design allowed for concurrent work on the submodules by the multiple team members. Testing and simulation of the modules was conducted independently of the other parts, ensuring that final implementation worked as desired. Bugs in the sub modules are mostly contained and should not affect other modules. A final benefit to this design pattern came from the presence of multiple copies of similar circuitry on the PCB. Copies of the modules can be added if another set of inputs is desired, such as a secondary data output or another set of linear photodiode arrays. This was crucial during assembly and testing of the PCB. To avoid damaging components on other parts of the board from shorts and to determine that signals were correctly connected, each portion was brought up independently. The logic allowed for each set to be tested without the need to populate the rest of the board at the same time.

The modules correspond to the following physical circuit parts: Sample Control, ADC, Data Output, LED Control and BRP Robot interface. Additional components, namely: Clock Manager and Sensor Clock Divider correspond to the external oscillator to provide clocking and synchronization. Each of these is now described in detail. As of the time of writing this report, the BRP robot interface is not included, but is available to be incorporated when needed.
6.7.2 Clocks

The 50MHz differential clock is handed off to the Digital Clock Manager. The Manager provides a 50MHz internal clock and 4MHz internal clock which can be used by the modules when needed. These clocks are kept in sync with each other to ensure proper data flow between the various modules.

The Sensor Clock Divider takes the 4MHz clock as an input. Using a counter, the clock signal is divided by 20 to get a 200kHz clock, which is to be used by the linear array. While the Sensor Clock Divider sets the rate at which the sensor is driven, the 4MHz clock controls the actual data acquisition rate. This is due to the 1 to 20 ratio desired between the sensor clock and the ADC clock. The 1 to 20 ratio is desired because 16 clock cycles are required to clock digital data out of the ADC. The remaining cycles are used as a delay, which is required due to the potential for rise time in the analog input voltage. This delay is explained in depth in B.2

6.7.3 Sample Control

The sample control module uses sensor clock and reset. The module synchronizes and controls the start integration (SI) pulse. This pulse is sent directly to connected linear arrays to begin a new sample collection, and internally to synchronize with the other modules. Upon receiving this pulse, the linear array clears the previous light intensity stored on each pixel’s internal
integrator and stores the charge on a sampling capacitor. After 19 clock cycles, the linear array begins to obtain the intensity profile of the current pattern from the FPI via the internal integration circuit, which is to be read during the next collection. The module sets the SI pulse high for 1 out of 129 clock cycles so that the ADC can evaluate the 128 pixels on each sub-array, without the sample changing. The 129th cycle serves two purposes, reading the 128th pixel and integration time. It also allows the previous SI pulse to be cleared from the sensor’s shift register. A pixel’s sampling capacitor is drained after the analog output multiplexer has been shifted to the next pixel. The 128th pixel needs time to transfer the charge from the integrating capacitor to the sampling capacitor before the integration capacitor is cleared of charge on the next SI pulse. Second, the 129th cycle adds to the overall integration time. If the laser light is weak in the future, the Sample Control module can be set to send the SI pulse on a higher number of cycles which would increase the integration time resulting in a larger collection of charge, as long as this time falls within the range specified by AMS. In effect, the cycles after 128 serve as phantom pixels.

6.7.4 ADC Control

The ADC Controller receives serial data from the ADC, as well as the 4MHz internal clock, sensor clock and internal SI pulse. Serial data arrives through a modified SPI bus, with master in slave out (MISO), chip select (CS) and 4MHz clock. Since the ADC does not need to be configured, the master out slave in (MOSI) has been removed. An internal counter keeps track of the sensor clock cycles so that 128 clock cycles after an SI pulse, the chip select is disabled so that other modules only receive the 128 pixels and not the extra phantom/integration time pixels. The linear array goes to a high impedance state after the 128th pixel is read, which could be interpreted as false data if the values were allowed to be read. If one were to increase the number of clock cycles between the SI pulse it would still be functional as it counts to 128 and then starts on the next SI pulse.

Within the ADC Controller, a delay logic circuit helps the ADC acquire an accurate reading of the voltage from the analog input. Due to the potential of rise time between the rising edge of the sensor clock and stability of the analog input, a delay between the sensor clock and the ADC chip select is required. The chip select starts an ADC sample and therefore the delay ensures accuracy. The maximum rise time as found during testing in B.2 is 0.25us. This means that the chip select must be delayed by at least one 4MHz clock cycle. It is currently set to five clock cycles in order to allow the clock to increase up to 20MHZ if needed.

The modified SPI bus works by shifting data in from the ADC. The ADC outputs two bytes as a 16 bit packet for each sample, of which the starting four are zero and the proceeding 12 are the value data. The first byte is the MSB and the second is the LSB for the converted value. Two
shift registers are needed as one ADC is connected to each 128 section of the linear array. Upon a high chip select, a counter is reset. This counter tracks the position of the 16 data bits. After cycle 15 occurs, each register is moved to a secondary storage register so that it can be moved to the FPGA output as a complete integer while new data arrives. The counter continues to 20 before resetting. This is so that the shift register does not continue to shift in data during the excess clock cycles, or while the sensor is in a high impedance state.

6.7.5 SPI Communication

During the development of this project, it became necessary to store large amounts of data from the hardware in order to facilitate the development of the algorithm. The initial solution was a board developed by FTDI, which could interface between SPI communication and USB communication. This allowed for data to be sent to a PC which could store and help analyze the data. This required the creation of a module which could take the parallel digital data from the ADC and send it to the FTDI board via the SPI communication protocol.

The SPI communication module takes three inputs and has a single output. As the FTDI board is designed to be the master in the SPI communication pair, it controls the communication clock and chip select signals. These two signals are inputs for the module. The third input is the 50MHz internal clock. The output is the Master InSlave Out(MISO) communication line. The Master OutSlave In(MOSI) is ignored because no data needs to be sent from the PC to the digital circuitry.

Within the module, several shift registers are employed to synchronize the external clock with the internal signals. While the chip select is enabled, the digital circuitry looks for falling edges of the external clock to shift out data. Since the USB communication reads only a single byte at a time, certain methods are employed to make data reconstruction possible. When the data is brought into the SPI communication module, it is split into the least significant byte(LSB) and most significant byte(MSB). The four leading zeros allow both bytes to be shifted left by a single bit. The least significant bit on the MSB is then assigned to be a one, and the least significant bit on the LSB is assigned as a zero.

After significant testing, it was discovered that, due to the indeterminate nature of the software, the FTDI board could not consistently acquire all of the data required. This method was then replaced by the Saleae Data Acquisition System which does not require specialized communication.
6.8 Digital Circuitry Testing and Simulation

Testing and simulation is important when designing digital circuitry, as the verilog description may not match the behavior intended and the FPGA induces delays due to internal latch and hold times. The team used Xilinx’s ISE environment to create test benches for simulation. The simulation may not be accurate itself, so the Nexys 3 development board from Digilent was employed. The board allowed for quick prototyping from the start while having a very similar FPGA. The Digilent board has a Spartan 6 xc6slx16 FPGA. It is very similar to the xc6slx9 FPGA employed in the final product, but differs in number of look up tables and flip flops, and has a different pin configuration. Moving from the Nexsys 3 to the PCB FPGA only required modification of configuration and a new constraints file. The use of the board allowed the PCB to be designed and fabricated at the same time.

The simulation used test benches that contained simulated inputs. The timing of signals and correct handling of logic was successfully verified. Of specific interest to this report is the timing characteristics of the Sensor Clock Divider, Sample Control and ADC Control.

Sensor Clock Timing

The sensor clock is expected to at 200kHz based on a single clock cycle over 20 4MHz clock cycles. In the simulation shown in Figure 6.56, the sensor clock has a 5μs period and therefore is running at the expected 200kHz. The simulation also shows the 20 4MHz clock cycles between each sensor clock cycle. Based on simulation, the sensor clock functions as designed. When tested on hardware as shown in Figure 6.57, it exhibits the same clock rate within ±0.5%.

![Figure 6.56: sensor clock timing simulation](image-url)
Sample Control timing

The sample control signal is expected to rise for a single sensor clock cycle every 129 sensor clock cycles. The simulation in Figure 6.58a shows that the timing between sample control pulses is 645μs. When divided by the 5μs period of the sensor clock, the result is 129 which is the correct number of clock cycles. Figure 6.58b shows that the sample control pulse is only high for a single clock cycle, and is not changing during the rising edge of the sensor clock.
All of these behaviors are the designed behaviors. During hardware testing, the signal showed identical behavior to the simulation. This is shown by the oscilloscope captures in Figure 6.59.

![Oscilloscope captures](image)

**(A)** scope of sample control timing  
**(B)** timing of a single sample control pulse

**Figure 6.59: sample control oscilloscope results**

**ADC Control Timing**

The ADC control module is a complex module that requires precise timing for data sampling and shifting to occur correctly. Of specific interest during simulation are the timing between the sensor clock and the chip select, and the correct output of data. As discussed earlier in this chapter, there must be a delay between the rising edge of the sensor clock, and the rising edge of the chip select. As shown in Figure 6.60, the 4.5 4MHz clock cycle delay behaves as expected. This figure also shows the transition of data from the serial input to the parallel output registers. Since the initial value of the `data_storage` register is identical to that of the `Parallel_Data_Out` register after the transition, it is confirmed that data is successfully shifted into the module.

As the simulation confirmed that the ADC control module functions properly, it can be inferred that it will function on hardware as long as the clock timings are still correct. Figure 6.61 shows
that the 4MHz clock, sensor clock, and chip select signals are all timed correctly when run on hardware.
Top Module Simulation
Since the clocks work correctly, it can be assumed that the rest of the logic works correctly if data is able to accurately flow from the ADC to the output. The testbench sent incrementing values at the ADC input on each Sensor Clock edge. The output shows the same value.

The data path is clearly shown in the figures in 6.62. The data starts in the simulated next_data register. This register is used to modify the simulation data being fed into the tested main module. The data is then fed into the data_storage register which is used to shift data into the module’s serial data inputs. Once the data is entirely shifted in, the data is then shown in the Data1 output, indicating that the data has been successfully input into the FPGA.

The project was synthesized and programmed to both FPGAs. The sensor behaves as expected with 129 outputs upon clocking per SI pulse. The ADC transmits 128 sets of 2 byte values per SI pulse.
6.9 Algorithm

The incoming pattern from the FPI needs to be processed by an algorithm to convert the changes in phase (fringes) into a force reading. The algorithm needed to provide accurate, repeatable results over multiple fringes that could be calibrated to the strain reading.

6.9.1 Development

As discussed earlier, the team had very little experience using optics, with knowledge obtained from discussions with Prof. R.S. Quimby, Prof. Cosme Furlong-Vasquez and others at WPI. Due to this, the team took the advice from Prof. Furlong to design the optics from a standpoint of utilizing math to get parts into their general positions. The process known as walking the beam was used while connecting the parts. This brute force process involves adjusting optics by physically moving them one at a time and monitoring the resulting intensity in order to find an optimal configuration. As a result, the team realized the algorithm would change a lot and would require calibration after adjustments.

The team collected initial data using a beam splitter, laser and optical table. During this process, the first FOS-N FPI was found to be unresponsive. A second was ordered and received from FISO. However, after a few days of collecting data, FISO representatives emailed the team stating that a production mistake had been found in a batch of sensors. With the new sensor included in the affected batch, a third FPI was obtained and bonded to the fixture.

Data from the output from the second sensor showed that the fringes do indeed change in a repeatable and predictable pattern, matching the teams understanding of the optics and FPI. In order to see the changes in the pattern effectively, a video was taken of the output and processed using Adobe Premier Pro. A differential analysis was performed by subtracting frame n from frame(n-1). The difference was amplified by applying a threshold filter. Figure 6.64 shows two
frames from this analysis with fringes present. The complete analysis video can be found at: https://www.youtube.com/watch?v=DlNC_iGsGGs

The linear array was set in front of this pattern and data was collected. The readings from the ADC were uploaded to Excel for analysis. Again, the signal was filtered by subtracting frame n from frame(n-1). To remove ambient light the average value of each pixel was removed during each linear array read. The fringes can be seen by highlighting any values that are above zero. Figure 6.65 shows the movement of fringes across the array, along with a change in direction as the FPI moves from compression to tension.
The need for miniaturization led to a coupler being used, leading to the end design of the algorithm being changed considerably. The coupler uses evanescent fields to propagate the modes of light from one fiber pair into another [10, p.45]. During this process, each output becomes a trigonometrical component of the original phase of the electromagnetic wave [10, p.44]. The team believes that this process is responsible for removing the fringes seen in the earlier setup, as only the sine/cosine portion of the intensity is received.
6.9.2 Background

The data obtained from the sensor needs to be processed before analysis can be performed. The signal may be noisy due to variations in the laser power, laser wavelength and coherence length [10, p.60]. Noise can also come in the form of vibrations on the fixture and from the optics. Furthermore, electrical noise presents itself on the linear array and the ADC circuitry. The signal needs to be reconstructed accurately before it can be used to find the phase difference. The ADC filter helps with this problem, but further digital signal processing is utilized to transform the input into a usable signal. Rajan classifies three methods of changing the incoming signal, "[...]phase-generated homodyne detection, fringe-rate measurement and homodyne detection methods" [10, p.58]. Phase-generated homodyne detection, also known as pseudoheterodyne detection, was of little use to the project because it requires the use of modulating the laser wavelength [10, p.58]. The fringe-rate method proved to be more applicable and was thus used. The fringe-rate method counts fringes as they pass the detector output over time, and the result is integrated to find the phase difference [10, p.58]. The two previous methods suffer from limits in resolution due to ambiguity in phase at $\pi$ radians. The homodyne method combines the two in order to remove this limitation and can be accomplished in several different ways such as extra couplers and actuators [10, p.58]. All methods need to count fringes as there is ambiguity in the wave number at $2\pi$ radians.
6.9.3 Principal of Operation

The algorithm works similarly to the ideas behind quadrature phase shift keying and differential Manchester encoding [28, pp. 113-120]. A discrete derivative is found by subtraction. This step also removes common ambient light from each sample. By tracking the order in which maximums and minimums in intensity are encountered, the state of phase change can be found. These four states of phase (Quadrature) correspond to:

- 00: From Minima, Intensity Increasing
- 01: From Maxima, Intensity Decreasing
- 11: From Maxima, Intensity Increasing
- 10: From Minima, Intensity Decreasing

From this point, the differential is integrated over time, which pseudo-performs the frequency recovery mentioned above. The resulting value is the current force on the FPI. If the same maxima or minima is encountered twice in a row, the FPI can be said to be in a state of switching direction of force, e.g. changing from compression to tension. In this case, the negation of the next values are integrated. Table 6.1 walks through how this works as a hypothetical fixture goes through tension and then is released causing compression as it moves back to steady state. This hypothetical situation causes two series of fringes to pass over, causing 4\pi radians of phase change in each direction. The switch occurs between the fourth and fifth peak event. Notice that the change in force direction is effectively encoded into the first bit.

6.9.4 Implementation

The algorithm was implemented in C to run on a softcore processor in the FPGA or on a microcontroller. A laptop computer and logic analyzer was used for most of the development to simplify memory constraints and allow for fast changes. Figure 6.67 shows the general layout of how the data is handled.
The team also built a Processing V3 utility 6.68 to visualize the waveforms. This allowed for playback of the changes in intensity and filtering either real time or in slow motion. For the current version of the project, the software also serves as a calibration tool to set maxima and minima values. At the time of writing this paper, extensive modifications are being done in an attempt to improve the algorithm. The primary goal is to regain the fringes seen earlier in the process. The fringes may be present but drowned out by noise or shifted in phase by the coupler. The current algorithm finds the average intensity over the entire array, causing it to suffer from ambiguity at $2\pi$ radians. Although unlikely, tension could switch to compression during a peak event thus causing the output to be completely incorrect. If the structure of the fringes are recovered, the same algorithm can be used. The only change required is to determine the quadrature phase over the array values instead of over time.
The incoming signal is noisy, a moving average smooths the response. Care must be taken to calibrate the moving average accurately as the ringing occurs in the forces due to the spring nature of the flexure. Over-damping of ringing by the moving average could cause events to be missed and under-damping will cause an erratic reading. In Figure 6.69, the blue dots are the raw sample and the orange dots are the result of a 11 point moving average.

\[ \int_{0}^{2\pi} |\cos(x)| \, dx = \sin(x) \ast \text{sgn}(\cos(x)) + c \]  

(6.42)
\[(\sin(x) * \text{sgn}(\cos(x)) + c)^2 \bigg|_0^{2\pi} = 4.0 \quad (6.43)\]

This value is then scaled to the set points and 1.3N.

\[
\frac{\text{current force} \times 1.3}{(4.0 \times (\text{Top} - \text{Bottom}))} = \text{Force} \quad (6.44)
\]

The force readings are then scaled according to the strain results. This is discussed in the Discussion chapter.

### 6.10 Algorithm Testing

The algorithm was tested using weights. The weights were added in increments of 20 grams until 200g was present on the fixture. This ensures that more than 1.3N of force (1.96N from 200g) was tested, causing the pattern to repeat past \(2\pi\) radians. Figure 6.70 show the results of these tests. The orange line is the resulting force in newtons, the blue is amplified sensor input and yellow shows the state in quadrature.

![Algorithm Output](image)

**Figure 6.70:** Adding and removing 10 sets of 20 gram weights

The spring like action of the fixture can be removed by applying a more gradual increase and decrease in force. Figure 6.71 was created by pressing down on the fixture with a finger and then releasing.
Figure 6.71: Pushing down with finger then gradually releasing
Chapter 7

Discussion

Each of the experiments presented in the Design Verification chapter of this report were conducted. In the following chapter, the team examines the results of each experiment and provides interpretations based on the presented results.

7.1 Fixture Testing Results

This section states the results found in the fixture experiments. The results are compared to the simulated/expected data to see whether or not they satisfy the hypothesis.

7.1.1 Axial Force Testing

The data gathered during the axial force testing portion of the fixture was processed using 6.8 and 6.9. The initial graph shown in figure 7.1 was plotted for the strain recorded for weights 0g to 680g. From this data, it was confirmed that the relationship between force applied and strain observed in the spoke is linear. It was also hypothesized and simulated that the strain in the spokes would be equal and followed the trend seen in 7.3. However, the results acquired from the experiment, graphed in 7.1 show that the strain experienced by the 3 spokes were slightly off from each other. More importantly, the rate at which the strain increases is higher in the experimented data, crossing the threshold of 1000u at a force of closer to 7N in spoke 1 and spoke 2. Interestingly, spoke 3’s behavior is much closer to the simulation than the other two spokes. Using the experiment data, a linear interpolation was done to extend the line of best fit to 13N. This is shown in graph 7.2. Using this data, the team concluded that this initial design of the fixture confirmed the basic working principle that was hypothesized. However, in order to achieve better results closer to the ideal simulation data, improvements can be made.
to the testing method, fixture design and set up to eliminate uncertainties in the data. These improvements are further elaborated in section 7.1.5.

**Figure 7.1**: Data acquired for 0g to 680g weights.
Figure 7.2: Results used to linearly extrapolate.
**Figure 7.3:** Simulated results for the tested fixture.
7.1.2 Non-Axial Force Testing

The data from the non-axial experiment included strain values for 3 individual spokes. The team hypothesized that the strain experienced by each spoke would vary linearly depending on the direction of the force applied. Three directions tested are shown in figure 6.17. The simulation results for each of the configuration is seen in figures 7.4, 7.5 and 7.6. These results did not align with the simulations the team had generated but instead revealed that in every configuration there was at least one spoke that experienced more force than the others as seen in figures 7.7, 7.8 and 7.9.

However, these results were extremely inconsistent even though the 3 trials were done for each. The error observed in the experiment data could be due to inaccurate resistors used in the circuit which have a significant effect on the voltage output. In addition, we used an instrumentation amplifier that was built on a breadboard using 3 operational amplifiers, increasing the possibility for inaccuracies in the circuit. When the weights were hung onto the needle, although the team tried to make sure to eliminate any instabilities caused by the weights swinging, it was impossible to hold it completely still, which could have contributed to the errors in the results. Furthermore, the part of the set up that acted as the contact point to the fixture was not completely stable and caused vibrations, which resulted in the forces not being completely transmitted onto the strain gages. Finally, these results could be improved by using off-the-shelf instrumentation amplifiers and more accurate resistors. In addition, a more stable set up could help eliminate inconsistencies in the data.
7.1.3 Simulation Results of Non-Axial Forces

Figure 7.4: Simulation results - Configuration 1.
Figure 7.5: Simulation results - Configuration 2.
Figure 7.6: Simulation results - Configuration 3.
7.1.4 Experiment Results of Non-Axial Forces

**Figure 7.7:** Configuration 1 experiment results
Figure 7.8: Configuration 2 experiment results
Figure 7.9: Configuration 3 experiment results
7.1.5 Fixture Testing Discussions

Based on these results, the team was able to develop a better understanding of how the forces are translated onto the fixture. The axial forces experienced by the needle can be linearly translated onto the fixture. However, the team realized that modifying the thickness of the fixture and conducting more experiments could lead to the exact force-strain relationship that was seen in the simulation. In addition, the team also showed that it was possible to detect needle deflection with the fixture but was not able to prove it conclusively during the experiment. Nevertheless, improving the experimentation conditions could result in data that can conclusively detect deflections in the needle. Finally, through the strain gauge experiment the team was able to confirm that an FPI sensor placed on the fixture would detect the forces experienced by the spokes of the fixtures. As a result, the data will allow the FPI sensor to be calibrated once it is affixed onto the flexure.

7.2 Algorithm Testing

The response was found to be promising but the issues of fringe loss, ringing and optical noise remain problems that need to be addressed in the future. Laser power fluctuations are a major issue too. Full phantom testing will be needed once these issues are solved. Figure 7.10 shows the results of the strain gauge calibrated to the FPI force reading. The fit used for this graph is in equation (7.1).

\[ \text{strain} = 112.44 \times \text{force}_{\text{applied}} + 6.58 \]  

This equation was used to find a correction factor to map the FPI to strain.

\[ 1000\mu \text{e} = 112.44 \times x + 6.58 \]  
\[ x = 8.835 \text{N} \]  

This changes the max force to 8.835N

\[ \text{Force}_{\text{cycle}} = \frac{8.835 \text{N} - 0 \text{N}}{72.498 \text{rad}} \times 2\pi = 0.766 \text{N/cycle} \]  

This graph shows that the sensor is usable, but there is significant room for improvement in the algorithm and strain characteristics of the fixture, as the top most value should correspond to 1.96N (200g).
7.3 Economics

Introducing this product to a medical environment would have two primary effects on the economics of cancer. First, the device presented in this report would reduce the cost of a biopsy procedure, by reducing the number of attempts needed to set the needle in place during a biopsy procedure. Second, the device may also reduce the number of return visits, as patients would not need to return to have biopsies re-done due to needle deflection.

7.4 Environmental Impact

Although the project does not directly affect the environment, the optical force sensor should reduce the number of biopsy needles used. As a result, less disposable needles will be used, reducing the amount of needles thrown away by the hospital.

7.5 Societal Impact

The largest impact caused by the completion of this project is the societal impact. It will help patients undergoing biopsies by allowing surgeons to remotely conduct a biopsy, through the use of a robot. By conducting biopsies in this manner, we can potentially reduce the invasiveness of the procedure, as well the potential of repeat visits caused by misdiagnosis.
7.6 Manufacturability

The cost to manufacture the fixture portion of this design would be approximately 15USD. The FPI sensor was approximately 265USD, totaling at 795USD for three of them. The price for the PCB, not including components was 430USD, but could be reduced, as the pricier “Super Swift” shipping option was chosen. Finally, the laser used cost approximately 50USD. In total, it would cost approximately 1300USD, not including labor. While somewhat expensive, this is a fairly reasonable price for a fiber optic system. This price could be reduced if the entire system is manufactured on a large scale.
Chapter 8

Recommendations and Conclusion

8.1 Recommendations

Although the team has made major progress toward implementing an MRI compatible force sensor into a surgical robot, there are some improvements that could be made to the overall design. In the following section, the team provides several recommendations to further develop the design presented in this report.

The first recommendation is to replace the laser currently used with a pigtailed laser diode. Currently, a 100mW 650nm red laser pointer serves as the laser within the system. While the laser pointer serves the same purpose, a pigtailed laser could connect directly to the fiber coupler, allowing for more stability within the system. The laser pointer also has an unstable laser diode that causes noise in the signal; this can be prevented by using a high quality pigtailed laser.

Currently, no collimating lens is used to focus the laser onto the linear array. While the design functions without the use of a collimating lens, the addition of a collimating lens would improve the resulting force measurements of the system. A collimating lens would more accurately focus the beam onto the linear array, and, as a result, the output from the system should theoretically become less noisy compared to the uncollimated laser.

Although the team’s fixture design includes three spokes to place FPI sensors onto, only one sensor was used in the final prototype, primarily due to the team’s limited budget. The addition of multiple sensors would allow for future work developing an algorithm to detect needle deflection, using the existing set-up, provided the fixture itself is moved to a position on the needle.

Multiple FPI sensors could also be placed on a single individual spoke. Currently, when using a single FPI sensor per spoke, when the FPI sensor outputs its maximum intensity, it is impossible
to tell whether or not the following sample is increasing or decreasing in force, if the intensity returns to a neither high nor low state. One solution to correct this issue is the use of two FPI sensors. By placing two FPI sensors on a single spoke, the output intensity from each could be compared, allowing for accurate force readings regardless of the current state of the FPI sensor.

The fringes could be recovered using the homodyne method mentioned before. The team is actively investigating these techniques and updates will be posted to the AIM Lab website when a solution is found. A group at Duke University found a way to recover the quadrature phase from fiber based Michelson interferometers using an extra 3x3 coupler [29]. The homodyne method could be achieved in part by using two laser sources. Since intensity is determined by the phase change in light, different wavelengths will change at different rates, thus giving a reference phase between them. Figures 8.1 and 8.2 demonstrate current efforts into this idea.

![Figure 8.1: Red and Green Lasers](image1)

![Figure 8.2: Resulting Yellow FPI](image2)

As mentioned in the Alternative Designs chapter, pressure sensors and pneumatics could be substituted for the FPI. Development of a leak proof thin tube could prove to be accurate and
have significantly simpler processing requirements. The circuitry allows for any analog based
sensor to be used, with only slight changes in the FPGA logic required.

8.2 Conclusion

Throughout the course of this project, the team designed, developed and produced a prototype
for an MRI compatible force sensor. Through the use of a Fabry Perot interferometry strain
sensor, a custom fixture and circuitry, the device has the potential to detect not only the stiffness
force acting upon a needle, but needle deflection, as well.

The continued development of the prototype will allow the device to be implemented into the
Bioengineering Research Partnership Robot. One major benefit to the designed device is the
use of standard biopsy needles. As a result, the device also has potential medical applications,
beyond laboratory testing.
Appendix A

Design Option Analysis Scoring

The following is a breakdown of the scoring system used in the design option analysis conducted by the team. The breakdown will supply a definition for each score relative to the criteria (i.e. What constitutes a 3 in Ease of Use? What constitutes a 2 in “Part Lead Time?”) as well as the weight, or multiplier, that each criteria had on rating.

Cost: Weight-97

- 1 - Parts for this design are beyond the budget of this project
- 2 - Parts for this design are within the budget, but expensive (+400USD)
- 3 - Parts for this design option are within the budget and inexpensive (<400USD)

Invasiveness: Weight - 35

- 1 - The design requires more insertions or causes more tissue damage to the patient when compared a traditional surgeon led biopsy
- 2 - The design would require the same number of needle insertions as a traditional surgeon led biopsy
- 3 - The design would produce results with fewer needle insertions

Ease of Use: weight - 57

- 1 - The device would require additional training and equipment to use
- 2 - The device would require additional training to use
- 3 - The device would not require any additional training to use
Appendix B

Part Lead Time: weight - 71

- 1 - Lead time is greater than five weeks
- 2 - Lead time is between three to five weeks
- 3 - Lead time is less than three weeks

Functionality: weight - 100

- 1 - Design cannot detect needle deflection or determine tissue layer/ other function
- 2 - Design has potential to detect needle deflection OR determine tissue layer
- 3 - Design has potential to detect needle deflection AND determine tissue layer

Required Knowledge: Weight - 64

- 1 - Design is based on purely theoretical knowledge
- 2 - Similar ideas and designs have been built and tested
- 3 - Similar ideas and designs have been built and tested in medical applications

Durability: Weight - 7

- 1 - Device is extremely fragile/requires extra equipment to handle
- 2 - Device is designed for a single use
- 3 - Device has minimal risk of breaking during use
Appendix B

Experiments

B.1 Fixture Design Testing

As the FPI sensors are extremely fragile and expensive, the team first wanted to test our design concept using strain gauges. The experiment serves as a proof of concept, measuring both axial and non-axial forces using strain gauges attached to the fixture, in place of the FPI sensors. This allowed the team to measure the forces at each spoke of the fixture, without potentially damaging the FPI sensor.

B.1.1 Equipment

The following equipment was used throughout this experiment:

- (3) Strain gauges
- (9) LM348N Operational Amplifiers
- (21) 1K ohm resistors
- (3) 120 ohm resistors
- (3) 23 ohm resistors
- (1) Power Supply
- (3) Multimeters
B.1.2 Methodology

B.1.2.1 Axial Forces

To begin, the team first attached three strain gauges to the fixture. The strain gauges are micro resistors, with the resistance value of each gauge changing slightly depending on the strain applied.

While attaching the strain gauges to the fixture, extra care was taken to ensure the placement of the gauges was identical to the placement of the FPI sensors. This would allow us to measure the force on each spoke identically between the FPI sensor and strain gauge.

From here, the team developed three separate Wheatstone bridge circuits, one for each strain gauge on the fixture. This allowed the output voltage of each circuit to be proportional to the strain applied to each spoke. Each bridge consisted of three fixed resistor values, two 1K ohm resistors and a 120 ohm resistor, as well as one strain gauge. The 120 ohm resistor was chosen as the strain gauge had a value of 120 ohms when under no strain.

In addition, each Wheatstone bridge was connected to an instrumentation amplifier circuit, to amplify the output and allow for comparison between each circuit’s output. Each amplifier circuit consisted of six 1K ohm resistors, a 23 ohm resistor and three LM348N operational amplifiers. The 23 ohm resistor is the gain resistor in the circuit, referred to as $R_{gain}$.

One of the circuits, including the Wheatstone bridge as well as instrumentation amplifier, can be seen below, modeled in Multisim.

The following equation was used to relate the strain of a single spoke to the output voltage of the amplifier circuit:

$$\frac{V_{\text{out}}}{(V_2 - V_1)} = (1 + \frac{2R_1}{R_{gain} R_2})^2$$  \hspace{1cm} (B.1)

In this equation, $V_2$ is the voltage between the 1K ohm and the strain gauge on the Wheatstone bridge, while $V_1$ is the voltage between the 120 ohm resistor and 1K ohm resistor on the Wheatstone bridge. The resistors $R_1$, $R_2$, and $R_3$ are all 1K ohm resistors located within the instrumentation amplifier.

Next, we developed a stand to hold both the fixture and the needle. The stand allowed a hook to be attached to the needle to hold the weights, and ensure that all spokes on the fixture would have an identical amount of force placed on them.

Weights were then added to the hook in increments of 20g, up to 600g, with the output voltage of each circuit recorded as additional weight was added. Using the following equation, these
output voltages were then converted into strain measurements placed on each spoke, using the following equation:

Once 600g of weight was placed onto the needle, the data points were then plotted on a graph displaying the weight placed onto the needle Vs. the strain of each spoke of the fixture. The results of this experiment can be found in the Discussion chapter of this paper, in section...

B.1.2.2 Non-Axial Forces

The team also conducted a second experiment to examine the effects of forces placed on the needle at varying angles. To do this, the stand was modified to hold the needle horizontally, with the fixture still attached to the base of the needle. From here, 100g of weight were added to the needle, in increments of 20g. As the needle bent, the force placed on strain gauges varied, allowing the team to observe how needle deflection affected the fixture by observing the output voltage of each circuit.

Once 100g and all data was recorded, the weights were removed and the fixture was rotated 90 degrees. By rotating the fixture, the needle deflection forces were placed on a different portion of the fixture, allowing the team to compare how different angles of deflection affect each spoke of the needle. The experiment was then repeated with the fixture at 90 degrees from the starting point, and again at 180 degrees from the starting point.

The strain of each spoke, calculated using the previous equation, was then graphed compared to the weight applied to the needle. The results of this experiment can be found in the Discussion chapter of this paper.
B.2 Analog Rise Delay Testing

When reading the output from the optical sensor array, there is a small delay between the rising edge of the input clock for the sensor the steady state analog output. This delay is maximized when the output goes from a low value, to a relatively high value (i.e. if there is a drastic change in light intensity from one pixel to the next). This delay becomes important when sampling the analog voltage with the ADC. The desired value is the steady state value that occurs after the analog set up time. Therefore, this delay defines the time after which the ADC should sample. It was also hypothesized that changing the length of the wire through which the analog voltage was transmitted, could affect the time of this delay. This is important due to the design in which the sensor and the ADC are connected to different boards with an unknown length wire between them. To determine whether this hypothesis was correct, the following experiment was done.

B.2.1 Equipment

- Nexys 3 Development Board
- AMS TSL1402R Linear Optical Sensor Array
- Bread Board
- Assorted Connection Wires
- (5) Wire of lengths varying from 5 to 25cm
- Electrical Tape
- Saleae Data Acquisition System

B.2.2 Methodology

First the sensor is connected to the Nexys development board according to the specific outputs defined in the Verilog code. Two of the Saleae channels are required for this experiment, one attached to the analog output of the sensor, the other to the clock signal that drives the sensor. Half of the sensor is then to be covered with black electrical tape to create the most possible contrast between adjacent pixels. The bit file can then be programmed onto the Nexys board. Just before sampling on the Saleae, begin shining a light, such as from a cell phone flash light, on the sensor. With the light shining on the sensor, begin the Saleae and wait for the sampling to complete. Choose ten different full array samples from the acquired data. On each of these
samples, find the maximum voltage change after a clock edge and measure the time between the clock edge and the analog voltage being stable. Record the data.

Once completed, replace the wire from the analog output of the sensor with the next size wire, and repeat the sampling process. Do this with all the sizes of wire.

**B.2.3 Data**

**Table B.1: Delay Testing Data**

<table>
<thead>
<tr>
<th>Wire Length</th>
<th>Delay Time(us)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5cm</td>
<td>0.2 0.21 0.21 0.22 0.22 0.24 0.23 0.23 0.23</td>
</tr>
<tr>
<td>10cm</td>
<td>0.22 0.24 0.23 0.25 0.22 0.24 0.24 0.24 0.25</td>
</tr>
<tr>
<td>15cm</td>
<td>0.21 0.23 0.22 0.21 0.21 0.23 0.22 0.23 0.2</td>
</tr>
<tr>
<td>20cm</td>
<td>0.22 0.24 0.22 0.23 0.22 0.24 0.24 0.23 0.22 0.24</td>
</tr>
<tr>
<td>25cm</td>
<td>0.23 0.24 0.22 0.24 0.24 0.25 0.23 0.25 0.25</td>
</tr>
</tbody>
</table>

**Table B.2: Delay Testing Averages**

<table>
<thead>
<tr>
<th>Wire Length</th>
<th>Average Rise Time(us)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5cm</td>
<td>0.22</td>
</tr>
<tr>
<td>10cm</td>
<td>0.235</td>
</tr>
<tr>
<td>15cm</td>
<td>0.22</td>
</tr>
<tr>
<td>20cm</td>
<td>0.228</td>
</tr>
<tr>
<td>25cm</td>
<td>0.239</td>
</tr>
</tbody>
</table>

Maximum Rise Time: 0.25us

**B.2.4 Conclusions**

The data collected shows that the length of the wire does not have a major effect on the rise time of the analog output voltage. While this disproves the hypothesis, it is a helpful result as it allows for the sensor to be moved off the main circuit board without potential problems. The experiment also gives empirical data for the maximum nominal rise time. The maximum delay must be taken into consideration when designing the timing between the sensor and the ADC sampling. Since the ADC is run on a 4MHz clock, the sample timing for the ADC must be delayed from the sensor clock by at least one clock cycle.
Appendix C

Engineering Drawings

C.1 Design 1

Figure C.1: Engineering Drawing of Design 1
C.2 Design 2

Figure C.2: Engineering Drawing of Design 2
C.3 PCB Schematics

Figure C.3: Engineering Drawing of FPGA
Figure C.4: Engineering Drawing of ADC

Figure C.5: Engineering Drawing of Filter
Figure C.6: Engineering Drawing of Power Connector

Figure C.7: Engineering Drawing of USB
Figure C.8: Engineering Drawing of Programming Circuit. From Paulo Carvalho
Figure C.9: Engineering Drawing of Power Circuit. From Paulo Carvalho
Figure C.10: Engineering Drawing of Output Connector. From Paulo Carvalho
Figure C.11: Engineering Drawing of Logic Connector. From Paulo Carvalho
Figure C.12: PCB Layout - Top Layer
Figure C.13: PCB Layout - Bottom Layer
Figure C.14: PCB Layout - Silkscreen Layer
Figure C.15: PCB Layout - Bottom Solder
Figure C.16: PCB Layout - Top Solder
Figure C.17: PCB Engineering Drawing for holes 1. This is the same file as the laser holder cutout.
Figure C.18: PCB Engineering Drawing for holes 2. This is the same file as the laser holder cutout

Figure C.19: 3d Render of PCB
## Figure C.20: PCB Bill of Materials
Figure C.21: Logo
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