FLOW MEASUREMENT VIA NOVEL FIBER BRAGG GRATING OPTICAL SENSOR

by

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ABSTRACT

A novel fiber optic sensor was designed and developed for the measurement of fluid flow. The sensor was specifically designed to measure flow within small channels such as those found in many biomedical applications without any metal parts (i.e. MRI compatible). At the heart of the device was a Fiber Bragg Grating (FBG). The FBG is an optical device that is very sensitive to axial strain. Specific mounts were designed to fit over the FBG and convert the pressure field, produced by the flow, into an axial strain that is then sensed optically by the FBG. Several different sets of mounts were designed, fabricated via 3D printing and tested experimentally. The mounts were developed to test two types of flow measurement. One set was a flexure mount designed to convert the radial compressive forces of a pressure wave into axial strain. The other set was a flow restriction disc paired with a fixed point on the fiber. The interaction of the flow with the disc would create a tensile force on the fiber and generate a proportional signal. The sensor was tested against theoretical values by measuring reflected wavelength values against known axial strains. The sensor was also tested in a closed loop system that generated forces similar to those found within the human body.

Conclusions based on experimental results were; (1) the FBG with the flexure mount was found to be sensitive and reliable in biological settings, (2) the FBG with the flow restriction disc was not able to generate consistent results to be considered useable.

Key words: Fiber Bragg Grating sensor, flow sensor
Chapter 1

INTRODUCTION

All sensors that are currently available to measure blood flow have vast drawbacks ranging from poor sensitivity to being oversized. As a result, the measurement of blood flow is largely left to blood pressure cuffs and other methods that do not directly measure the velocity of the blood, but in turn, attempt to determine it from other measurements. Most of these sensors are also susceptible to interference from other devices if used in conjunction. The ability to measure blood flow accurately and in vessels of various sizes would lead to better care for patients suffering from poor circulation or other more serious cardiovascular diseases.

Some of the most common methods for measuring flow come from blood pressure cuffs and plethysmographs that focus on the average of the flow instead of the instantaneous and localized flow in individual vessels. If the velocity of blood moving through a vessel could be accurately measured in combination with MRI imaging, a more efficient care model could be created for patients at developing risk levels for strokes and heart attacks, along with preventative care for patients with family history of heart disease.

The aim of this thesis is to design and develop a novel sensor whose primary application is the measurement of flow in small vessels. The sensor is comprised of a Fiber Bragg Grating (FBG), different mounts fabricated via 3D Printing and associated optical instrumentation. Fiber Bragg Gratings are optical devices that are highly sensitive to axial strain, specifically in tension. By using a FBG it is possible to
take advantage of many traits that are inherent to optical fibers. Some of these are; (1) Due to the FBG’s small diameter, it is possible to make a sensor that is minimally invasive and could even be used for small children - a particularly difficult clinical problem, (2) since the fiber and associated mounts can be fabricated from non-metallic materials the device is inherently MRI compatible, (3) the FBG sensors are linear over a large dynamic range of stain levels, (4) the FBG sensor and associated instrumentation has a very small time constant and thus can provide almost instantaneous flow measurements, (5) the optical signals are immune for EMI produced by other electronic or RF devices and (5) optical instrumentation is inexpensively available.

1.1 Scope of this Word

A major portion of this work revolves around the design and development of a thermoplastic flexure mount or flow restrictor that is capable of converting radial forces into axial forces or generating proportional axial forces via flow restriction. Two designs were investigated for their effectiveness in generating useable signals. The two designs were a flexure mount and a flow restricting disc combined with a fixed point on the fiber. Testing was done in a closed loop system of latex tubing and a large reservoir of water.

1.2 Overview

In the remainder of this thesis, background information, surveys of other systems, and results are presented. Chapter 2 provides a look into Fiber Brag Gratings and their different properties. Chapter 3 looks at the core principles that were considered in the process of the work including physiology, properties of fluid
mechanics and other devices that are used for similar measurements. Chapter 4
discusses the design and development of the different designs that were evaluated in
the results presented in Chapter 5. Finally, Chapter 6 summarizes the results and
looks at possibilities for future work.

1.3 Accomplishments

There are several accomplishments of note in this work.

1. Two methods were investigated for producing an all optical flow sensor. A
   third was developed and is planned to be experimentally evaluated in the
   future.

2. A calibration test-bed was developed to directly relate the introduced strain to
   fluid velocity.

3. The sensor was successfully demonstrated to be able to handle pressures and
   velocities that would be in the ranges produced within the human body.
Chapter 2

FIBER BRAGG GRATINGS

This chapter is an overview of Fiber Bragg gratings and their use as sensors.

2.1 Introduction

Fiber Bragg Gratings (FBGs) are well known for their use in dispersion compensators and optical multiplexing. In addition FBG have found use as strain and temperature sensors. A Fiber Bragg Grating is a specially modified optical fiber that has been exposed to intense UV radiation to ‘write’ a periodic pattern of alternative refractive indexes into its core. By exposing the silicon core to intense UV radiation the effective refractive index, $n_e$, is permanently changed at those location of highest UV exposure. This change in refractive index causes an impedance mismatch that results in light being reflected at this boundary. By creating a periodic pattern of these reflections (i.e. optical grating) it is possible to design very narrow band spectral filters (i.e. only select wavelengths of light are reflected or transmitted). This reflected wavelength is known as the Bragg wavelength and is defined by the effective refractive index of the fiber and the spacing of the grating (See Figure 2.1).

By slightly altering the spacing of the grating a different wavelength will be reflected if a continuous broadband light is initially transmitted down the fiber. In many cases, this change in wavelength is measureable to the tens of picometers, which is comparable to the same change in spacing [6]. Any external event that changes the spacing of the grating on this small scale, such as pulling the fiber or heating the fiber,
will produce a change in the reflected wavelength. This means that FBGs are extraordinarily useful for measuring very slight changes in both strain and temperature. In fact, temperature can be measured simultaneously to strain without modifying the fiber since the core and cladding will expand or contract according to an increase or decrease in the local temperature to the grating. [15][23]

2.2 Fiber Bragg Grating Structures

The structure of the grating is as important as the spacing or periodicity. There are six main types of FBGs, each with their own characteristic style of interacting with light. The first FBG created was a uniform, positive-only index change. This style of FBG has a periodic high-low alternation of index of refraction with the grating perpendicular to the longitudinal axis of the fiber. This produces a narrow band of wavelengths reflected with the Bragg wavelength as the peak. The bandwidth is dependent on the variations of the refractive indices in the core. The second type of FBGs has similar properties in being a uniform, positive-only index change, but the grating is at some angle, theta, to the longitudinal axis of the fiber. This is known as a Tilted Fiber Bragg Grating (TFBG) and acts very similarly to a perpendicular grating while also resonantly exciting the cladding modes. [18]

Dispersion can be added to the FBG by making the grating in a chirped style. Chirping is the process of adding a linear variation to the grating spacing. This variation allows more and more wavelengths to be reflected as they travel farther down the fiber. This can be utilized in a variety of areas. A phased array can use signals from a chirped FBG to rapidly transmit data without the need for multiple encoders or transmission lines. It also can decrease the need for controller compensation from polarization mode dispersion and thereby increase the rate at
which data can be transferred. By chirping the fiber, it is possible to reduce the problem of imperfections within the fiber that are inherent in its manufacturing, if it is not a polarization maintaining (PM) fiber. Polarization maintaining fibers have an elliptical core instead of circular core. This elliptical core prevents a second polarization from being excited and is similar to the concept behind a single-polarization (SP) fiber. In an SP fiber, the second polarization state is not maintained or guided and thus escapes the waveguide. Both of these options are more expensive than regular optical fibers and typically have higher losses, especially at long distances. [3][19]

Other types of FBGs include discrete phase shift structures, Gaussian apodized structures, as well as superstructures combining many different aforementioned structures. These will not be discussed as they were not considered for this project.

When looking at a FBG, the linearity is obvious in the governing equation:

\[ \lambda_B = 2n_e \Lambda \]  \hspace{1cm} (2.1)

In this equation, the reflected (Bragg) wavelength is related to the spacing of the grating (\( \Lambda \)) by a factor of two times the effective refractive index. [6] This is a first order equation and can easily be modified to accommodate chirped FBGs, TFBGs, or even superstructure FBGs.
2.3 Fiber Bragg Grating Production

Fiber Bragg Gratings are fabricated by manipulating a photosensitive optical fiber that has been chemically processed to hold any changes permanently when exposed to intense UV light. It is then burned in the correct spacing to generate the alternations or grating. [15]

In order to get the periodic index alternation that defines a Fiber Bragg Grating, the chemically treated fiber needs to be exposed to a pattern of uniformly spaced fringes. These fringes need to alternate between strong intensity and no intensity of UV light. This alternation, or interference pattern, is accomplished by passing a high intensity UV laser through a beam splitter and then is recombined in a lens to focus directly on the fiber. There is often a phase mask to help with the
alternation, allowing for ±1 diffracted orders on the fiber. There can be some adjustments done in the mirrors of the beam splitter to aid in the differences but is primarily done via the binary phase mask. After the exposure is complete, the fiber is released from the tension it was “burned” in, resulting in a slightly smaller grating than was initially applied. It can continue to be exposed in this state for fine adjustments to the reflected wavelength’s base value. [17]

![Diagram of Fiber Bragg Grating Production](image)

Figure 2.2 Fiber Bragg Grating Production [1]

### 2.4 Fiber Bragg Grating Reflectivity

The reflectivity is one of the most important properties of Fiber Bragg Gratings. It is one of the primary determinations of signal strength since it requires that a narrow band of wavelengths be reflected back to the spectrum analyzer from the broadband light source. This signal strength is important because it determines the
allowable losses that are anticipated from couplers, splitters, or any other defects that may be inherent in the development of the system. The center wavelength is ideally determined by the spacing of the alternating indices in the grating, but has a finite Gaussian distribution associated with it. The full width, half maximum of the spectrum is also determined strongly by the reflectivity of the grating. A larger width of the spectrum means that the sensor may not be as sensitive as originally predicted. An ideal FBG would have a reflected signal of a delta function, but this is not possible since all devices have some finite bandwidth. There are three equations that govern reflectivity via coupled mode theory with constant modulation amplitude and periodicity. First, the definition of reflectivity [2]:

\[ R(l, \lambda) = \tanh^2(\Omega l) \]  

(2.2)

In this equation, \( l \) is the grating length, \( \lambda \) is the wavelength interacting with the grating, and \( \Omega \) is the coupling coefficient, which can be defined as:

\[ \Omega = \left[ \frac{\pi \Delta n}{\lambda} \right] M_p \]  

(2.3)

Again, \( \lambda \) is the interacting wavelength, \( \Delta n \) is the change in refractive index within the grating (i.e. the difference between the highs and lows that comprise the alternations of the grating), and \( M_p \) is the portion of the fiber mode power in the core. This can be determined by:

\[ M_p = 1 - \left[ \frac{2\pi}{\lambda} \sqrt{\left( n_{co} - n_{cl} \right)^2} \right] \]  

(2.4)

in which \( a \) is the radius of the core of the fiber, \( n_{co} \) is the refractive index of the core, and \( n_{cl} \) is the refractive index of the fiber cladding. [16] Thus a change in grating length, \( l \), caused by stain or temperature will produce an associated change in reflectivity given by Equation 2.2-2.4.
Chapter 3

CARDIOVASCULAR SYSTEM AND DEVICES FOR MEASURING BIOLOGICAL FLOW

This chapter will look into the physiology of the cardiovascular system from the heart to the vessels that carry blood. It will describe the sizes of the vessels and the flow rates within them. It will also look into devices that are already available for the measurement of flow.

3.1 Cardiovascular Physiology

When looking into measuring different forces in the human body, it is important to look into the physiology involved. The main physiology measured by this sensor is blood flow, with a secondary focus on in-muscle forces, so the most important system to look at is the cardiac cycle.

The cardiac cycle comprises of the heart, arteries, capillaries, and veins. The heart is a four chambered pump that moves about 5.25 liters of blood every minute. This volume is based on a basic governing equation [9]:

\[
\text{Cardiac Output} = \text{Stroke Volume} \times \text{Heart Rate}
\]  \hspace{1cm} (3.2)

According to Edward Laskowski of the Mayo Clinic, the average heart rate is between sixty and one hundred (60-100) beats per minute. [24] Paired with the average stroke volume according to Edwards Lifesciences being fifty to one hundred milliliters (50-100mL), it is simple to calculate this average output. [25]
Figure 3.1 Cardiac Output [8]

The output from the heart goes through thick walled arteries. The blood from the right ventricle goes through the pulmonary artery to the lungs to be oxygenated. This oxygenation happens primarily via diffusion in the capillaries of the lungs. Here, carbon dioxide in the blood moves passively through the very thin walled vessels due to a concentration difference. Oxygen in the air in the lungs does the same into the blood. This blood then returns via the pulmonary vein to the left atrium of the heart. Approximately eighty-five percent (85%) of the blood moves passively to the ventricles from the atria, where it is then pulsed out from the left ventricle to the rest of the body. From the left ventricle, blood passes through the mitral valve, into the aorta, and then to the rest of the body. Typically, blood pressure relates directly to the pressure in the arteries and is also known as arterial blood pressure. The pressure varies between is maximum and minimum during systole and diastole at the average values of 120/80 mmHg. These values are at the very edge of the desired range according to the American Journal of Emergency Medicine. These values do not
change tremendously depending on which side of the body is measured. In fact, if there is a large difference, the patient may want to seek medical advice and be checked for an obstructed artery.

![Artery and Vein gross anatomy](image)

**Figure 3.2 Artery and Vein gross anatomy**

There have been many studies on how fluid flows through the circulatory system and how the blood moves. Many of the veins that bring the blood back to the heart from their respective appendage have one way valves to prevent the blood from flowing in the opposite direction when there is not an incoming pulse. The arteries have special walls that allow them to contract and further move the blood as it exits the heart. They must be thick walled to handle the pressure from the heart, but this contraction is a byproduct of the smooth muscle cells that make up the center layer of the artery. [21]

The sizes of vessels vary greatly from the arteries to the capillaries. This means the flow through each vessel would change to the corresponding size. The overall velocity of the net volume must remain the same as it moves through splitting vessels.
<table>
<thead>
<tr>
<th>Vessel</th>
<th>Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Artery</td>
<td>10mm – 5mm</td>
</tr>
<tr>
<td>Arteriole</td>
<td>5mm – 1mm</td>
</tr>
<tr>
<td>Capillary</td>
<td>8-10μm</td>
</tr>
<tr>
<td>Veinule</td>
<td>0.5mm – 5mm</td>
</tr>
<tr>
<td>Vein</td>
<td>5mm – 15mm</td>
</tr>
</tbody>
</table>

Table 3.1 Average human blood vessel sizes

3.2 Flow

Flow is typically cast into two categories, laminar and turbulent. Laminar flow is known for being a smooth or a parallel layered flow. This means there are no eddy currents that move perpendicular to the rest of the particles. The velocity of the flow does not necessarily force it to be a laminar flow. It is very possible to have a fast moving flow that is also turbulent. When looking into blood flow, it is typically modeled as a laminar flow. This means that each pulse of blood through the circulatory system is an individual instance of laminar flow. At blockages or acute narrowing points in the blood vessels, this laminar flow is forced to become turbulent. This turbulent flow happens because the energy required for passing the same amount of blood through the narrow opening increases. To be more specific, a critical Reynolds number is exceeded by the blood. The Reynolds number is defined by the product of average fluid velocity, vessel diameter, and blood density, divided by the blood viscosity:

$$ Re = \frac{(\bar{v}D\rho)}{\eta} $$ (3.3)
In long vessels, this Reynolds number is considered large but decreases rapidly as vessels begin to branch into smaller networks. [11]

Since each beat of the heart provides a separate pulse of pressure through the blood, it is possible to look at each pulse individually and determine the velocity at which the blood is moving. Based on the velocity of the blood, the pressure it exerts on the walls of its vessel, and the average cross sectional area of different types of vessels, doctors can quickly find blockages, monitor implanted stents, and predict possible flow interruptions.

3.3 Flow Measurement Devices

There are multiple devices that are used to indirectly measure flow in a biological system. These devices can be used to calculate flow from a variety of other measurements that are much easier to get. They range from a net pressure difference to visual approximations of individual particles in the blood.

3.3.1 Sphygmomonometer and plethysmograph

Most commonly, a blood pressure cuff and stethoscope are capable of measuring blood pressure. A blood pressure cuff is known as a sphygmomanometer. It works by literally cutting off the blood flow to the arm and gradually allowing it to begin flowing again. The first number in the measurement relates to the systole portion of the cardiac cycle and marks when blood first starts to move through the brachial artery. The second number in the measure relates to the diastole portion and marks when the artery is no longer blocked from the pressure of the inflated cuff.

Pulse oximeters are capable of measuring the amount of oxygen in the blood, but some can also measure blood flow via plethysmograph. A plethysmograph
measures the change in volume of an organ, body part, or body as a whole. By monitoring the volumetric change in a limb, it is possible to determine flow based on the volume change over time. Most plethysmographs are used to measure the lungs or whole limbs for functional capacity or compliance, but the change in blood in the fingertip can be quickly and easily measured in the same manner. Measurements done in air are less accurate than those done in a chamber but are quicker and less expensive to do. [12]

3.3.2 Ultrasound

It is possible to measure flow using ultrasound and the Transit Time Principle or Doppler sonography. It assesses structures or fluids moving in relation to the probe, either toward or away from. The best measurement is of relative velocity based on a frequency shift of a sample volume. This is an excellent example of noninvasive measures that give maximal information. The flow can be visualized on-screen for the ultrasound and vessel size can be measured simultaneously. To determine pressure, additional devices can be used for maximizing information. Quality of imaging and measurements is highly dependent on the user and their ability to locate and maintain a position on the body for the duration of the procedure. This is highly variable from user to user.

Ultrasound flow monitoring using TTP is governed by:

\[ L = c(dt) + v(dt) \]  

(4.1)

Where L is the distance between the emitter and the sensor, dt is the time period the system is run on, v is the flow velocity, and c is the speed of sound in the fluid being measured. This process for measuring flow with ultrasound requires the emitter-sensor distance to be precisely controlled and the runtime monitored carefully. If
either of these values strays, the calculation becomes inherently flawed. This method is typically used for “clean” liquids, devoid of contaminate or particles, like pure water. It is very accurate, delivering accuracies within 1% of their actual values. The largest problem with this method is the need for a counter that works at very high speeds on the order of nanoseconds.

![Ultrasound TTP diagram](image)

Figure 3.3 Ultrasound TTP diagram

Doppler sonography works in a different way, by measuring fluids that have high particle counts and measuring the frequency shift of a specified particle. This frequency shift is proportional to the velocity of the flow. Doppler sonography is typically the type of ultrasound used to measure flow in the body due to the high amount of red blood cells in the fluid being measured. The average volume of blood is approximately 55% plasma, leaving the reaming 45% to be composed of solid red blood cells and platelets. This large number makes Doppler measurements much easier. [10]
3.3.3 Industrial Flowmeters

There are many ways to measure flow outside of the body, in pipes and tubes. These measures are difficult to translate to effective medical procedures. Flowmeters come in a variety of types but the two most common are turbine flowmeters and vortex flowmeters on the mechanical end, and optical flowmeters on the electronic end.

Turbine flowmeters, or axial turbines, measure flow by translating some mechanical work into a readable measure. These turbines look very similar to a propeller for a boat and function in very similar fashion. The turbine is placed in the center of a flow as it requires flow to completely surround the turbine. The fluid exerts a force on the turbine blades as it passes over them and causes the turbine to rotate. The rotation speed is proportional to the velocity of the fluid at steady state. This model is effective because it only slightly reduces the fluid's ability to flow and
causes less pressure loss than many other flowmeters. The major downside to this type of measure is less accurate than other types.

Vortex flowmeters rely on a vortex trail that was first witnessed by Theodore von Kármán in which a fluid passes over a slightly off centered cylinder in their path. As they pass over this body, vortices spawn on alternating sides in a highly predictable frequency pattern that is based on the fluid’s velocity. These vortices then impact on a piezoelectric sensor that produces a pulse that is sent to some output for use. This is typically used in systems where introducing a turbulence does not affect the outcome as a whole.

Optical flowmeters utilize a particle’s ability to scatter light in a unique way to measure flow. As different particles pass through the first of two lasers, the reflections and scatterings are recorded with many photo detectors. Further downstream, a second laser also illuminates the particle and its properties are recorded as well. The difference in time from the readings along with the known distance between the two lasers is used to calculate the velocity of the fluid. [5]
Chapter 4

DESIGN, DEVELOPMENT, AND METHODOLOGY

The main goals of the project were to design a sensor that was sensitive, small enough for a variety of vessels, and made of nonmetallic materials. The basis for this sensor was a Fiber Brag Grating with an added mount that would be used to translate the forces. Two methods were adopted with an array of possibilities in each; the first was a flexure mount and the second was a restriction disc.

4.1 Flexure Mount

There were two flexure mounts developed for this trial. A third design was considered but never developed. The first design was a cage based design that was a skeletonized in structure, allowing for fluid to flow freely around and into the mount. The second mount was similar in design but completely closed in structure, forming more of an expanded pipe. This design was completely closed to the surrounding fluids. Both of these types of mounts are capable of measuring temperature as well as axial strain. As the ambient temperature around the FBG changes, the grating slightly expands or contracts accordingly. This is compensated by the mount by slightly offsetting the mount from the grating.

4.1.1 Cage Mount

The cage mount was based on work done by Greg Behrmann in measuring tendon forces. This mount, however, was developed using additive manufacturing while his was made bymachining steel components. The cage mount was based on a
pinned arch that translated radial compression into axial strain. A pinned arch uses basic mechanics to convert any force applied perpendicular to the axis to a force directed along the axis. This happens by utilizing the properties of a pin join at the two ends where the arch meets the fiber, or ground in a free body diagram. The pins generate reaction forces that can only work perpendicularly to the initial force applied to the arch because the reaction force that is parallel to the initial force will end up being cancelled out.

![Figure 4.1 Cage mount structure](image)

It is important to understand the mechanics that occur during a force application to the arch to understand how it is used in this project. It is possible to generate a closed form solution for a curved beam assuming the curvature is circular in nature. This means the whole arch (being the circumference of the circle) must vary identically as the radius to some nonphysical, calculated center point. The two ends of the arch
cannot move either horizontally or vertically and therefore have two reaction forces applied to them in the corresponding directions. The horizontal reaction forces relate directly to the tension being applied to the fiber and therefore the Fiber Bragg Grating.

Figure 4.2 Circular arch pinned at both ends, uniform vertical load [1]

For a thin arch, it is possible to determine the horizontal deflection at one end by using the energy principles in structural mechanics. More specifically it follows Castigliano’s first theorem which states “the partial derivative of the strain energy, considered as a function of the applied forces acting on a linearly elastic structure, with respect to one of these forces, is equal to the displacement in the direction of the force of its point of application.” [13] This means the displacement can be calculated from the applied force for a linear elastic system. When measuring a pressure wave with the arch, the applied force is a uniformly distributed vertical load along the length of the arch. By combining all this information, the horizontal deflection can be determined with a single equation [14]:

\[
\delta_{HA} = \frac{R^3}{EI} \left[ (A_{HH}H_A) + A_{HM}(\frac{M_A}{R}) - LP_H \right]
\]  

(5.1)
In this equation, \( R \) is the radius of the circular arch, \( E \) is the modulus of elasticity of the arch material, \( I \) is the moment of inertia of the cross section of the beam, \( A_{HH} \) and \( A_{HM} \) are constants that apply to \( \theta \), the half angle of the arch, which related points A and B to the centerpoint and therefore the radius of the circle. \( H_A \) is the reaction force at A in the horizontal direction, and \( M_A \) is the moment at point A generated by the applied loading force \( L_P \). The loading force can be determined using

\[
L_P = wR \left[ 2\theta^2SC + \left( \frac{3}{2} \right) (2\theta^2 - \theta - SC) + ((\theta - SC) - 2(C - \theta C)) \right] \quad (5.2)
\]

In this equation “\( \sin\theta \)” and “\( \cos\theta \)” have been replaced with \( S \) and \( C \) respectively in the interest of space. Since the arch is pinned, there will be no displacement or moment, causing \( \delta_{HA} \) and \( M_A \) to become zero, simplifying the reaction force greatly. Because of this, the equation for the reaction force becomes:

\[
H_A = \left( \frac{L_P}{A_{HH}} \right) \quad (5.3)
\]

Here \( A_{HH} \) is easily determined using:

\[
A_{HH} = 2\theta \cos2\theta + (\theta - \sin\theta \cos\theta) - 2\sin\theta \cos\theta \quad (5.4)
\]

It is possible to generate a much simpler model by looking at a triangular design instead of a pinned arch. In this model, the arch is replaces with a collection of trusses that are connected with pins. This allows free movement in a single axis at any joint. One end of the structure is considered pinned while the other is considered a roller joint. By defining the second end as a roller joint, it is free to move along a single axis. This allows for the calculation of the reaction force in the horizontal direction to be determined by:

\[
R_H = \frac{lw}{4h} \quad (5.5)
\]

where \( R \) is the reaction force, \( l \) is the original distance between the two ends, \( W \) is the force applied as a single vector to the zenith of the triangle, and \( h \) is the height of the
triangle. This value only corresponds to a single arch, so the value can be multiplied by the number of arches on a given mount. The displacement of the second end comes directly from this reaction force:

\[
\delta_H = \frac{RHL}{AE} = \frac{(L^2W)}{(4hAE)} \tag{5.6}
\]

There were four arches oriented orthogonally to one another to create a symmetrically identical mount. The fluid flowing around the arches would interact similarly on all struts and could be added together due to linear relationships among that. The major benefits to this design are in its simplicity of modeling and minimal material requirements. On the other hand, this design was incredibly fragile and had other interactions with the fluid that were not originally modeled. Since the fluid could flow freely into the mount, but not out of it, there were forces acting to push the arch outward instead of inward. This caused an axial strain to be generated as the fluid pulled the arch in a way more like a flow restrictor than the force converting cage. This interaction also caused the cage to have very high stress concentrations at the point where the arch connected to the collar that connected to the fiber as a pin join. These stress concentrations caused premature failure of the mount, rendering the design unusable.

4.1.2 Pipe Mount

The pipe mount was developed after the failure of the cage mount. It is modeled similarly, but provided a more structurally sound model. The skeletonized design was taken and rotated for all angles around the fiber, instead of four orthogonal components. This provided three major advantages not present in the cage mount: fluid could not get into the mount, fluid interacted with all sides of the mount instead of four separate struts, and the mount was more durable. The largest problem with this
design is that the mount absorbed some of the water it was submerged in for the first thirty minutes, causing a drift in the data over time. This was accounted for by leaving the mount submerges for a calibration period before starting to measure data.

By having the mount as one structure and sealing the collars to the fiber, there was no way for fluid to push the mount upward or axially instead of compressing the mount. The largest problem introduced by this design in the inclusion of static pressure when measuring. Since there is a difference in the internal pressure of the mount and the ambient pressure of the environment surrounding it, the mount is constantly in a state of compression when it is in a fluid environment. This can be accounted for by measuring the static pressure of the system and subtracting it from the total pressure being read with a basic manometer.

The change to the integrated measure of the flow is important because it is impossible to mount the sensor directly in the middle of the vessel. The sensor needs to be in the middle of the vessel to correctly measure the flow because of the boundary conditions of the fluid and the vessel. As the fluid approaches the wall of the vessel, flow must go to zero by definition. In the open cage method, the flow could not be measured by at least one of the struts if they were near the wall of a vessel. This means the measured force would be significantly decreased due to a lack of measurement by at least twenty-five percent of the mount. By switching to the closed pipe mount, the integrated forces are measured, making it less important for the sensor to be in the center of the vessel. The flow all around the sensor will be enough to compensate for the lack of a portion that is near the boundary.
The pipe mount, as well as the cage mount, are less than three millimeters in diameter at their largest point. This means that they are small enough to fit into nearly any of the larger blood vessels like arteries or veins. Capillaries are still too small for this to be a viable option as their average size is approximately four microns [22]. The smaller arteries are on the same scale as the pipe mounting and would be the limit in terms of use without blockage of flow. This is important because it allows for the measurement of very specific blood vessels without influencing how they function, which is a key factor in developing a sensor. This size also means that it can be used in pediatric patients that have much smaller blood vessels than an adult. This alone means that children can gain huge advancements in medical treatment during procedures by allowing for the monitoring of blood flow to specific locations. Because of the ability to monitor so specifically, treatment options may become available that was originally closed off for them.
The design of the mounts both allow for laminar flow to pass over the body with minimal forces being applied since they have a fluid design that works to interact primarily with the pressure waves generated from the pulsatile flow that is generated by the cardiovascular cycle. It also works with the pulses of blood in the legs that are generated by the contractions of the various leg muscles.

4.2 Flow Restriction Disc

The second type of mount that was developed had a simpler design in hope of simplifying the calibration terms. The basis for the design was simple; a flow restrictor would pull on the sensor that was proportional to the flow that interacted with the cross-sectional area that was exposed to the flow.

4.2.1 Small Disc

The first disc that was developed was smaller than the largest diameter of the pipe or cage mounts. This stemmed from the idea that the disc was to be mounted perpendicular to the longitudinal axis of the vessel. The disc had a diameter of 2mm and a small hole in the center that was nearly identical to the size of the fiber. The disc was positioned on the fiber at the end of the sensor region to maximize its potential for the forces to act directly on the sensor. The fiber was fixed via clamps on the tubing that fed the fiber into the closed loop system of the testing apparatus. This method allowed the sensor to be in-line with the flow without occluding the tubing used for the flow. The principle behind this mount was simpler than the pinned arch of the cage or pipe mount. The disc would act as a flow restrictor. There would be no force acting on the sensor when there was no flow since it would sit at its rest position. When there was flow in the tubing, the disc would restrict a portion proportional to the
cross sectional area that was blocked. This was easy to determine the proportional amount of force by determining the area of the disc and subtracting that value from the cross sectional area of the tube. This area of interaction would happen only during a period of flow and gently push the disc downstream, causing the sensor region to stretch. The velocity of the liquid could be determined empirically with the vortex flow meter that was already in-line with the disc sensor.

The problem with this method was the reliance on the sensor to sit directly in the center of the tube in order for there to be any signal from it. Again, due to fluid mechanics, flow must go to zero at a boundary condition. This sensor is highly susceptible to being pushed to the edges of the tube and therefore will not produce a reliable signal. As flow velocity increases, the likelihood of a turbulent flow in this type of system increases. Because of this, the disc sensor is typically pushed to the edges and moves around them instead of staying in the center. The small diameter and therefore small area of the disc prevent it from reliably interacting with the flow. The most common result of this is a bend in the sensor region instead of a stretch. This means there is a very small force applied axially on the fiber itself, but not a measurable signal that would have resulted from the disc. The ideal interaction of the fluid with the disc is shown in Figure 4.5.
A larger disc mount is currently in development with the expectation that the increased area will allow this type of mount to also be used. It is an unlikely way to measure flow \textit{in vivo} as it requires the fiber to be fixed at a single point in order for the stretching to take place when the force is applied to the disc. It will, however, provide a better understanding as to how different mounts and blockages will interact with the flow of blood as it moves through the body.
4.3 Other Mounts

There were a collection of other mounts that were considered and designed but never developed. They attempted to ease some of the problems that the original mounts had like lack of interaction or improper interaction with the flow. These mount were hybrids or variants of the original designs.

4.3.1 Umbrella Mount

The umbrella mount was an attempt to remedy the lack of interaction with the disc mount. By having a shell that was open to the direction of incoming flow, the likelihood of the mount staying in the center of the tubing was greatly increased as the force would be directed back to the center of the mount instead of being pushed around the edges. There were two models being considered: spherical and elliptical. The spherical mount would have caught more of the flow in its shell and gotten its increased force through volume capture. The elliptical mount would have had a higher cross sectional area and would have directed the forces farther from the edge of the mount. The capture of flow would have occurred from a curvature that was
gentler than that created by the spherical mount but a larger area to interact with the flow. This greater area would have allowed it to more easily move from the edges of the tube.

4.3.2 Teardrop Mount

There was also a variant of the pipe mount designed but never developed. The design stemmed from a flaw in the testing apparatus with the flow not being unidirectional as it should have been to mimic flow in the body. The apparatus was fixed instead of going through the process of developing and modeling this mount. The mount would have had two sides that were radially symmetrical but completely different on the upstream and downstream ends. The end that is upstream and first interacts with the incoming flow is exactly the same as the pipe mount. This part of the model goes to the original midway point of the mount. At this point, it changes to an elongated model that follows a truncated cone. This shape allows for the backward flow to smoothly pass over the mount without generating as much of a signal. The reverse flow was significantly lower than the initial pulse because it was a restorative force, trying to refill the pressure generating bulb as quickly as possible and was doing so from both ends of the system. Because of this, it could be mitigated by allowing the flow to move over the gentle slope instead of the bulbous shape of the pipe mount. The modeling became considerable for this type of mount because of the two separate regions and the way it would interact with two different types of flow.

4.4 Testing Apparatus

The idea for developing the testing apparatus was simple; recreate the way fluid moves in the body and be able to freely measure it. This meant similar size
vessels, pressure levels, and a closed loop of fluid. Water was used for sanitary and simplicity reasons. The multiple variants will be discussed in these sections.

4.4.1 Peristaltic Pump System

This first iteration of the system planned to utilize a peristaltic pump to generate a measured amount of flow per unit time. The average human heart pumps about 5qts or 4.7L per minute. This is more easily measured by flowmeters as 75gal/hour. Many peristaltic pumps are capable of generating this level of flow in standard tubing.

The peristaltic pump in this system was a double roller rotating pump and is considered one of the most common styles. The tubing sits in a channel between a hard surface and a point tangential to the circle the heads generate. As the heads spin around they are in contact with the tube for a short period of time. As they continue to move, the point of contact also moves along the path, pushing anything in the tubing forward. This action generates a pulsatile flow by forcing flow through the pinching of the tubing and moving it forward. The pumped being used had two heads and generated pulses that were approximately a quarter of the circle’s radius per rotation.
The pump was controlled within its own system and could be varied based on the type of size of the tubing being used. It was also controllable via RS-232 and LabView. It was set for 75 gallons/hour, which was its highest setting. The fiber was inserted to the tubing in the opposite direction of flow. This proved to be a problem as the flow was very strong and continually pushed the fiber and sensor out into the reservoir the water was being drawn from the loop. It was not possible to put the fiber on the other hand because the flow on the entrance point of the pump was laminar until the roller head passed to move what has filled that area of tubing. The fiber also could not be threaded through the head due to a risk of breaking the fiber. This lead to the need for holding the fiber near the end of the exit tube to prevent ejection.

As this was very early in development, there were no other sensors integrated into the system. These were added into the second iteration for developing an empirical benchmarking of the FBG sensors and mounts.
4.4.2 Bulb Pump System

The second iteration had almost no changes save order adjustments after its original inception. The bulb pump is the pressure driver in the system and requires a constant human interaction to work. The system is simple and easily modifiable to adjustments in the sensor.

The system comprises a reservoir of excess water that feeds into an otherwise closed loop with sensors integrated along it. From the reservoir, a bulb with an approximate volume of 8oz or 235mL is manually compressed to generate a pressure wave, this wave moves through latex tubing past the mount and FBG and through a thin film monometer and a vortex flowmeter, back to the junction and into the reservoir. When the bulb is released, water from the reservoir refills it so another pulse can be generated. The pulses are difficult to control and therefore to measure. This was the reason for introducing the monometer and flowmeter. Since it is difficult to accurately generate measured pulses like in the peristaltic pump, values must be related empirically to reach any level of reliability or certainty.

4.4.2.1 Manometer

The monometer used for calibration in the system was the PX409-050AUSB thin film monometer by Omega. This monometer was chosen for its easy connectivity via USB to the computer and LabView. It is regarded as a high accuracy manometer and is capable of delivering values on average to ±0.08 BSL (Best Straight Line) and a maximum of ±0.14% BSL. The BSL measurement is a combined measure used by companies to describe linearity, hysteresis, and short-term repeatability. It is determined via least squares method for the generated ideal reference line. The 050 model measures an absolute pressure in the range 0 – 50.00 psi. Since the scenario
being measured was at room temperature and atmospheric pressure at base, the lowest value that would typically be seen by the sensor is 14.69psi. The when converted from mmHg to psi, the average blood pressure generated by the heart is only 2.32/1.55psi (from 120/80mmHg) and is well within the range of the sensor.

The downside to this sensor is its sampling rate. It has a normal sampling rate of 6Hz which is very short compared with the rate needed to sample the ECG of the heart. This is remedied within the system by making pulses that are considerably longer than one second for calibration. In testing, the pulses lasted approximately one second. For purposes of this thesis, the whole ECG was not considered, only the largest peak that moves the blood throughout the body.

Figure 4.7 Manometer
As a thin film manometer, it is a diaphragm style manometer and must therefore be based on the concept of an aneroid or mechanical gauge. There is a diaphragm, or thin membrane, that separates regions that have different pressures. The deformation of this membrane is determined by the difference in pressure on either side. By having a known pressure built into the main body of the manometer, any pressure change on the other side can easily be measured via interpolation between values of deflection for known pressures on the other side.

4.4.2.2 Flowmeter

The flowmeter used in the system is the FLP04-L1NA vortex flowmeter by Oval. It is a vortex flowmeter that is capable of measuring flow in the range 0.105 – 1.05 gallons/minute. By controlling the pulses from the bulb, it is very easy to stay within this range for an entire recording period. The output of this particular flowmeter is a pulse output. The output sits on one of two rail values that are set within the calibration software. As flow enters the upstream side, it passes over an offset cylinder causing alternating vortices, or swirling turbulences, as it continues to move downstream. Slightly further downstream a piezoelectric sensor site in-line with the initial cylinder. Each time a cortex collides with the sensor, a pulse is sent as output and causes a swing to the opposite rail value. These pulses all correspond to a volume that passes the sensor equivalent to 0.8133 mL. It is possible to measure the velocity of the flow by taking this volume and dividing by the cross sectional area of the tubing. Since the tubing has a 3/8” diameter it is a 0.9525 cm diameter, which makes a cross sectional area of 0.7121 cm$^2$. This value is constant around the entire loop of the system so the flowmeter does not need to be at any one specific point for accurate measurements.
4.4.2.3 FBG Sensor Position

The FBG sensor and mount enter the loop via a wye connector to allow the fiber to be positioned in-line with the other sensors. The raw end of the fiber points downstream to reduce the chances of breakage from an overly compressive force. There is a length of tubing used to introduce the fiber that is the same as the entire loop and is clamped closed to prevent any backflow that could produce turbulence in the flow before it reaches any of the sensors. The sensor is approximately 2” from the wye connector to reduce any interactions with the hard plastic and flow and maximize the measurement of the flow in the tubing. Because the fiber is clamped at multiple positions in its introduction tubing, there is no chance of slippage due to flow interactions. There are four pinch clamps that hold it in place in the tubing. Because
the tubing is soft, the small fiber is not at risk of breaking as it is embedded in the appressed layers.

4.4.2.4 Other Materials

The rest of the system comprises a large reservoir and latex tubing. The reservoir is capable of holding 5gal and outlets at the bottom to a single piece of tubing that connects to the loop of the measurement system. The reservoir top is open to the rest of the room and any variations in temperature and pressure in the area. The tubing for the entire system is a 3/8” diameter tube with a 1/16” wall. This system works to model the large to medium sized arteries, or the thick-walled arteries where pressure is the greatest, immediately leaving the heart.

Because of the materials that comprise this system are not stiff, there are some variations in the values at different points. Compliant tubing is required because the vessels of the human body are not rigid pipes. This generates a more realistic system but complicates the measurement process.

Figure 4.9 Block diagram of system
Figure 4.10 Full closed loop system (Bulb)
Chapter 5

EXPERIMENTAL EVALUATION

This chapter will discuss the instrumentation and process for gathering and evaluating data from the system. It will cover the entirety of the system and the process used to determine the effectiveness of the sensor. Section 1 will provide information on the transmission of broadband light and the receiving of the Bragg wavelength light. Section 2 shows the controlled strain on a FBG that shows the agreement between the physical sensor and how it should theoretically behave. Finally, Section 3 shows the connection between the sensors and how the wavelength shift was empirically determined.

5.1 Instrumentation

The testing system requires a broadband light source, as well as a spectrum analyzer for monitoring the central wavelength of a reflected band of light. This is accomplished by connecting the two through a coupler and then to the fiber. A variety of fibers were used in the process of testing, ranging from one to three separate reflected wavelengths. The linearity and design tolerances were identical and can be ignored through most of the calibration and data analysis processes and looking only at the change in wavelength instead of the specific wavelength for measurements.

The specific instrumentation models and manufacturers can be found in Table 6.1. There were two computers used for data collection. Since the system was upgraded during data collection. The first was a Dell Precision PWS690 and the
second was a Dell Precision T7500. Both were running a LabView interface that is connected to the spectrum analyzer via a RS-232 connection. The Spectrum analyzer is a compact analyzer designed specifically for Fiber Bragg Gratings, the FB200 by Yokogawa. It is coupled to the Thor Labs broadband light source by a 3dB coupler made by Blue Road. This setup is shown in Figure 5.1.

<table>
<thead>
<tr>
<th>Component</th>
<th>Manufacturer</th>
<th>Model Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Broadband Light Source</td>
<td>ThorLabs</td>
<td>ASE-FL002</td>
</tr>
<tr>
<td>Coupler</td>
<td>Blue Roads Research</td>
<td>BRR-35S</td>
</tr>
<tr>
<td>Spectrum Analyzer</td>
<td>Yokogawa</td>
<td>FB200</td>
</tr>
</tbody>
</table>

Table 5.1 Components, manufacturers, and model numbers

Figure 5.1 Light source, Spectrum Analyzer, 3dB coupler
The FB200 is used to find the center wavelength reflected by the FBG in LabView. The sampling rate is variable with a maximum reporting value of 100Hz. This value was used in the first half of data collection, but later slowed to 50HZ for smaller file sizes. The Labview interface was modified from the original VI supplied by Yokogawa to include both the flowmeter and manometer data. These data values were written to individual spreadsheets while simultaneously being displayed in real time. The spreadsheets were then imported to MATLAB for further data analysis.

5.2 Verification of FBG with Theory

In an effort to verify the Fiber Bragg Grating was working according to its theoretical behavior, as well as to verify that all instrumentation was working properly, an axial strain test was developed. The fiber was mounted on a set of translational stages. The initial spacing of the stages was recorded and considered the base reference value. The stages were then moved apart 10 times in 20µm increments for a total of 200µm of displacement. According to the governing equation of FBGs, this should create a shift in wavelength of approximately 1nm. This occurs because the equation that governs FBGs accounts for the spacing between a single period, not over the length of the sensor. The results from this test can be seen in Figure 5.2. The tests were repeated three times with the same result. They are not visualized in the graph to reduce cluttering.
5.3 In-System Testing

Once the FBG was verified to be working according to theory, the fiber with its corresponding mount was inserted into the testing system. Here, the sensor was subjected to a pattern of pulses from the bulb system. The signal from the FBG and mount, manometer, and flowmeter were simultaneously recorded and displayed in the LabView interface. The same series of six pulses were applied to the system for each mount at least five times in a row, with each set writing a new spreadsheet of data. The pulse series always followed the pattern of a pair of short pulses with no relaxation between them, followed by four full length pulses with uniform separation.

Figure 5.2 Testing of FBG vs Theory
All data from the spreadsheets were then imported to MATLAB for analysis. The flowmeter data was converted from pulses to cumulative volume. The process for this data conversion is as follows: rectification of pulses, multiplication of number of pulses by volume moved per pulse, cumulative addition of volume. The data from the manometer was taken exactly from its original form and then subtracted by the standard atmospheric pressure of 14.64 psi. This allowed for a gauge pressure to be extracted from the absolute measurement. The data from the FBG and mount was taken directly than had the base value subtracted to show only the change in Bragg wavelength. This process allowed for multiple fibers with different base values to be used without different calibrations. The three data were plotted in parallel to show the overlay of data and as a second check for any malfunctions in the system.
It is possible to empirically determine the connection between pressure and velocity with a simple equation:

\[ P_{total} = \frac{\rho u^2}{2} + P_{static} \]  

(6.1)

Since the static pressure was subtracted in the process of analysis, the dynamic pressure of the system is the only nonzero measured value, causing the equation to become:

\[ P_{total} = P_{dynamic} = \frac{\rho u^2}{2} \]  

(6.2)

This is a second check of the velocity of the fluid velocity measure by the flowmeter process of dividing the volume by the cross sectional area of the tubing. This method is complicated since it relies on specific values for the temperature and density of the water to remain the same throughout the measurement. Instead, it is possible to extract velocity from pressure in a method similar to that used in the flowmeter. The manometer generates its measurements in pounds per square inch which when divided by the cross sectional area of the tubing, leaves a value in pounds. The density of water at room temperature is 0.0361 lbs/in\(^3\) and can be used to convert the pound value to a volume. Finally, by dividing by the amount of time that the pressure wave is active for, it is possible to determine a value with units [in\(^3\)/s] which is a flow value. By using these two checks, it is possible to create a set of calibration values for the FBG and mount.

With these values known, it is possible to then correlate the wavelength change from the FBG and mount to a flow value. When the reflected wavelength change is plotted, it is a very sharp initial peak, a stable plateau at the value that corresponds to the specific pressure or flow of the fluid moving over it, and an exponential decay as the mount relaxes back to its initial position. This is exactly what is expected when
comparing to the waveform generated by the manometer. It also has a sharp initial peak and decays exponentially as the bulb refills for the next wave. From here it is possible to empirically determine the wavelength shift from any given pressure or velocity from the manometer and flowmeter values. In the pipe mount case, the wavelength shifts approximately 0.02nm for a pressure wave of approximately 6psi. This pressure wave moves approximately 17mL of fluid per second. Other values and their corresponding wavelength shifts are show in Figure 5.3 and are repeatable over multiple data sets.

<table>
<thead>
<tr>
<th>Pressure (PSI)</th>
<th>Wavelength Shift ((Lambda_0) (nm))</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.001</td>
</tr>
<tr>
<td>2</td>
<td>0.002</td>
</tr>
<tr>
<td>3</td>
<td>0.003</td>
</tr>
<tr>
<td>4</td>
<td>0.005</td>
</tr>
<tr>
<td>5</td>
<td>0.010</td>
</tr>
<tr>
<td>6</td>
<td>0.020</td>
</tr>
</tbody>
</table>

Table 5.2 Pressures and corresponding wavelength shifts

It is easy to interpolate values for any of the other pressures that might arise due to the very clear exponential curve. Any value that lies between known pressures can be determined by following the trend curve seen in Figure 5.4.
Figure 5.4 Best fit of Wavelength vs Pressure

Figure 5.5 Comparison of FBG, manometer, and flowmeter data
The pipe mount generates clear changes in the Bragg wavelength in accordance with the pressure waves measured by the manometer. It is also easy to see the sharp initial peak followed by the exponential decay that is typical of the system as the bulb generates a strong wave and then must relax and refill to send the next pulse. The initial double pulse is used to show the sensitivity of the system. The second pulse occurs before the system has a chance to relax, causing the plateau into the next peak instead of a full relaxation.

Figure 5.6 Small disc data

The flow restrictor disc did not generate any useable data. As seen in Figure 5.6 there was no significant change in the wavelength when the flow restrictor disc
mount was used. This occurs from the disc being pushed to the wall of the tubing where flow goes to zero to meet boundary conditions. Because of this, there is no way to reliably determine a wavelength shift for any pressure or flow values.
Chapter 6

CONCLUSION AND DISCUSSION

6.1 Summary of Results

A fiber optic sensor was developed to measure flow in a human body-like setting. The sensor was developed from a Fiber Bragg Grating and mount created via additive manufacturing, also known as 3D printing. These mounts served one of two purposes; one converted radial forces into axial forces that can then be utilized by the FBG, the other worked as a flow restrictor disc to generate an axial strain by forcing flow to interact with a blockage. By looking into two possibilities for measuring flow, it is possible to gain a better understanding of how different sensors interact with flow and what must be considered when implementing them in an in vitro scenario.

The mounts were created with additive manufacturing and designed in RhinoCAD. With precision additive manufacturing it is possible to create structures on the micro scale that are biocompatible. These two features allow for sensor development that is capable of being inserted in blood vessels.

Models were developed and adapted for determining the axial strain applied to the Fiber Bragg Grating. These models were then tested against theory and in a system that is comparable to some areas of the human body.

6.2 Significance of Results

A novel microsensor was developed from a Fiber Bragg Grating was successfully developed. It was designed to measure flow in the vessels of the human
body but may have other applications that were not yet tested. Compared to other sensors for measuring flow, it is capable of measuring very localized values, instantaneous values, and pressure as well as flow. Temperature sensing can be added by offsetting the mount from the grating region slightly.

The sensor has demonstrated capability in \textit{in vitro}-like scenarios when compared to average values for pressure and vessel size. It is also useful for measurements during an MRI or in pediatric patients.

6.3 \textbf{Future Work}

There are several steps to move the sensor forward and allow for implementation. The overall set up is fragile and needs to be reinforced to prevent breakage in the smaller vessels or implantation. Secondly, the process for developing the sensors is slow and labor intensive. Each mount must be individually printed and applied by hand to a Fiber Bragg Grating. If these were to be made en masse, a better system should be developed. Finally, the sensor should be tested against a variety of autoimmune defenses to ensure biocompatibility. Since the sensor is developed exclusively from glass, silicon, and thermoplastics, it should not have any issue with compatibility, but the testing must be done to allow for clinical use.
REFERENCES


