Fabrication and Characterization of Polymeric Sutures and Brain Stents

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Abstract— **Biodegradable polymers are used in the production of a variety of biomedical devices, including sutures and stents. Sutures are medical devices used to facilitate healing by binding tissue together. Biomedical stents restore various pathways, such as clogged arteries. The sutures were fabricated through the process of extrusion at varying collection speeds. These stents were fabricated with poly-L-lactic acid (PLLA) and acrylonitrile butadiene styrene (ABS) through extrusion, followed by 3D printing. The polymers were then characterized by differential scanning calorimetry (DSC), tensile testing, and X-ray diffraction (XRD). In order to determine what qualities the optimal biodegradable medical device should have, as well as how it would behave once used in the body, the sutures and stent would be tested under a number of conditions, such as pH, temperature, and rate of degradation. Results of suture testing found that the thinner polymers are stronger than thicker polymers. In addition, the PLA brain stent was stronger than the ABS stemt in both axial and transverse loading.**

I. Introduction

The fabrication of polymers could give rise to a variety of powerful and advanced medical products that will help combat today's most prevalent and serious healthcare challenges. These products can then be further developed to construct advanced biomedical devices in the future. One particularly popular material is polylactic acid (PLA). PLA is a biodegradable polymer formed by the processes of polymerization and condensation. It is commonly used to 3D print medical implants.

A stent is a biomedical device constructed out of metal alloys to open up blood pathogens through the arteries in the body. Brain stents clear a path for increased blood flow between certain sectors of the brain. The device is predominantly composed of nitinol, a metal alloy of nickel and titanium. Although nitinol possesses strength and rigidity, patients now desire a biodegradable alternative to nitinol stents. As a result, a sturdy stent with a polylactic acid base is a less invasive and flexible choice.

A. Stroke Prevention

Cerebral atherosclerosis is the buildup of plaque in the blood vessels of the brain. The plaque in the brain leads to ischemic strokes, which account for 87 percent of all cases of strokes. The staggering amount of strokes that occur due to plaque buildup motivates research that strives to find a solution to such a common health defect [1]. An example of a solution to this problem is a brain stent. The stent opens up the artery narrowed by the plaque, thereby restoring blood flow to the brain.

II. Background Section A: Polymer Basics

A. Fabricating Polymers

3D printing is a process in which a filament in a threadlike structure is heated up to a liquid state. The polymer is dispensed through a nozzle onto a print bed, where it creates the 3D model layer by layer, allowing fabrication of highly

sophisticated shapes depending on the thickness of the nozzle being used. After being processed, PLLA can be modified through cutting, filing, sanding, painting, or gluing to customize the shape further. The polymer sometimes jams the nozzle while flowing because of its stickiness, which creates clumps that block the opening. On the other hand, ABS flows easily from the nozzle because it has a higher print temperature and can also be customized further after printing [2].

Since PLLA is biodegradable and relatively soft compared to other materials, the products made of PLLA through 3D printing are not meant for activities that induce wear and tear. Rather, it is suitable for models, containers, prototypes, light-use toys, and medical devices that will biodegrade in the body. Polycaprolactone (PCL) is hydrophobic and has a slower degradation time, so it is often used for drug delivery. ABS is much harder and therefore can be utilized in the production of frequently handled items.

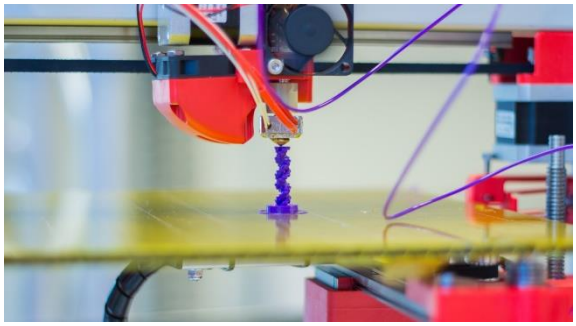

Figure 1 3D Printing [1]

Extrusion is the process of melting a plastic and then shaping it with a nozzle into the desired structure. Polymer pellets are fed and packed into a hopper and then through a feed throat into the main barrel. There is a large amount of friction and thermal energy that results in even heating throughout the barrel. When the plastic is in a liquid state and has been rotated with the screw over the length of the barrel, it is pushed through the die or nozzle. This gives the final shape of the extruded polymer [3].

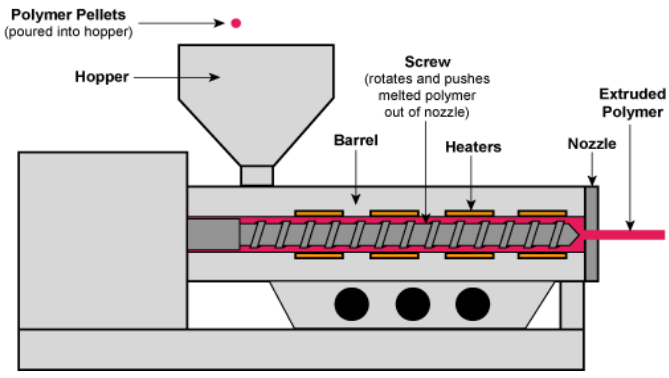

Figure 2 Extrusion Process [2]

B. Characterizing Polymers

X-Ray diffraction is a method that analyzes the percentage and degree of crystallinity in materials. The technology achieves this by generating X-rays in a cathode ray tube. When the X-ray is directed toward the sample, the light is diffracted at multiple angles and distances, which can be related by Bragg's Law. From the areas of diffraction, images can be produced displaying regions of crystallinity. With X-ray diffraction testing, the optimal degree of crystallinity, which controls hardness and flexibility, can be determined for sutures.

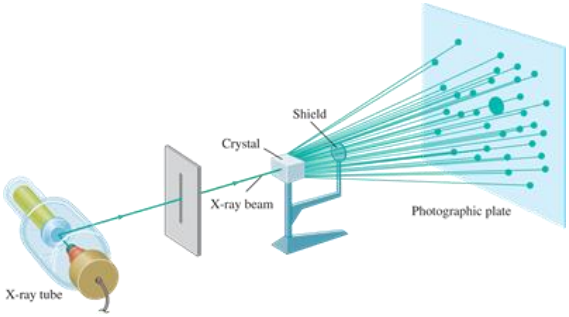

Figure 3 X-Ray Diffraction [3]

Differential scanning calorimetry (DSC) is a technique used to observe polymers. A reference and sample substance are heated and maintained at the same temperature throughout the experiment. The difference in heat required to cause the same change in temperature is measured. The heat flow is dependent upon whether the process is endothermic or exothermic. Through this, several different temperature points of the substance, such as the glass transition temperature, crystallization temperature, melting temperature, and oxidization or decomposition temperature can be determined [4].

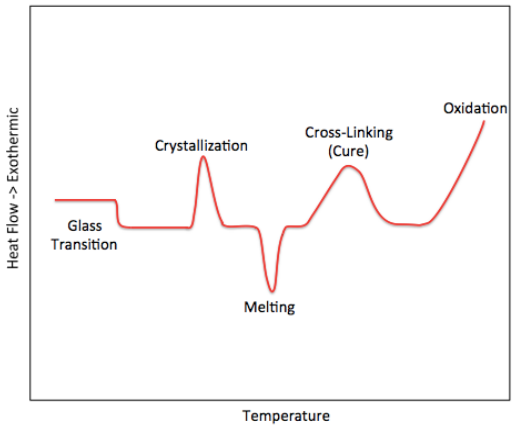

Figure 4 Differential Scanning Calorimeter Data Points [4]

Tensile testing is the procedure in which a material's reaction under tension is measured. It is usually depicted on a Load vs. Extension graph. This type of graph is usually proportional until it reaches the proportional limit and then yields (or, more colloquially "necks"), beginning the process of

plastic deformation. Load can also be referred to as stress, and extension can be referred to as strain. The relationship between stress and strain is described through Young's Modulus [5].

C. Biomaterials

A biomaterial is a non-living material that interacts with biological systems through a medical device [6]. Biomaterials can be useful in drug delivery and organ replacement. They are also used to fabricate devices, such as scaffolds, bone screws, and most topically, brain stents.

There are several classifications of biomaterials: bioinert, bioresorbable, and bioactive materials. Bioinert materials interact minimally with surrounding tissue, while bioactive materials do the opposite, interacting with surrounding bones and even soft tissue. Bioresorbable materials dissolve in the body and are replaced by advancing tissue [7].

D. Poly-L-Lactic Acid (PLLA) Properties

Usually composed of sugarcane and corn stalk, Poly-L-lactic Acid (PLLA) is a stereoisomer of polylactic acid, which means that it is essentially the same compound, but with a differing atomic special arrangement. PLLA is ideal for biomedical use because of its properties. For example, it has an amorphous, semi-crystalline structure and is well known for its biodegradable application. In addition, like other polymers, PLLA has strong covalent bonds between its atoms and weak London dispersion forces. These bonds allow PLLA to be strong, yet processable at the same time [8]. PLLA is currently used in biomedical devices, such as sutures, GTR membranes for dentistry, bone pins, and implantable drug delivery systems.

PLLA, $(C_3H_4O_2)_n$, is formed through the process of polymerization between lactide and metal catalysts in solution. Theoretically, based on a DSC scan, PLLA has a glass transition state of 60-65℃, a melting temperature of 173-178℃, and oxidation and decomposition temperatures greater than 240℃ [9]. Due to its semicrystalline structure, it never exceeds a crystallinity of 40% and has varying physical properties, which is ideal for a biomedical device that needs to be strong, yet flexible.

E. Polycaprolactone (PCL) Properties

Polycaprolactone is another semi-crystalline biodegradable polymer that possesses more flexibility than PLLA. However, it has a slower degradation time. As a result, it is used for a controlled release of drug delivery over a long period of time. It degrades through the hydrolysis of its ester bonds. PCL is hydrophobic, so cells do not adhere to and thrive on it very well. This issue is often solved by mixing PCL with other polymers to create a compatible hybrid [10].

PCL, $(C_6H_{10}O_2)_n$, has a low glass transition and melting temperature of -60°C and 58-60°C, respectively. It has a crystallinity of 56% and its tensile strength ranges from 10.5 to 16.1 MPa [11].

F. Acrylonitrile Butadiene Styrene (ABS) Properties

As an oil-based thermoplastic, Acrylonitrile Butadiene Styrene becomes malleable when heated. Unlike most solids, ABS has no true melting point because it is purely amorphous. As a result, ABS is printed at a high temperature to ensure it is completely liquefied. However, when it cools, it hardens and becomes very tough. This is shown through its strength and high impact resistance. Another property of ABS is high heat resistance due to its high glass transition temperature. An advantage of ABS is that it is flexible, tending to distort and bend before actually breaking [2].

Due to these properties, ABS is suitable for products that will frequently undergo stress. These products range from high-wear toys to automotive trim components. It can also be used for biomedical devices even though it is not biodegradable, as it will be able to withstand high bodily temperatures and pressure from the artery walls. Furthermore, degradability over time is possible with exposure to UV rays and water [2].

II. Background Section B: Polymer Application

A. Brain Stents

Brain stents in modern science have been applied to the maintenance of blood flow. Currently, models of brain stents are used in a procedure called cerebral angioplasty, which treats intracranial stenosis, the narrowing of a brain vessel due to plaque building. An angioplasty catheter is routed to the affected brain artery, where an expandable balloon near its tip inflates, allowing the passage of blood through the affected artery [13].

Figure 5 Stent Function in Artery [5]

Modern medicine also applies brain stents to patients who have suffered from aneurysms. Aneurysms occur when blood abnormally builds up in a weak point of a vessel in the brain. If left untreated, the aneurysm could burst, leaking blood throughout the brain and causing a hemorrhagic stroke. Even after the aneurysm bursts, there is still a possibility of more aneurysms occurring in the same place. Therefore, it is extremely important to treat an aneurysm as soon as possible. Common treatment includes inserting coils into the aneurysm to release the clot in the blood and placing a stent in the same area to keep the coils stationary. The stent is essential to prevent another aneurysm from building up in the same place, as well as preventing the coils from invading the blood vessel [13,14].

Stents can also be used to treat ischemic strokes, during which a blood clot results from an insufficient amount of blood circulating through the brain. Since ischemic strokes account for a majority of stroke cases, finding the best way to treat them is essential. By inserting a stent, the clot will be

pushed into the walls of the vessel. The vessel will be expanded so that blood can flow properly [1].

Issues arise with the presence of a stent located permanently in the cerebral cortex of the brain. A stent constructed of nitinol, causes it to be stationary in the brain, leaving the patient prone to bleeding. The body attempts to reject a stent in the body by increasing its clotting factor even more. Physicians combat this reaction by prescribing blood thinning medication, which prevents platelets in the blood from clotting. This leaves the patient with a high risk for blood loss. However, if the stent is biodegradable, this would no longer be a problem. The stent would degrade after a while, preventing restenosis from occurring, a condition in which the tissue on the vessel walls would attempt to grow around the stent and cause complications within the brain [15].

B. Sutures

Sutures are used to stitch together wounds and surgical incisions. They can be degradable or nondegradable. Degradable sutures are usually made from synthetic polymers, such as PLLA, and are generally used for the closure of soft tissue that require minimal support and swift recovery [16].

Non-biodegradability is an inconvenience because the patient is required to undergo another procedure to remove the suture when the wound has healed. If biodegradable sutures were used, the patient would wait for the sutures to dissolve. The problem with current degradable sutures is that they do not last long enough and may dissolve before the wound is healed [16]. This issue can be solved by experimenting with the different properties of each polymer. One could then select the most appropriate settings for extruding the polymer that will allow it to last for the desired amount of time.

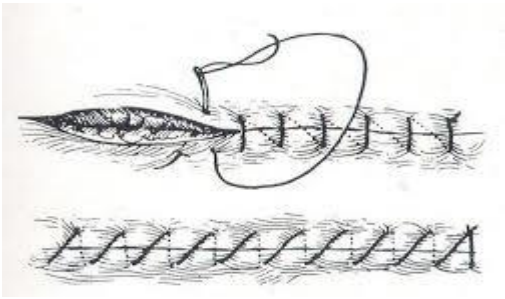

Figure 6 Suture Use [6]

III. Methods/Experimental Design

A. Extrusion of PLLA

The extrusion process involved heating the polymer pellets in the Malvern RH2000 Capillary Rheometer. The polymer was then dispensed through a nozzle in its molten state. As the filament was extruded, the strand was collected on a rotating spool. The rate at which the polymer was collected varied in the experimental design. Multiple variables in the extrusion process were altered in order to extrude polymers with different properties.

The variable of highest importance was the collector, or pickup speed. The pickup speeds were represented by numbers from 1-24 on a dial, which were converted to meters per minute. Overall, the rate at which the spool of polymer was collected directly varied with the diameters of the polymer strand samples. The diameters were measured using a micrometer and recorded.

The speed at which the spool rotated was a simple numeric value and does not include the exact parameters at which the machine was spinning. In order to collect accurate data, a tachometer was held next to the spinning dial to get the speed in meters/min.

Figure 7 Extruder: RH2000 Capillary Rheometer

B. X-Ray Diffraction of Sutures

Sample G was chosen to be observed through X-ray diffraction (XRD), because it had the most consistent and smooth trial during extrusion. A portion of the sample was heated at 120°C for 17 hours and compared with the original extruded sample. It then underwent X-ray diffraction in the lab. This test was done to analyze the difference between the amorphous structure of the non-heated sample and the crystallinity of the heated sample. Other small lengths of sample G (cut to 5-15 mm) were also heated at varying temperatures and amounts of time. Each treated piece of filament was examined using XRD as well. As the pieces were scanned, each diffraction scan was examined for concentric, circular marks, which denoted signs of crystallinity development. These marked sections of the scans were then converted into graphs that compared the intensity of the marks to either the chi values or the 2-theta values, which show the degree of crystallinity either across the radius of or around the circumference of the scan, respectively. Since crystallinity correlates with general strength and hardness of a polymer, the XRD is a suitable test for determining the desired rigidity of sutures.

C. 3D Printing in the Fabrication of Brain Stent Models

The printing of a PLLA brain stent in lab consisted of loading the spool of PLLA into the nozzle of printer. The design of the stent constructed through the TinkerCAD software was uploaded and then constructed. This design and polymer were then tested for their mechanical properties for the fabrication of the ideal brain stent.

Through the fabrication of brain stent models with varying polymers, errors arose with the 3D printer. The original design of the stent was cylindrical in shape with dimensions of 2.5 mm x 10 mm. This design could not be fabricated with the printers provided in lab due to its extremely small size. These repeated errors led to a new path for the fabrication of 3D models that were increased by a scale factor of three.

D. Compression Testing

The 3D printed stents were tested to see how much force they can withstand. This was done because the stent would have to endure pressure on all sides in vessels, so it had to be made sure that the strongest material was chosen. Both samples were cut in two, and each piece was compressed either vertically or horizontally on the mechanical testing machine. The ABS pieces were 4 cm tall and 3 cm wide, and the PLA pieces were 5.5 cm and 6.5 cm tall and 3 cm wide. The pieces were stuck onto the platform with packing tape to ensure that they did not slip off of the machine.

The top press was first positioned so that it just barely touched the top of the stent piece and was then automatically lowered at a speed of 10 mm/min until the plastic broke. The breaking points of the two different materials were compared afterward.

E. Tensile Testing

Several extrusion samples were tested for tensile strength to analyze the resistance of the material to stress. Polymer samples G, I, and L were chosen for tensile testing because they were of various thicknesses and diameters, with I being the thinnest (90 μm), L the thickest (296 μm), and G the medium (258 μm). Each sample was measured and cut into an 8 cm strand and subsequently attached on both ends with small squares of high friction paper, which helped ensure the strand would not slip. The mechanical testing machine was also used for this test, but with clamps instead of flat compressing modules. The samples were stretched at the same speed of 10 mm/min until they snapped, and the breaking points of the samples were compared.

F. Differential Scanning Calorimetry

Approximately 7.00 mg of PLLA was measured and sealed in aluminium crucibles and placed into the differential scanning calorimeter (DSC). The machine was programmed to heat, cool, and reheat the samples. As the temperature changed, the DSC was supposed to measure the flow of heat passing through PLLA and translate the data into a graph. However, the Mettler DSC 823E provided in the lab was under maintenance during the lab period, so all the necessary steps were taken to prepare the lab, but the data was pre-recorded.

G. Relevant Equations

One way to characterize polymers is through the process of X-ray crystallography. X-rays are directed at a sample, and the angle and location of the beams are recorded. Bragg's Law represented as:

$$n\lambda = 2d \sin\theta$$

This law is a calculation that explains the relationships between the wavelength of the X-ray (λ), the angle at which the wave leaves the sample (θ), and the distance it travels (d) [12]. Bragg's law is utilized in the characterization of PLLA, because it allows the molecular structure and properties of a material to be seen.

Stress is a key property in the characterization of polymers. A biomedical polymer must be able to withstand high amounts of force and stress while in the body. Stress, its unit being Pascals, is the force applied over the area of the material. It is calculated through the equation

$$\text{Stress} = N/m^2.$$

IV. Results

A. Crystallinity Post-Heat Treatment

The X-ray diffraction of Data Point G prior to and post heat treatment resulted in a conclusion about the crystallinity of the polymer. Prior to heat treatment, PLLA Data Point G's diffraction scan showed no real structure. This led to the conclusion that PLLA post-extrusion is highly amorphous in its structure. After this same sample was heated at 120 °C for 17 hours and tested again, entirely different results were shown. The appearance of the polymer was opaque and brittle. It had shrunk in size from the original sample. The new X-ray diffraction scan showed rings in the structure. The results all yielded a similar pattern: after being treated, regardless of the intensity of the treatment, the PLLA transformed from being amorphous to having at least a partially crystalline structure. Each diffraction scan possessed two concentric circle marks of varying boldness, close to one another but slightly far away from the center. Those that had been heated for many hours had brighter marks, in contrast to those that were heated for mere minutes, whose marks were fuzzy and at times, nearly undetectable.

In order to determine Bragg's area per sample, graphs were analyzed in terms of the total area. Bragg's area is the area of crystalline structure in the sample. The equation for finding degree of crystallinity, or Bragg's area, is (Bragg's area)/(Total area). The percent Bragg's area for the 5 min at 150°C was 51.56%, 15 min 150°C was 54.74%, 30 min at 150°C was 52.84%, and 17 hours at 120°C was 61.42%.

B. Polymer Extrusion

Through the extrusion process, the results of changing the pickup speed are shown below.

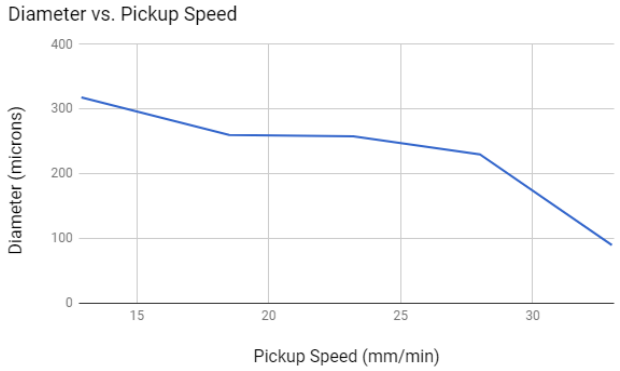

Figure 8 Extrusion Results

Although this graph is supposed to be linear, errors due to collection method caused variation in the diameters of the suture. Errors encountered in lab included setting the extrusion machine at the incorrect temperature. PLLA melts at any temperature between 150-160°C. Therefore, in order to extrude at the proper viscosity, the machine was heated to a temperature between 180-190°C. It was found that at lower temperatures the filament would not flow through the extrusion process smoothly, but at higher temperatures the PLLA would burn or flow at a rate too fast for proper extrusion. Another source of error was improper packing of the polymer pellets. Pockets of air in the barrel caused the extruded polymer to come out opaque and brittle, rather than the desired clear and flexible appearance. A brittle polymer cannot be collected in spools and is not flexible enough to be used as sutures, so the pellets were repacked in order to extrude usable filaments.

Through experimentation, errors due to collection method caused variation in the diameters of the suture. Errors encountered in lab included setting the extrusion machine at the incorrect temperature. PLLA melts at any temperature between 150-160°C. Therefore, in order to extrude at the proper viscosity, the machine was heated to a temperature between 180-190°C. It was found that at lower temperatures the filament would not flow through the extrusion process smoothly, but at higher temperatures the PLLA would burn or flow at a rate too fast for proper extrusion. Another source of error was improper packing of the polymer pellets. Pockets of air in the barrel caused the extruded polymer to come out opaque and brittle, rather than the desired clear and flexible appearance. A brittle polymer cannot be collected in spools and are not flexible enough to be used as sutures.

Although there were some errors, it can still be concluded from this data that as the pickup speed increases, the diameter of the filament decrease. Due to the polymer being extruded in a molten state, the diameter can be shaped to whatever the extruder desires.

C. Compression Testing

The transverse and axial compression testing of both ABS and PLLA led to some unexpected results. PLLA is usually brittle and easy to break, while ABS is generally more flexible and tends to bend before breakage. However, after conducting the compression tests, it was found that the stent made from PLLA withstood more load than the ABS stent did. This is likely because the design of the stents caused the materials to perform differently; the joints in the stents might not have fused as well in ABS as they did in PLLA.

The compression testing results were not all unexpected, however. Both the axial tests of PLLA and ABS withstood more load than the transverse tests before breaking. PLLA aligned vertically withstood about 140 Newtons while when aligned horizontally, only withstood about 30 Newtons. Likewise, ABS aligned vertically, withstood about 65 Newtons, and when aligned horizontally, only withstood about 7 Newtons.

For both PLLA and ABS, the transverse tests did not exhibit much extension. For instance, PLLA, when compressed vertically, only extended about 22 mm and ABS extended 18 mm. On the other hand, when aligned and compressed horizontally, both exhibited more extension. PLLA extended about 30 mm and ABS extended 24 mm.

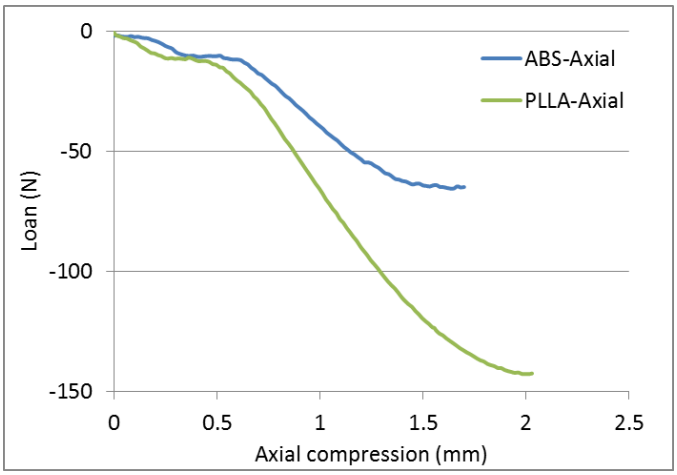

Figure 9 Axial Compression

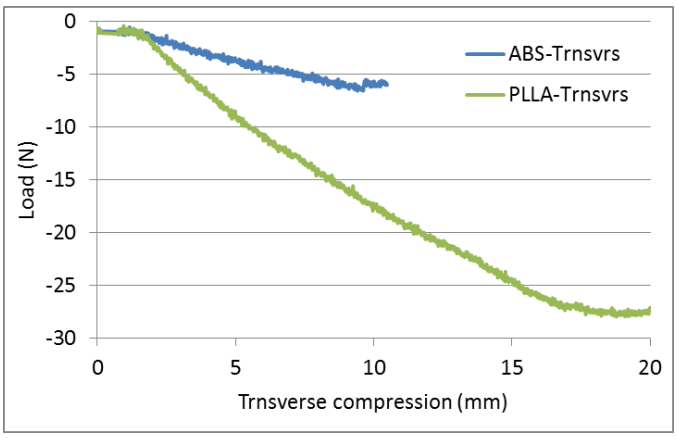

Figure 10 Transverse Compression

D. *Tensile Testing*

After testing polymer samples G, I, and L for tensile strength, data was gathered in order to determine which filament would perform the best as a suture. The force exerted on the filament (in Newtons) was compared to the distance that the filament was stretched by the force (in millimeters).

It was predicted that the thickest filament (sample L) would be the strongest, the thinnest filament (sample I) would be the weakest, and the filament at medium thickness (sample G) would be the medium. However, experimentation and data collection revealed that sample I was able to stretch the most, demonstrating the greatest tensile strength. Sample G had similar results, lasting a few seconds less than sample I before breaking. Sample L was the weakest by far, snapping after about 40 seconds of being stretched. The tensile strength of each sample was unexpected when compared to its thickness, but expected in terms of orientation. The thinner filament is highly orientated, so it was harder to break. The medium filament is slightly orientated, and the thicker filament is not orientated, which is why they broke faster than expected. These results showed that thickness does not necessarily correlate to strength; the orientation of the polymers have to be taken into account as well.

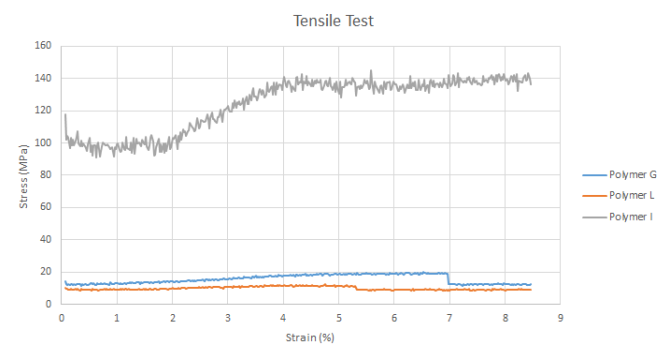

Figure 12 Tensile Data

E. *Differential Scanning Calorimetry*

The expected result of a poly-L-lactic acid differential scanning calorimetry scan is that as the temperature of the sample increases, the PLA will be highly exothermic when it crystallizes and oxidizes, and highly endothermic when it melts. PLLA does not cross link. This means that there is no bond that links PLLA chains together so this section of the graph will not be present. If the sample that is given is already crystalline, then the graph will only show endothermic processes.

Due to the differential scanning calorimeter being under maintenance, the graphs of previous experiments were provided. Other polymers such as nylon 6, polyethylene, and polystyrene were also tested with differential scanning calorimetry to determine their properties. Through the DSC scan for Nylon 6, it can be seen that Nylon 6 has no glass transition temperature because it is highly crystalline and has a high melting point of about 220°C. Polyethylene does not have a glass transition state and melts at approximately 110-120°C. In differential scanning calorimetry, if any discrepancies occur between the heating, cooling, and reheating process, these discrepancies are seen as additives to the real material. Polystyrene had a unique result in that it was discovered that the material did not have a melting point. The material was found to undergo glass transition at approximately 140-150°C. See Appendix for graphs.

The DSC scan for PLLA was different from the rest. The first heating curve seemed apposite. It had a glass transition temperature, no crystallization temperature and had a melting point. However, when it was reheated, the properties were completely different. The material began to exhibit a crystallization temperature and a lower melting point. It was then concluded that the reason for these disparities was that the temperature at which the PLLA samples were heated was so high that it transformed the PLLA into a different polymer with different properties.

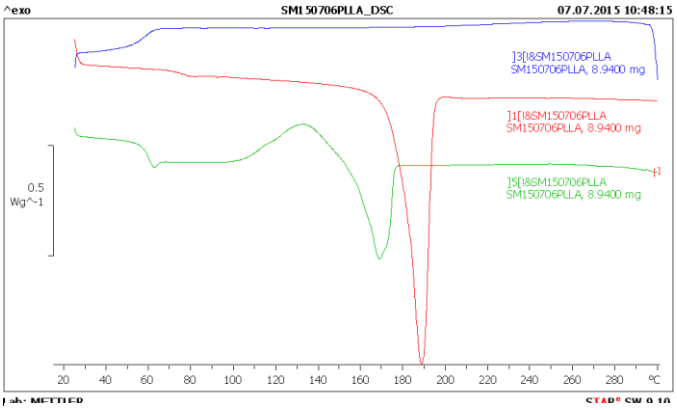

Figure 11 DSC PLLA

V. Conclusion

From the results of the numerous experiments conducted on the polymers as both sutures and stents, several conclusions were reached. If PLLA is heated to a high temperature (below the melting point) and then cooled, its structure changes from amorphous to crystalline. Its crystallinity depends on both the temperature to which the polymer was heated and the amount of time for which the heat treatment lasted. Furthermore, when the PLLA is extruded and wound onto a spool at a specific speed, the thickness of the filament is dependent on the temperature that the extruder heats the polymer to, as well as the rate at which the filament is pulled out of the extruder and collected. The thickest filament produced was Sample L, which had an average cross-section diameter of 0.2958 mm. The thinnest filament was Sample I, whose average cross-section diameter was 0.00353 mm. Also, the differential scanning calorimeter was used to determine how heat flows through PLLA and several other polymers while passing through different states. PLLA itself is exothermic when crystallizing and oxidizing, and it is endothermic when it melts. The presence of a glass transition

depends on if the sample is already crystalline or amorphous before testing.

Testing the biodegradable stents and sutures yielded results that allowed for the determination of which polymer and thickness, respectively, would be most appropriate and practical for the medical devices. Compression testing of the stents showed that the PLLA stent was less pliant in the axial direction than the ABS stent, but more pliant in the transversal direction. The stents' performances varied because of three factors: material, design, and fabrication. It is likely that, because of the design of the stent (or the fashion in which the stent was fabricated), the PLLA stent was able to withstand more stress than the ABS stent, despite ABS being the harder and stronger material. Therefore, PLLA would be a better material for stents, because the stent is more likely to face larger pressure on its walls from the vessel than on its ends. While testing the sutures for tensile strength, it was observed that, while the thickest suture (Sample L) was expected to be the strongest, the thinnest suture (Sample I) performed the best, lasting much longer and therefore enduring the most stress. This means that thinner sutures are better suited for medical use, as they would be able to withstand the natural stretching and twisting of the skin as the patient moves around after being treated.

In spite of these findings, there is still more research to be done. More tests need to be performed on the brain stent to simulate its behavior when it is implanted in the body, such as temperature, pH, and rate of degradation tests. Cold drawn fibers could be observed with x-ray beams to determine whether they are amorphous or crystalline. Heated polymers could also be tested for tensile strength and the results could be used for comparison. Even more innovative research could be done if polymers could be monitored as they are being fabricated. This would allow for easy characterization of these polymers and more accurate results.

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Appendix

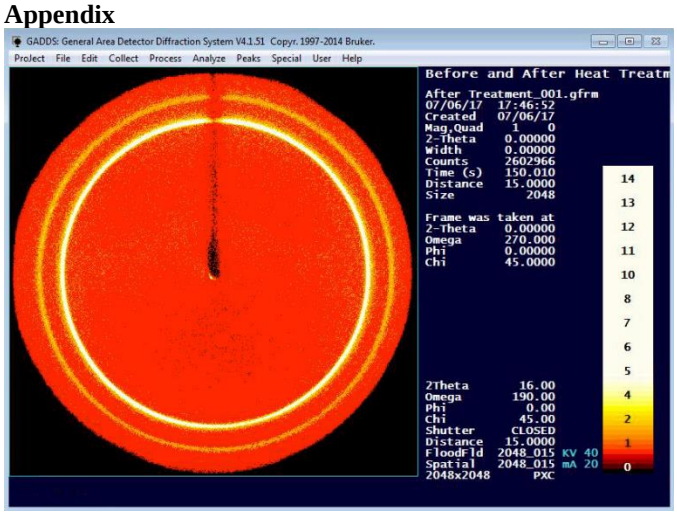

Figure 13 After Treatment 001

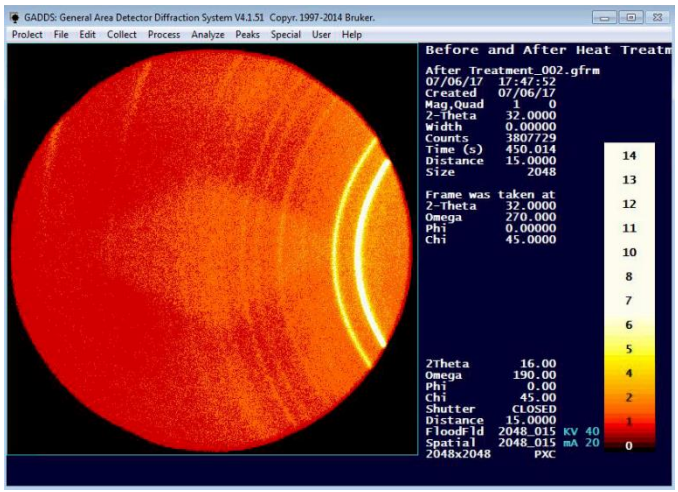

Figure 14 After Treatment 002

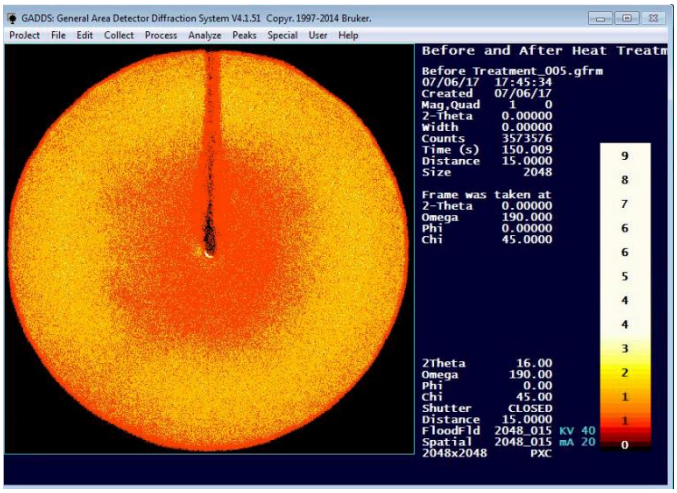

Figure 15 Before Treatment 005

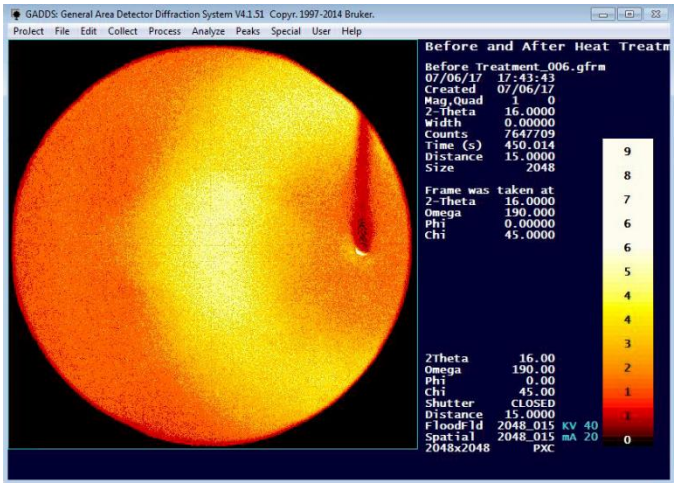

Figure 16 Before Treatment 006

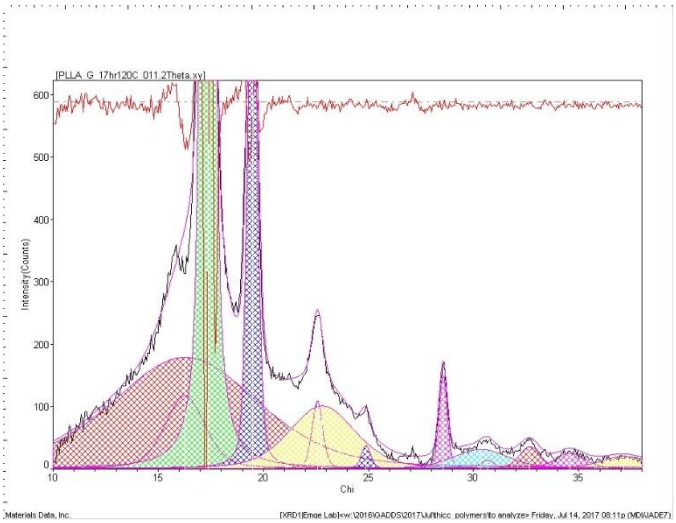

Figure 17 2 Theta Angle Crystallinity