Abstract—Polymers are attractive candidates for synthetic organ replacement because of their biodegradable and biocompatible properties. This study creates and evaluates a polymeric aortic vessel replacement designed to help those with connective tissue disorders, such as Marfan syndrome, by preventing potentially fatal aortic enlargement and rupture. Differential scanning calorimetry, extrusion, tensile strength tests, and X-ray diffraction were used to ascertain the effect of different tests on the molecular and physical characteristics of polymers. Following these experiments, fused deposition 3D printing was used to create an aortic vessel replacement prototype.

I. INTRODUCTION

Polymers are a broad class of materials with a wide range of possible characteristics. They can be easily processed at lower temperatures and pressures in shorter amounts of time relative to other materials, such as metals or ceramics [1]. Consequently, the use of polymers to design effective biomedical devices is rising.

Current biomedical research lends itself towards total tissue replacement as a method of treatment for chronic connective tissue disorders. However, synthetic replacements using polymers have accelerated development and have the potential to be more reliable and customizable. Combined with the potential capabilities of 3D printing technology, many view the future of polymeric biomedical devices with optimism [2].

This project aims to address a specific complication of Marfan syndrome and other connective tissue diseases via the characterization and fabrication of a polymeric blood vessel, specifically the aorta. To improve the prognosis of patients with these disorders, the device needs to be customizable, efficient, and inexpensive. The first step was to experiment with the given polymers to find the optimal parameters to design and manufacture the device. Experimentation in the lab with differential scanning calorimetry (DSC), extrusion, tensile strength testing, and X-ray diffraction were used to evaluate the proper polymer for the success of the device. Two biocompatible polymers which are able to function alongside the body’s tissues were utilized in the prototype of the biomedical device. Thermoplastic polyurethane (TPU), specifically the commercialized NinjaFlex, was used to mimic the elastic inner layer of the aorta, and polylactic acid (PLA) was used to act as the strong, rigid outer layer.

II. BACKGROUND

A. Cardiovascular manifestations of Marfan syndrome

Marfan syndrome is one of the most well-known genetic connective tissue disorders, affecting approximately 1 in every 5,000 people [4]. It is an autosomal dominant disorder that compromises the body’s ability to create structurally sound connective tissue throughout the entire body, including the skin, intestines, and blood vessels [3]. This inability to create effective connective tissue is especially critical in the aorta, the first blood vessel through which oxygenated blood leaves the heart, as it is subject to the highest blood pressure.

All blood vessels, including the aorta, are comprised of three layers attached by connective tissue. When the connective tissue is weak, as in Marfan syndrome, the pressure the heart exerts on blood vessels can cause the vessels to expand; this expansion is known as an aortic aneurysm. This expansion and subsequent thinning of the vessel walls leads to a higher risk of aortic dissection, where a tear forms in a layer of the
aortic wall, allowing blood to flow between the layers [5]. A visualization of aortic aneurysm and dissection is provided in Figure 1. Internal bleeding due to an aortic dissection will quickly result in death if left untreated.

Figure 1. Mechanics of aortic aneurysm and dissection [5]

B. Treatments of aortic aneurysm

Synthetic aortic vessels are used to replace anomalous vessel walls and to mimic the elastic and flexible nature of the aorta. Grafts that utilize portions of the patient’s own artery or vein can be used, but this can disrupt blood flow to the areas from which the vessels were extracted from. Inadequate blood vessel growth to the graft can cause it to die, leading to further complications. Until these issues are resolved, the need for a synthetic facsimile of the aortic vessel is vital [6].

Today’s primary method of aortic aneurysm treatment involves the implantation of a Dacron sleeve. This device is a graft made of a flexible and transparent polyethylene terephthalate (PET) mesh, which is a woven form of polyester. It is able to conform to the patient’s aorta without the formation of any kinks or wrinkles. Attached via sutures, or stitches, the Dacron sleeve decreases the risk of expansion and leakage. If the sleeve is not attached properly during surgery, however, there is a greater risk of aortic enlargement despite the presence of the graft. Another difficulty of the Dacron sleeve is that it must be made by hand and prepared for the patient’s aortic dimensions, creating potential for human error and delaying treatment [7].

With a 3D printable aortic treatment, design files can quickly be changed to fit the patient’s specifications and dimensions, sent over the Internet, and printed in fewer than 12 hours. This technology and design expedites the process and minimizes human error.

C. Polymer basics

A polymer is a long molecular chain made of many repeating units. These are the fundamental building blocks of natural materials such as hair and DNA. Scientists can produce similar polymeric structures in the lab for medical use. The inherently useful properties of a polymer may be enhanced by a variety of additives and treatments during the manufacturing process, broadening its applications. PLA and TPU were the polymers used in this model.

1) Fabrication of polymers: A rheometer is a machine used to extrude polymer in long, thin fibers that can be used for further tensile strength and X-ray diffraction testing. Polymers, in the form of pellets or powder, were inserted into the rheometer and were subjected to heat and pressure to change its physical properties. Melting point and other data about the polymer were gained from a differential scanning calorimetry (DSC) scan. DSC is a technique used to study how a material’s heat capacity is changed by temperature, revealing the transitions such as melting points, glass transitions, phase changes, and curing. A DSC scan of PLA can be seen in Figure 7. This information was used to adjust the settings of the rheometer to melt and extrude a polymer of better shape and consistency.

2) Characterization of polymers: One way of characterizing the polymer is by its tensile strength. An Instron 5869 was used to test the strength of a polymer by elongating the fiber from one end. The machine collected data such as elongation and force applied, which was used to determine Young’s modulus, revealing the stiffness of PLA fibers extruded under certain conditions. Another way to characterize the polymer is via X-ray diffraction, or XRD. This involved the use of an X-ray beam to ascertain aspects of the molecular structure of a substance. The monochromatic X-ray beam, produced by a cathode ray tube, hits a sample at an incident angle and diffracts in several different directions. The resulting diffraction patterns and graphs were used to calculate the percentage crystallinity.

3) Characteristics of PLA: PLA is a polymer produced from natural materials such as corn and beets that is commonly used in 3D printing technology because of its versatility and biodegradability. Its molecular characteristics can be altered to change the rate at which it degrades, and its high strength and thermoplastic properties make it the perfect candidate for industrial packaging and biomedical engineering applications. When heated within its melting point range, 157 – 170°C, PLA can be liquefied, cooled, and reheated without significant degradation [8].

4) Characteristics of TPU: TPU is a biocompatible block copolymer, or a substance that is made up of segments of chemically distinct monomers in a polymer chain [10]. It consists of an alternating sequence of hard and soft monomer segments. The hard segments are made up of isocyanates, while the soft segments are made of reacted polyl. The combined properties of isocyanate and polyl allow TPU to be flexible, elastic, and tough. TPU’s hard and soft characteristics can be used for molding and flexible tubing. It is resistant to impacts, abrasions, tears, and weathering, and has a melting point of 57°C [9].
III. EXPERIMENTAL PROCEDURE

To identify the materials and characteristics best suited for the device, the techniques of extrusion, tensile strength testing, and X-ray diffraction were used. Once those were identified, the prototype was manufactured via 3D printing.

A. Polymer extrusion

An RH 2000 capillary rheometer was used to extrude the polymer into long, thin fibers for testing. Polymer pellets or powder were inserted into the barrel of the rheometer. The sample was then compressed into the heating zone and pushed with a plunger through a cored metal cylinder, a die, with a diameter of one millimeter and pulled through by a spooling machine. The fiber was collected on paper bobbins and stored for later use. The variability of the temperatures set on the rheometer and the speed at which the fibers were collected changed the fibers’ stiffness. Moreover, the fibers were extruded at two different temperatures, first at 160°C and then at 170°C. For each temperature, multiple spools of fiber were collected at different drawing speeds ranging from 5 meters per minute to 65 meters per minute.

B. Tensile strength testing

Tensile strength was measured using an Instron 5869. Both ends of the fiber sample were secured in a grip. Then, a force was exerted on one end of the polymer. The machine pulled the fiber until it broke; if the fiber did not break after reaching double its initial length, the trial was stopped. Inputting the fibers’ diameter and length allowed the Instron to measure load and extension. From these values, Young’s modulus, a measure of stiffness, were calculated for each fiber.

C. X-ray diffraction

The samples were prepared by trimming a segment of a strand of fiber and adhering its ends to a cardboard slide, which had an opening punched into it. For each sample, multiple pieces of the same diameter fiber were bundled together across the opening of the cardboard slide, as seen in Figure 2. The sample was then mounted on the sample holder inside the diffractometer chamber so the X-ray beam would pass through the prepared samples.

Upon mounting the fiber, the X-ray beams were diffracted off of the fiber’s molecular structure and the end location of these beams was registered by the detector. The locations were then interpreted as a collection of points and represented in the form of a detector image of X-ray scattering about $2\theta = 0^\circ$. The scattering images were then integrated to produce an intensity versus $2\theta$ graph and used to calculate percent crystallinity.

The PLA fibers extruded at 170°C and drawn at different speeds (0 m/min, 14.75 m/min, 25.5 m/min, 41.4 m/min, 65 m/min) were tested (1) in their initial states, (2) after they were stretched from the tensile strength test, and (3) after annealing at 100°C for approximately 12 hours. In addition to PLA, a sample of TPU was also analyzed. The data were collected at 20° on a Bruker Vantec-500 area detector using Rigaku Osmic mirrors and a Bruker FR571 rotating anode generator (Cu target, $I = 1.5418\text{A}$). Detector images were collected at 2$q = 0^\circ$ or $30^\circ$ for 3 minutes at a sample distance of 10 centimeters with no sample movement.

D. Fused deposition 3D printing

Fused deposition 3D printing is a method of creating objects through additive manufacturing which utilizes heated plastic, or polymer, dispensed layer by layer through a nozzle.

1) CAD Design Process: A dimensioned model, or 3D blueprint, was created using Autodesk Inventor computer aided design (CAD) software. There were two key aspects to the device’s design: the C-clips and the multilayered aortic vessel.

The C-clips were C-shaped interlocking cuffs intended to protect the suture lines – by which the main vessel would be attached to the natural tissue – and prevent them from bursting due to any weakness post-surgery or due to inherent weakness in the remaining connective tissue. They were also designed to prevent blood leakage through the locking seams, which was important if they failed in their primary purpose and the vessel did rupture. Another benefit to the design was ease of surgical implantation, intended to reduce the contribution of human error to the chances of device failure. The C-clips were designed as interlocking pairs to bypass the need for any suture lines along the device itself.

The multilayered aortic vessel was composed of an inner and an outer layer that mimic the properties of the intima and media and the adventitia, respectively. The intima and the media are similar enough that only one inner layer is necessary for the artificial vessel replacement. The inner layer was composed of TPU because it was the only available polymer with relatively flexible properties. The outer layer was composed of PLA, a harder polymer intended to prevent the inner layer from loosening over time or expanding too much during blood transport. TPU and PLA are water insoluble polymers and therefore would last longer in the body. Polymer longevity reduces the frequency of device replacement and maintenance surgeries.

The device would be assembled by placing the inner layer in the outer layer. Both layers would then be attached to the patient’s tissue by suture lines on either end. Next, the C-clips would be locked around the suture lines.
With CAD, the device’s design is customizable to the specifications of each patient. Additionally, since CAD files are digital, they can be transferred over the Internet to local printing sites within 12 hours. This reduces wait time before surgery, therefore improving the patient’s prognosis. Furthermore, each digitally assembled version of the device can be visualized through a CAD-specific professional tool to produce a 2D blueprint file, otherwise known as a drawing, as seen in Figure 12.

![3D CAD model of aortic vessel replacement](image)

**Figure 3. 3D CAD model of aortic vessel replacement**

**IV. RESULTS**

The polymer’s characteristics are vital in the development of a biomedical device. The analyses used to select polymers for the biomedical device included tensile strength testing and X-ray diffraction to ascertain its physical properties.

**A. Extrusion**

The first three grades of PLA polymer that were fed into the extruder were 6202D, Purasorb, and 18. All three varied slightly in texture and stiffness, but were too brittle to be drawn. Due to improper storage methods and the passage of an extended period of time after the polymers were opened, the molecular weight of the first three grades of PLA decreased enough that they could not be extruded as desired. The molecular weight, or the average weight of each polymer strand, had reached a critical point at which they were too small to entangle with each other and had no strength. The grade of polymer that was successfully extruded was PLA 2003D; at 160°C and 170°C, the resulting fibers were adequately flexible and strong.

For both extrusion temperatures, the speeds at which the fibers were drawn was changed and their diameters were recorded. When the data were graphed in Excel, the expected inverse quadratic relationship was found between diameter and speed. A linear graph resulted from graphing diameter squared versus 1/speed. Faster pulling corresponded to longer and therefore thinner fibers from the polymer.

Since one graph had a less than ideal $R^2$ value, the line of least regression was only approximately accurate. Thus, the relationship suggested was verified by the data and the mathematical model.

**B. Tensile strength**

When producing a synthetic aortic vessel, it is vital to account for how the material will behave under the high-tension conditions of the heart. One such testable indicator is a material’s stiffness. Stress is calculated as

$$\frac{\text{load}}{\pi r^2}$$

where $\pi r^2$ is the cross-sectional area of the sample, and load is the force exerted on the sample (Newtons). Strain is calculated as percentage change in length $L$,

$$\frac{\Delta L}{L_0} \times 100$$

The Instron measures force, the amount of load pulling the fiber, and extension, the fiber’s change in length, in order to calculate stress and strain. Stress is equal to force over the area of the fiber’s cross-section (the area over which the force is being exerted); strain is equal to extension over the fiber’s initial length. Young’s modulus is stress over strain.

In order to perform the tensile tests, three fibers with different drawing speeds were collected from the 160°C and 170°C fiber groups. Three pieces of each sample were collected from each fiber group. Each fiber was then taped at both ends to give the Instron clamps a better grip and to ensure that the effective (initial) length of each fiber from clamp to clamp was a five centimeter control. The fibers were then clamped and pulled by fine adjustment of the Instron until they were taut. After the load and extension values measured by the Instron were tared, each run was performed until a load of approximately two Newtons was reached, until the fiber reached double its initial length, or until the fiber snapped. The collected data was then exported to Excel, where the linear part of each run was selected to produce a linear regression line, the slope being the Young’s modulus. The moduli produced by the trendlines with the best $R^2$ values were recorded, and the moduli of

![Linearized graphs of spooling speed vs. diameter for PLA fibers extruded at 160°C and 170°C](image)

**Figure 4. Linearized graphs of spooling speed vs. diameter for PLA fibers extruded at 160°C and 170°C**
the samples with the same extrusion temperature and drawing speed were averaged out and displayed on a table.

<table>
<thead>
<tr>
<th>Temperature (°C)</th>
<th>Speed (m/min)</th>
<th>Young’s Modulus (Average MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>160</td>
<td>7.8</td>
<td>1.13143</td>
</tr>
<tr>
<td>160</td>
<td>20.2</td>
<td>2.68295</td>
</tr>
<tr>
<td>160</td>
<td>65</td>
<td>0.3898*</td>
</tr>
<tr>
<td>170</td>
<td>7.8</td>
<td>0.79787</td>
</tr>
<tr>
<td>170</td>
<td>20.2</td>
<td>2.96407</td>
</tr>
</tbody>
</table>

*annealed; only one sample

Figure 5. Average Young’s modulus of PLA fibers after different extrusion conditions

The similarity of the slopes indicated consistency in the controls, giving further validity to conclusions drawn from the data. On average, Young’s modulus increased by a factor of 2.37 when drawing speed increased from 7.8 m/min to 20.2 m/min for a polymer extruded at 160°C. For a polymer extruded at 170°C that factor was 3.71. While there was not sufficient data to draw a true mathematical correlation between drawing speed and Young’s modulus, there was enough data to prove that a relationship exists; faster drawing speed produced a much stiffer fiber, and slower drawing speed produced a much softer fiber. Note that a fiber with a modulus of around 1 has almost no practical building application, while a modulus of around 2.5 is just about commercial grade. Simply by changing the drawing speed, the stiffness of the polymer can be manipulated to suit different purposes. In this case, the stiffness of the polymer must be designed to stretch to accommodate blood transport but retain aortic shape upon release of blood pressure.

The change in stiffness occurred due to a change in preferred orientation. This is where the interchain structure between the polymer units are changed and aligned as such that it is not as easy to pull as low orientation. Stretching occurs when the polymeric chains have to be reoriented in order to follow the load vector as it pulls the polymer towards uniform chain motion. When there is no reorientation to be done, there is no elongation of the fiber to be observed.

Additionally, the annotated section of the table shows the one annealed sample’s Young’s modulus at 65 m/min. Based on the previous conclusion, one might expect the fiber with such a high drawing speed to have had a high Young’s modulus. However, the important difference is that the fiber had been annealed, meaning that it had been heated for an hour to make it more crystalline. The higher crystallinity of the annealed fiber overrode the high drawing speed’s effect on the polymers stiffness, and the fiber broke almost immediately upon exertion of an initial load. In light of this, long-term heat treatment should be avoided when working with biomedical devices that rely upon their flexibility to resist breakage in highly vulnerable areas where implementation is necessary.

C. X-ray diffraction

The X-ray diffraction trials revealed information about the molecular structure of PLA and TPU, most notably the degree of crystallinity. Untreated, elongated, and annealed PLA fibers in addition to untreated TPU were analyzed. The scattering images produced by untreated PLA drawn at low speeds showed relatively little intensity, which indicated that the samples had a largely amorphous structure. In comparison, the scattering image of annealed PLA had well-defined rings of higher intensity, signifying greater organization or crystallinity. This sample also exhibited some degree of preferred orientation, as shown in the scattering pattern seen in Figure 16. The pattern was a cone about 0° 2θ with the intensity maximum located around 20° and with non-uniform intensity located around the χ torsion angle; this orientation occurs when the majority of the polymer chains are oriented in the same direction. The sample that was placed in the Instron and drawn at a speed of 7.8 m/min exhibited a similar scattering pattern to the annealed PLA sample, which showed that heat treatment and extreme elongation resulted in preferred orientation.

Integration of the scattering images produced graphs of Intensity versus 2θ and, from these, the percent crystallinity of each sample was determined. Percent crystallinity was calculated with the following formula:

$$\text{Percent crystallinity} = \left( \frac{\text{crystalline peak area}}{\text{total area}} \right) \times 100$$

The annealed fiber had sharp and narrow peaks and a percent crystallinity of 27%, as seen in Figure 17. The percent crystallinity of the elongated PLA and TPU samples could not be calculated, as they were almost all completely amorphous and exhibited no crystalline peaks as seen in Figure 19. Thus, these data suggested that annealing caused the amorphous polymers to exhibit some crystalline domains, and stretching the polymer did not produce any appreciable amount of crystallinity but did produce some preferred orientation in the amorphous polymer, a first step towards creating crystals.

More crystalline polymers, such as the annealed and elongated samples, tended to be more brittle and dissolve less readily. Conversely, more amorphous materials like the untreated PLA and untreated TPU samples were more flexible and dissolved more rapidly. The brittle properties of the more crystalline samples of PLA would have allowed the outer layer of the device to mimic the rigidity of the aorta’s outer layer. However, if the PLA fibers used to print the device were highly crystalline as a result of annealing or elongation and thus brittle, the outer layer would have been more prone to rupture in the high-pressure environment of the aorta. Knowing the influence of treatments such as elongation and annealing on the physical properties of the fiber helped determine which treatments should be applied to the fiber before use in 3D printing of the device.
D. 3D printing

The dimensions of the sample device that was successfully printed were 21.4 mm for the diameter of the inner layer; 23.4 mm for the diameter of the outer layer; 50 mm for the length of the main vessel, which is composed of both the inner and outer layers; 2 mm for the combined thickness of the inner and outer layers, each layer being 1 mm thick; 12.7 and 14.7 mm for the radii of the inner and outer C-clips, respectively, both of which are 20.3 mm thick and 10 mm long; 1.8 mm for the inner C-clips locking piece, and 2.5 mm for the outer C-clips locking piece.

A Lulzbot Taz 6 printer was used for printing the TPU elements, and an Ultimaker 3 printer was used for printing the PLA elements. Two pairs of C-clips, three versions of the TPU inner layer, and one PLA outer body were printed. Three versions of the inner layer were printed to test the different tolerances between the inner and outer layers, as the inner should have snugly fit within the outer. One print had the same diameter as the original CAD specified, one was scaled up by five percent, and one was scaled down by five percent. Ideally, the differences in diameter of the inner layer would not affect the inner layer’s ability to conform to the size of the outer layer when inserted. However, the smallest version did not touch the walls of the outer layer, although it might later have been shrink-fit to do so. The original version of the inner layer fit perfectly within the outer layer, with ideal tolerance that prevented it from sliding out, even when pulled. The largest version had the same diameter as the outer layer and would not fit in the outer layer. Thus, if the original design was followed, the original version would have been the best choice.

However, it was soon realized that the outer layer was too rigid to properly accommodate blood flow. When the largest inner layer version had the same diameter as the outer layer, having only one layer composed of TPU began to seem like a superior alternative.

Another issue arose when it was observed that a support lattice was printed on the inside of each vessel layer. The lattice could not have been removed, and appeared as if it would have promoted clotting and interfered with blood flow if left inside. This seemed to be an issue with the raft algorithm of the 3D printer software attempting to support a print that should not have needed it. Future print setting optimization would resolve this issue.

In addition, every round surface was rendered dodecagonally, as shown in Figure 6. An unnatural geometric structure would have prevented the device from fulfilling its purpose. For example, implantation may result in unstable suture lines. This change was likely due to faulty print file transfer to the printer software, because the default import settings for a given printer often need to be adjusted for the type of file it is integrating. This necessary adjustment likely failed to happen when the file was passed off to a third party for printing.

These issues occurred with the first iteration of the design in the prototyping process and can be remedied with relative ease. Most importantly, these first attempts demonstrated that the design was printable, and that, when assembled, performed as intended.

V. CONCLUSIONS

Characterizing polymers extruded at different temperature and drawing speed combinations allowed for a greater understanding of the ideal material for the vessel replacement. Evidently, a polymer’s diameter, whether or not it was annealed, and whether or not it was stretched all determined its suitability to replace the aorta.

A. Significance

The ultimate aim of this device was to improve the quality of life for those suffering from connective tissue disorders and their subsequent aortic complications by producing a customizable and more effective alternative to the current intensive aortic valve replacement surgery. Because the heart is one of the most sensitive vital organs in the body, connective tissue tears near or at the heart can place the patient in a life-threatening situation. A complete replacement of the aorta ensures peace of mind for the patient, because they avoid chronic issues such as suture degradation on an already weakened area or the possibility of multiple invasive surgeries.

More broadly, the application of biocompatible polymer devices to intractable medical problems poses an opportunity to supplement and enhance the fields of regenerative and reparative biology. The aortic vessel replacement provides a case study in addressing a historically critical need with modern technological solutions.
B. Future work

Given the time frame and restricted resources provided for this project, material testing was limited. Only data from two distinct samples of PLA proved conclusive. These data indicated that a relationship exists between the drawing speed of a fiber and its Young’s modulus. However, there were not enough data to form a mathematical correlation between the two. In the future, more time would be devoted to tensile strength and X-ray diffraction testing on a wider range of PLA samples.

Another step would be to perform axial and transverse compression testing on the 3D printed prototype using an Instron 5869. Compression testing would show the crush resistance of the device, indicating whether or not the device is structurally sound enough to act as an aortic vessel replacement. From there, multiple iterations of the model would be produced to develop a device with a structure that may withstand the desired load, undetermined as of now.

Further design alterations to the device could also be examined. One option for improving the device is to implement a layer of scaffolding that would allow living tissue to regrow and repair over time. This would avoid multiple dangerous surgeries performed on a patient to replace the artificial aorta as the patient grows.
APPENDIX

Figure 7. DSC diagram of PLA, showing the melting temperature and the other behavior of the polymer at different temperatures

Figure 8. Graph of elastic region of PLA extruded at 170°C and drawn at 7.8mm/min, obtained from Instron testing

Figure 9. Graph of elastic region of PLA extruded at 170°C and drawn at 7.8mm/min, obtained from Instron testing

Figure 10. Young’s modulus on polymer extruded at 170°C and drawn at 20.2mm/min

Figure 11. Young’s modulus on polymer extruded at 170°C and drawn at 20.2mm/min
Figure 12. CAD drawing of aortic vessel replacement

Figure 13. XRD detector image of Untreated PLA, exhibiting a diffuse scattering pattern

Figure 14. XRD detector image of PLA with draw speed of 50 m/min, exhibiting preferred orientation

Figure 15. XRD detector image of PLA with draw speed of 65 m/min, exhibiting a more defined preferred orientation

Figure 16. XRD detector image of annealed PLA, exhibiting crystalline structure

Figure 17. XRD peak profile fitting of annealed PLA, exhibiting its amorphous and diffracted regions
Figure 18. XRD peak profile fitting of typical intensity versus 2 theta of untreated PLA

Figure 19. XRD peak profile fitting of Polyurethane(Ninjaflex) and PLA(Annealed, Untreated)
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