GAIT CYCLE ANALYSIS AND MECHANICAL DESIGN OF ANKLE FOOT
ORTHOSIS TESTBED

A Thesis Presented

By

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Ankle Foot Orthosis (AFO) is a type of ankle brace for providing assistant force support on the patient's ankle. The research was focused on the structure redesigning, gait cycle analysis and Rapid Prototyping-Ankle Foot Orthosis testbed design.

It is tried to provide a new assembling AFO design, so that lower cost along with a standard process are achieved in order to build a customized AFO. The formal AFO structure was an integrated design. The integrated design was separated into three parts, a. customized base, a. function bar and a standard cuff. The customized base was built by rapid-prototyping method. The surface of the patient's ankle was modeled by the 3D scanner and then built by the 3D printer. The function bar worked as a spring. When the patient is walking, the bar could store the energy and release it. In this process, the weight of the patient is transferred to the spring's potential energy and then the energy is transferred as an assistive force. During designing the structure, it has been tried to provide a range of materials and sizes to meet the different needs of patients who need different assistant levels. The Finite Element Analysis (FEA) was completed on several regular materials. There was no way to formulate or calculate the shape of the AFO, since a customized AFO should be based on the certain patient. As a result, the RP-method was only for assistance, and hence, this work focused on providing the guidance for designing an eligible AFO for a certain patient at a certain status. The standard cuff has a range of sizes, and it could be adjusted by the patients at any time.
The testbed for the AFO was based on the gait cycle analysis. For the testbed, an ankle foot simulation system was built. Using this testbed, the AFO can continue working for several hours, so that fatigue results for the proposed AFO can be obtained. The underlying principle here is to test the AFO design rather than simulating the human ankle foot. Hence, some necessary simplification was applied. For this, some necessary degrees of freedom were retained while the rest were eliminated. Three concept designs were compared and then one final design was built. A simple PID controller was designed and implemented as the control system for the testbed. The test result demonstrated the feasibility of the design and applicability of the testbed.
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1 INTRODUCTION
1.1 Introduction and Problem statement

Paralysis is one of the most common disabilities resulting from stroke; paralysis patients lose control of their limbs and body. To possibly achieve the best quality of life, patients need rehabilitation and medical devices that could help them walk normally. [1] Ankle Foot Orthoses (AFOs) are a kind of rehabilitation device that can help patients who have injured ankles. [2] AFOs support and control patient's ankle movements. Since the process of the rehabilitation takes a long time, patients need to wear the orthoses for several months or even years. Therefore, the patient's comfort should be a first consideration of AFOs. Custom-fit AFOs can satisfy this need. Building special AFOs according to an individual patient’s foot anatomy can make the patient more comfortable when they wear the orthoses. [3] However, high cost and fabricating of customized AFOs is a stressful process to patients. Fabrication of one AFOs unit takes approximately 4 hours by experienced orthotist and it leads to a high cost. The casting form of the patient’s leg cannot be stored for more than two months due to the warehouse limitation, so the entire fabrication process starts over when patients need new AFOs. Thus, the combination of two technologies, i.e. 3D laser scanning and rapid prototyping (RP), is proposed as an alternative method to create custom-fit AFOs with low cost, less fabrication time, and lasting digital data for a patient's foot anatomy. Current RP AFOs', which are designed by Biomedical Mechatronics Laboratory (BML) based on a posterior leaf spring AFOs, satisfy user's comfort and AFOs' performance in terms of flexibility.
and support. [3] However, current AFOs lead to several problems, i.e. inefficient fabrication, limited use depending on type of injury, and poor durability.

Figure 1-1 Current Design of Custom-fit AFOs [4]

As seen in Figure 1-1, current AFOs have a large cuff and it increases fabrication time (16.7 hours) and cost (approximately $2,500) due to the inefficient fabrication. Considering the alignment of the AFOs in the volume of build platform, a 3D printer can only fabricate one unit of AFOs at once. Thus, the volume of waste is much greater than the product volume, which is lower manufacturing efficiency. The posterior leaf spring AFOs perform well for a drop foot patient, but certain patients who have unstable ankle or ankle deformities cannot use it. In addition, since current AFOs are made of nylon 101, Accura SI 40, or Somos 9120 UV, its durability is questionable and it turns out a weak structure to support patient's body weight. Therefore, current Rapid Prototyping AFOs must be redesigned in terms of less cost and manufacturing time, general patients' usage, and increasing durability and support of weight.
In addition, embedded sensors are required to monitor patient's rehabilitation status and notice the orthoses condition such as material fatigue. Doctors can diagnose a patient's condition with a numerical value. Patients and orthotists can know when they should replace the old device with a new one, so that it helps patients to avoid a second injury when the old device breaks down. In order to embed sensors, proper sensor type, sensor-mounting location, and data transmission are carefully considered.

2 Significance/Motivation

The current version of custom-fit AFOs provide high rehabilitation performance and comfort. However, the fabrication of this type of AFOs is inefficient because of a large barrier covering user's calf. It increases material usage and leads to high cost. In order to reduce the cost, the current AFOs version must be redesigned. Redesigned AFOs are consists of three parts, bottom (blue), standard cuff (green), and modular interchangeable load bearing rod (red), as seen in figure 2-1. The bottom of AFOs is customized using 3D scanner and rapid prototyping considering the patients comfort. The standard cuff is designed for attaching the AFOs to the patient’s leg and foot. The rod provides assistant torque to help patient's ankle move during gait. The cuff and rod should be adjustable depending on different size, ankle torque, and rotational range of patients.

![Figure 2-1 New Design Concept of Custom-fit AFOs [4]]
The geometry of rod and cuff should be general shape instead of customized. This general geometry leads to low cost because it allows mass production. Also, general geometry of rod and cuff should be adjustable depend in different size and weight of patients. In addition, new design of AFOs should be able to provide different range of force or torque based on the patient's condition. For example, if the rod is connected to cuff along with replaceable spring, different range of torque can be provided depends on patient's need by diverse spring constant. Redesigned AFOs are created so that patients would not feel painful, tired or uncomfortable during long rehabilitation time.

In this project, the new design concept of RP AFOs will lead to several significant innovations for custom-fit AFOs.

The proposed AFOs design is manufactured in three parts. Only the base part is customized; the last parts and assembly part are mass-produced. It leads to low cost and high manufacturing efficiency (high productivity within short manufacturing time and decreasing ratio of waste to product).

The adjustable connection is manufactured with simple geometry of rod. The joint connection and the flexible rod with rigid mounting are optional depending on patient conditions. The movable assembly or the flexibility of the rod can define the ankle rotational range and dynamic stability. The simple geometry of rod leads to mass production and increase durability of the rod.

Since the new concept of AFOs is manufactured in three parts, each part can be selected with different material. The vulnerable part can be replaced with certain material along with high Young's Modulus. In terms of support and flexibility, proper material of each part can be selected.
3 BACKGROUND

3.1 Background on assembling 3D Printed Ankle Foot Orthosis (AFOs)

The traditional AFOs are designed and fabricated to fit a range of patients, so they are called prefabricated orthoses. [3] These prefabricated orthoses could not provide enough comfort and orthotic function due to individual difference between patients. The rise of custom-fit AFOs can solve these problems and satisfy the comfort and functional requirements. The current process of design and fabrication of custom-fit AFOs are mostly done by manual labor. [3] The process is time-consuming, costly and in need of skilled orthotists. Moreover, the long and expensive process may be repeated several times for a whole rehabilitation therapy. However, the wide use of 3D scanning and Rapid Prototyping (RP) method in medical applications makes it possible to optimize the former fabrication process. The idea of combining these two technologies enables the fabrication of custom-fit AFOs to become timesaving and cheaper.

The idea of using RP and 3D scanning for AFOs originated from a capstone team at NU MIE under the guidance of Prof Mavroidis. [2] Current RP AFOs design would not have been possible without the contribution of those knowledgeable people. Richard Ranky has contributed a lot in AFOs design and embedding sensors on it. Mark Sivak, Seth Sivak, Alyssa Caddle, Kara Gilhooly, and Lauren Govoni have done a great job for their foundational work, as well as the contribution of Alan Argondizza in the design and fabrication of the AFOs testbed. [2]
AFOs is used to help various types of ankle disability such as posterior tibial tendon dysfunction, severe flatfoot, arthritis of the ankle, ankle sprains, lateral ankle instability, tendonitis, and drop foot. Current generic AFOs by Biomedical Mechantronics Laboratory at Northeastern University are fabricated based on a posterior leaf spring, which provides a full range of plantar and dorsiflexion, as shown in Figure 3-1. The posterior leaf spring AFOs are the proper rehabilitation device especially for drop foot, since it helps to return the foot to neutral position in the swing phase. Since a posterior leaf spring AFOs is a semi-flexible structure, it allows for normal tibial progression in stance phase. [5] However, a posterior leaf spring AFOs is not useful for certain types of patients, i.e. severe swelling or edema, unstable ankles, other ankle-foot deformities, etc. [4] A generic posterior leaf spring AFOs is unable to fit the patient’s deformed foot or foot with ailments. In this case, the 3D printed posterior leaf spring AFOs can be quickly made depending on every individual patient with a low cost. [4]
In the medical field, 3D scanner is widely used to capture 3D measurement of patient's face or body anatomy. [5] 3D scanner is able to quickly capture 3D measurement without contact to patient. It also provides consistent results and is easy to operate. The most important advantage of 3D scanner is easy transporting of 3D modeling. The RP technology in medical application has been used for medical devices, physical anatomical parts, and medical models. [6] Medical RP method benefits assured technology, fast turnaround, and use of U.S. Pharmacopeia approved material such as silicone rubber, epoxy, urethane, etc.

The combination of 3D scanning and rapid prototyping can provide patient-specific data input corresponding to anatomical features (via 3D scanning), as well as a means of producing a patient specific form output (via RP). [3] The procedures of the design and manufacturing the 3D printed AFOs are clearly shown in figure 3-2. Using 3D scanning to acquire digital models of surfaces for patient leg and ankle can transfer the patient-specific surface data to Computer Aided Design (CAD) software. Next step is building the model of custom-fit AFOs and optimizing the model in CAD software by unitizing the surface data. Rapid Prototyping would be used for the fabrication after the model of AFOs imports to RP machine, i.e. stereo lithography (SLA). [3] Meanwhile, the digital modeling and fabrication process could efficiently reduce the manual labor and the data is easily recorded and portable. All the stages of the AFOs modeling are stored digitally, it is easy to transmit the model and produce repeatedly.
Considering the gait phase of the patient who injured the right foot, after the right foot initially contacts on the ground, the center of mass will be moved gradually toward the right foot. This transition of body weight will help push off the AFOs backside. Because of this pushing off, the AFOs is bent and this bending stores the energy. Then, when the center of mass is moved to the left foot and the right toe is off from the ground, the stored energy will be released to return to the neutral ankle position. Therefore, the main function of the AFOs backside can be approximated as a linear spring does. The purpose of the AFOs design section was trying to provide a standard method to build a customized AFOs rather than a certain design. As we mentioned, our principle was to meet the individual requirement of the patient.
3.2 Background on Ankle Foot Orthoses (AFOs) testbed

The AFOs was used for the patient’s rehabilitation. Because the rehabilitation period for a patient was long, the AFOs would break or lose their function. For the patient, the most important thing was the safety. When the AFOs didn’t work well or would even break, we need to replace the AFOs as soon as possible. If not, the patient would fall down and get hurt. Fatigue test was a test which could provide the patients and the doctors an indication of when they need to replace the AFOs. The AFOs testbed was developed for the AFOs fatigue test. ASTM defines fatigue life, \( N_f \), as the number of stress cycles of a specified character that a specimen sustains before failure of a specified nature occurs. [3] Actually, fatigue test was applied for a certain material not an object. Therefore, the fatigue test machine was only used for material specimen. Obviously, since the AFOs was not a material specimen, a fatigue test machine was not suitable. Theoretically, the best way to test the AFOs was to find a patient, build AFOs for him and let the patient do the test. But, a patient should not participate a test which the AFOs hasn’t been proved. Any patient should not be hurt. To fix this problem, we need a testbed which can simulate a patient’s motion. When the testbed was applied a patient’s motion data, the AFOs can be tested as though a real patient was wearing it. However, there was one thing should be clarified that the principle purpose for this testbed was to do the AFOs fatigue test. We were not trying to develop an ankle simulated device. For this reason, some necessary simplification was applied. The AFOs test result was provided for proving the eligibility
of the testbed. In future work, we would do more tests for the AFOs and provide more
dynamic data of the AFOs.

4 Assembling 3D Printed Ankle Foot Orthosis (AFOs) Design

4.1 Overview of Assembling 3D Printed Ankle Foot Orthosis (AFOs)
Assembling 3D Printed Ankle Foot Orthosis was based on Rapid Prototype Ankle Foot
Orthosis (RP-AFOs). RP-AFOs were divided into three parts, the base, the functional bar
and the cuff. The base was built by 3D printed method after 3D scanning the patient’s
foot. We reduced the volume of 3D printed object in order to reduce the 3D printing
material usage. The workspace of 3D printer is limited. Reducing the volume of 3D
printed object means the 3D printer can build more objects at once. Then the cost of 3D
printed part was lowered by reducing the cost of material and increasing the amount of
printing at once. The cuff was a standard design which had a range of sizes to suit the
patient’s lower leg. A standard part is obviously cheaper than a custom one and the
patient can replace the cuff at any time with low cost. The key point of designing the
assembling AFOs is looking for a structure or method to replace the formal functional
part of RP-AFOs. FEA method was applied to verify if the new structure’s deformation
and stress under a certain torque could meet the patients’ needs.

4.2 The design requirements of assembling AFOs
To design a satisfactory AFOs, some requirements need to be clarified at first.

- The patient’s safety and comfort must be guaranteed
- The AFOs’ function must meet the patient’s needs
The durability of AFOs should be as long as possible.

The cost of AFOs in order to ease the financial burden of the patient was reduced.

The first requirement is necessary and it is the most important one. The AFOs serve the patient’s rehabilitation. The efficiency of rehabilitation takes a back seat to the patient’s safety. For example, if a patient fall down due to the defect of design, the design will be considered as a failure. As regards the comfort, the rehabilitation period is a long time, so the patient needs a comfortable device or brace which doesn't affect his/her daily life. Moreover some other factors need attention. Some patients may have an allergy to the 3D printed material or have abrasion after long wear. Therefore, when we design the AFOs, we should pay more attention to these kinds of problems.

The second requirement means that we need to develop a highly effective AFOs. The minimum requirement of function is to help the patient to adjust his/her gait to normal gait. In this Assembling 3D printed AFOs project, the deformation and stress under a certain torque are the indicator of whether the AFOs meets the requirement. If the stiffness is big enough to support the patient, the design would be consider as eligible. To meet all the requirements, we need to follow the doctor’s guidance.

For the third requirement, the durability of AFOs was much involved with the material. To determine the fatigue feature of AFOs, Finite Element Analysis (FEA) method is not suitable. A testbed which is used for fatigue testing is necessary. In the next section a testbed for stiffness and fatigue testing was designed and built.
The last requirement is the main purpose to design the assembling AFOs. Less material, easy manufacturing and standard part applying would lower the cost. In some conditions, the patient may need to adjust the stiffness by replacing the function part. For example, there were two materials with different stiffness. The patient needed high stiffness AFOs at the beginning of rehabilitation. After a period of treatment, the other material with lower stiffness will be used when the patient doesn’t need that much support anymore. We would like to find a range of materials to meet different statuses of rehabilitation for a patient.

4.3 Initial design of assembling AFOs

Figure 4-1 represents an overview of initial design. The formal integral RP-AFOs is a posterior leaf design. Based on the structure, the initial design is also a posterior leaf. The AFOs design has a simple structure and uses less material compared to the other kinds of AFOs structures. The base part is 3D printed and the cuff is standard and replaceable. For the functional part, the initial design has a bar which can rotate around the top of the base part. The concept of this design is to meet the different demands of patients. The shape and the length of the bar are easy to adjust, so we can adjust the stiffness of the bar. A simple model which we designed is easy for most materials to shape.
The detailed design (figure 4-2) at the connection part between the function bar and the base shows two promising features. Feature A is a small space between the bar and the base. If a compression spring was inserted in this space, the spring would be compressed when the bar was bended forward. Depending on what stiffness level the patient needed, the stiffness of AFOs was easy to adjust by applying different springs with a range of stiffness. Moreover, changing the number of spring which was inserted also could adjust the stiffness of AFOs. Theoretically, if the height of the space was long enough, the stiffness could be changed by adjusting the distance between the spring and the pivot of the bar.
Feature B is a mechanical stop for the safety of the patient. The patient doesn't need to bend backward too much, so feature B can help to prevent the foot drop in order to prevent the patient’s falling down and guarantee safety.

The initial design provided a concept that instead of using the functional bar to provide the torque, a spring also can work. Based on the initial design, there were three concepts which could be applied.

- Flexible bar with spring
- Rigid bar and spring
- Flexible bar only

For the first concept, the flexible bar and spring can work together. Since the concept was based on regular posterior leaf AFOs, the flexible bar played a leading role in providing the torque. The spring was an assistant for helping adjust the stiffness and make the AFOs more comfortable.
The second concept made a change that the flexible bar became a rigid bar. The change made the spring become the only torque provider. The rigid bar with spring concept could not meet the first requirement of designing AFOs, because the rigid bar made the patient uncomfortable and may cause the patient’s falling down. Flexible bar without a spring theoretically has no difference with RP-AFOs. We considered the flexible bar with spring concept and flexible bar only concept together, when we developed the flexible bar.

4.4 Development of Initial Design of AFOs
In section 3.3, we have provided two reasonable design concepts. One is flexible bar with spring and the other one is flexible bar only. We considered the spring first, because we need to figure out the exact function of spring. We made the connection between the base part and the bar become a rotation joint so that when the function bar rotated around the connection the spring could be assumed as the main torque provider. In what situation could the spring be the main torque provider? When the patient was walking, the function bar was bended forward by his leg and the connection joint would rotate first. The bar would not begin to bend until the spring could not be compressed anymore. That means the torque provided by the spring was less than the torque applied by the patient. For the flexible bar without spring concept, the rotation connection was a surplus. Because, as we mentioned before, the rotation connection was used for the spring. When the spring was not applied anymore, the connection joint between the base part and the bar became rigid.
Therefore, we classified the design concepts depending on the connection type. One is rotation connection and the other one is rigid connection.

4.4.1 Rotation Connection with spring

4.4.1.1 AFOs with compression spring

AFOs with compression spring used a compression spring or something like it to provide the torque. Figure 4-3 shows a design which was using a piece of rubber to work like a compression spring. Part A is a place for fixing the rubber. To simplify the design, the whole assembling structure was mounted directly to the back of AFOs base part.

![Diagram of AFOs with compression spring](image)

**Figure 4-3 AFOs with compression spring Overview**

The detailed design of the assembling structure is that it included three parts, front plates, back plate and bending rod. In order to easily assemble the rod to the plate, front plate had a clearance, as seen in figure 4-3(front). The clearance allowed the back plate to adjust its position relatively to the front plate during assembly process. In order to attach the assembling structure to AFOs, a flat surface of AFOs need to be develop. In addition,
the bolt size should not be too big, because a big bolt would make the patients feel uncomfortable.

The dimension of AFOs would also be an important factor. The proposed concept should be developed with proper dimensions, i.e. width, height and thickness of rod, mounting plate size, location of rubber, mechanical stopper, and bolts. Furthermore, the material of the rod should be selected by mechanical properties.

4.4.1.2 AFOs with torsion spring

The AFOs with torsion spring was a concept design with two separate rotational shafts and one torsion spring which was shown in Figure 4-4. Two shafts were connected to the bottom of AFOs to work as a revolute joint. The revolute joint could allow plantarflexion and dorsiflexion. The two-shaft structure also could be bended and allowed rod’s eversion and inversion. Applying a torsion spring which provided an assistant torque could help the ankle joint for either plantarflexion or dorsiflexion. The material of the AFOs applied was also an important factor. The flexible material applied to the shafts could also provide additional assistant torque for plantarflexion and dorsiflexion. Based on the nominal torque data, the desired spring constant could be calculated.
The AFOs with torsion spring had some disadvantages. The disadvantage of this design was that the large spring stiffness was required for providing the assistant torque. It was hard to customize a torsion spring with large coefficient due to the size limitation of the connection area. Even if the large stiffness torsion spring was used and the size limitation problem was solved, the contact area between the torsion spring and the AFOs base would be easy to break due to the stress concentration. In order to avoid the stress concentration, the thickness of the AFOs base and the radius of shaft should be increased. The increased dimension would reduce the patient’s comfort.

4.4.1.3 AFOs with Extension Spring

For both AFOs with compression spring design and AFOs with torsion spring design, the disadvantage was the connection area was too small. That means we could not apply a big spring. If we need to insert a big coefficient spring, the size of the spring must be big.
To solve this problem, the extension spring design could provide more space for connection area, so we designed several structures for AFOs with an extension spring.

Figure 4-5 AFOs with extension spring

Figure 4-5 shows a way to insert an extension spring. The red arrow indicates the extension spring. As we see, the extension spring design was easy for replacing the spring. By adjusting the hook location reference to the rod, different lengths of the springs could be applied. We also could design an adjustable hook which could be moved up and down on the rod in order to adjust the pre-extension of the spring. However, the design also had some drawbacks. First, if the spring was dislocated or broken, patients would be more likely to get a second injury. For example, when the rod was bended forward, the spring broke. At this time, the support to the patient would disappear immediately. And the rod’s movement with the patient’s leg would lose control. It would cause the patient’s falling down. Second, when an extension spring was working, the spring should be straight. For the extension spring design, when the rod was bended
forward, the spring cannot be straight all the time. The spring would contact the rod and cause some friction.

Figure 4-6 shows an optimized design for AFOs with extension spring. The optimized design could keep the spring straight and avoid the contact between the spring and the rod. The back cover of the connection area also could work as a mechanical stop to provide resistant torque when the rod was bended backward. The mechanical stop would help to avoid foot drop and keep the patient’s safety.

![Image of AFO design](image)

**Figure 4-6 Second design development**

Detailed design is shown in figure 4-7. The rod was designed to rotate along with the ankle joint. To limit the movement of the rod, the inner faces of the base and the cover can work as a mechanical stop to restrict the ankle rotational angle. When the rod rotated forward (dorsiflexion), the extension spring would provide resistant force as the spring was elongated. And the bottom of the rod would interfere with the two inner faces (inner face of base and inner face of cover). When the rod rotated backward (plantarflexion), the rod would interfere with the inner face of the cover.
Both of the forward and backward rotational angles could be adjusted by the distance between the inner face and the bar. Larger space allowed bigger range of rotational angle. In the rotational range, the spring generated assistant torque for patient’s ankle. In order to set desired assistant torque, the stiffness of spring and specific geometric dimension must be defined with desired assistant level.

### 4.4.2 AFOs with rigid connection
Rigid connection design means that the spring was removed and the flexible rod was rigidly mounted to the AFOs base (Figure 4-8). The rigid connection design was developed in terms of the material selection instead of the structural design. The possible materials of the concept are Polypropylene Homopolymer, carbon fiber and some other flexible materials.
Figure 4-8 AFOs with Rigid Mounting (left) and Embossed Rod (right)

Since the rigid mounting would cause stress concentration around the connection, especially between the AFOs base and the bottom of the bar, we needed to increase the dimensions of this part to strengthen the connection. To make the patients more comfortable, we tried to add some features in the structure. As shown in figure 4-9, the unique feature of the rigid design was inserting a compression spring between the cuff and the top of the rod. The added feature on the cuff could provide the patients more degrees of freedom instead of limiting the movement of the leg between forward and backward. The added spring should not affect to the assistant torque a lot. Because of the limited volume of the connection part based on the leg anatomy, the spring cannot extend or compress much. Therefore, the spring connection would work as a planar joint and it provided joint kinematics with 6 degrees of freedom between leg and foot. The length of the rod was assumed as 240 mm. When the rod was bended forward by 10 degrees, the displacement of the rod top would be $240 \text{ mm} \times \sin(10^\circ) = 41.68 \text{ mm}$. Therefore, the cuff could be moved about 42mm away from the top of the bar. The deflection of the bar
delivered assistant torque to patient's ankle joint. That's the rigid design's principle of work.

Figure 4-9 Planar Joint between the Cuff and Rod

4.5 Structure Design Conclusion
As we discussed before, two types of concept designs were provided. For the design with the spring, the function of the spring was to provide assistance torque for the patient's ankle. No matter what kind of springs we applied, the torque which was provided by the spring was too small compared to the torque provided by the function rod, so the FEA for the design with the spring was not necessary, due to the spring was only used for torque assistance. To deeply analyse the two types of concept designs, Finite Element Analysis (FEA) method was needed. Therefore, we will provide detailed analysis in the following sections. The base part was made by a certain selected 3D printed material and the cuff was standard. Hence, we only needed to analyze the mechanical properties of the rod. For the rod, there were two features that needed to be analyzed. Stress condition and deflection condition of the rod which were applied by a resume torque. Furthermore, the
effects of different dimensions and materials of the rod would be analyzed. We could not just simply provide a certain dimension and material for the rod, because different patients need different levels of support. We should provide a range of dimensions and materials to meet the different requirements.

The normative torque profile was measured on healthy subject by the medical center (Figure 4-6). The applied force to the top of the rod could produce equivalent ankle torque. FEA was performed with linear dynamic conditions and 25 unit force. The normative torque profile with 25 unit force was assumed as a virtual patient who can generate only 33.3% of healthy subject ankle torque when the body weight of healthy subject is assumed as 75 kg.

![Figure 4-6 Normative Torque Data from the Ankle Movement](image)
4.6 Dynamic Analysis

According to figure 4-6, 1.5 Nm/Kg is the peak torque in a gait cycle. As we assumed, the weight is 75Kg and the patient only can provide 33.3% torque of a healthy person, so the torque should be $1.5 \times 25 = 37.5$ Nm.

Figure 4-7 Dimension information of AFOs

Figure 4-7 shows some dimension information of the AFOs. The force which applied on the top of the rod is $37.5 \div (0.365 - 0.064) = 125$ N. When the peak torque was applied, the angle of ankle was about 12 degree. The deflection of the rod was from $301 \times \sin(10^\circ) = 52$mm to $301 \times \sin(15^\circ) = 78$mm. The principle of the FEA was to decide that when the dimension and material were decided, would the stress and deflection be in a required or safe range. In the following FEA sections, the rod would be considered as a beam, and ANSYS APDL would be used for FEA.
4.7 Finite Element Analysis of Rod

4.7.1 Deflection Analysis

As rod was considered as a beam, beam 188 element type was applied for FEA. Beam 188 method needed two parameters, Elastic Modulus and Poisson ratio, to complete the analysis. Due to the Poisson ratio was almost the same among the regular materials, we selected some materials mainly reference to the Elastic Modulus. According to the datasheet of the materials, different type of materials had huge difference in terms of the elastic modulus. Figure 4-7 shows a column chart about elastic modulus of some selected materials. For the Poisson ratio, most of the selected materials’ ratios were around 0.33.

![Figure 4-8 Elastic modulus of selected materials](image-url)
First of all, we made the dimension of the rod $5 \times 25 \times 265\text{mm}$. Some selected metal Materials, Aluminum 1060 Alloy, Aluminum 2014 Alloy and Aluminum 7079 Alloy, were applied. For the three selected materials, the elastic modulus was around $69000\text{MPa}$ and the Poisson ratio was 0.33. We could see, the deflection of the metal rod which was applied selected material under a 125N force was around 40mm. It means that the stiffness of the rod was a little higher than required. Therefore, we need to reduce the stiffness of the rod.

![Deflection of rod](image)

**Figure 4-9 Deflection of rod**

To adjust the stiffness of the rod, there were three methods.

1. Change the shape of the rod to a curving beam.
2. Adjust the dimension of the rod.
3. Change the materials of the rod.
For the first method, $0^\circ$ to $5^\circ$ curving of the rod were applied.

Table 4-1 ANSYS results for different degree

<table>
<thead>
<tr>
<th>Degree</th>
<th>ANSYS Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 degree</td>
<td><img src="image1.png" alt="Image" /></td>
</tr>
<tr>
<td>1 degree</td>
<td><img src="image2.png" alt="Image" /></td>
</tr>
<tr>
<td>2 degree</td>
<td><img src="image3.png" alt="Image" /></td>
</tr>
<tr>
<td>3 degree</td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
<tr>
<td>4 degree</td>
<td><img src="image5.png" alt="Image" /></td>
</tr>
<tr>
<td>5 degree</td>
<td><img src="image6.png" alt="Image" /></td>
</tr>
</tbody>
</table>
Figure 4-10 Deflections of 0 – 5 degree

Figure 4-10 shows the deflections of 0 – 5 degree were between 42.5mm to 42.6mm. It means pre-bending the beam could not increase or decrease the stiffness of the rod/beam. Therefore, method one was not proved and we would try to adjust the stiffness by adjusting the dimensions of the rod/beam and changing the materials.

For the second method, the rod was considered as a beam, so we could use mechanics of materials method to analysis. As we known, the maximal deflection function of a beam under a certain force should be

\[ y_p = \frac{Fl^3}{3EI} \]

\[ I = \frac{bh^3}{12} \]
\( y_p \) is the maximal deflection of the beam. \( F \) is the certain force which was applied. \( l \) is the length of the beam. \( E \) is the elastic modulus. \( I \) is the moment of inertia of the beam. \( b \) is the width and \( h \) is the height of the beam section. Hence, the deflection of the beam would be

\[
y_p = \frac{4Fl^3}{Eb^3h^3}
\]

According to the function, there are five variables. \( F \) was provided by the patient and \( l \) was decided by the length of the patient’s leg, so we cannot adjust them. \( E \) could be adjusted by changing the materials. \( b \) and \( h \) could be adjusted by changing the dimension of the beam section.

Firstly, we selected the Aluminum 1060 Alloy to do the analysis. The width of the section, \( b \), was changed from 5mm to 8mm to proceed the FEA. Table 3-2 shows the ANSYS results for different height of the section. According to the ANSYS results, the deflections of the rod/beam from 5mm to 8mm are 42.517mm, 24.608mm, 15.499mm and 10.385mm. According to the deflection function, the deflections of the rod/beam from 5mm to 8mm should be 43.15mm, 24.97mm, 15.73mm and 10.54mm.
Table 4-2 ANSYS results for different b

<table>
<thead>
<tr>
<th>h</th>
<th>ANSYS Results</th>
<th>Function Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>5mm</td>
<td><img src="h=5mm" alt="Image" /></td>
<td><img src="h=5mm" alt="Image" /></td>
</tr>
<tr>
<td>6mm</td>
<td><img src="h=6mm" alt="Image" /></td>
<td><img src="h=6mm" alt="Image" /></td>
</tr>
<tr>
<td>7mm</td>
<td><img src="h=7mm" alt="Image" /></td>
<td><img src="h=7mm" alt="Image" /></td>
</tr>
<tr>
<td>8mm</td>
<td><img src="h=8mm" alt="Image" /></td>
<td><img src="h=8mm" alt="Image" /></td>
</tr>
</tbody>
</table>

Figure 4-11 shows the comparison between ANSYS and function results. We could see the deflections which we got from ANSYS and function were almost the same, so we would use the function to get the curve of deflection directly.
Figure 4-12 shows the deflection vs. the height of the section. When the height changed from 3mm to 10mm, the deflection of the rod changed a lot. For the materials which had elastic modulus around 69000 MPa had a range of height to meet the requirement of the deflection from 52mm to 78mm. The width h and the height b had the same effects on the deflection. According to the function, changing the height h a little could change the deflection a lot which comparison to the width b. Therefore, we just consider the adjustment of the height h.
According to figure 4-8, the elastic modulus between different types of material changed a lot. Due to the second method for adjusting the stiffness was applied after we selected the material, adjusting the dimensions of beam section only could adjust the stiffness in a small range, and the results depended on if the elastic modulus could provide an eligible range of stiffness.

Figure 4-13 provided a deflection vs. elastic modulus curve based on the third method. The elastic modulus changed from 20000 MPa to 100000 MPa and the height h changed from 5mm to 8mm. According to the figure, the elastic modulus from 20000 to 100000 and the height from 4mm to 7mm would satisfy the requirement. The figure would give us an indication to decide what kind of material should be selected when a patient’s requirement was provided.
Therefore, if we were designing a rod for the virtual patient which we mentioned before, Aluminum Alloy may be the most suitable material. Some plastic materials such as ABS, Nylon and PEEK cannot be used, due to the small elastic module. For the other metal materials, such as cooper, brass and steel, the density were too big. It means that the rod was too heavy for the patients and it would make the patients uncomfortable. For the patients who need big torque assistance, Carbon fiber had very big elastic modulus and it was very light.

The deflection analysis only could provide a reference for material selection and rod design to meet the torque and displacement requirement of the patients. Stress analysis was needed to decide if the design was safe for the patients. When the torque was
applied, the maximal stress should not bigger than the yield stress, if not the rod would break and the patient would get hurt.

4.7.2 Stress Analysis

The stress analysis was necessary for the safety consideration. If the maximal stress under the certain torque was bigger than the yield stress, the rod would break. It means the AFOs could not provide enough torque to support the patient’s weight and the rod could not guarantee the patient’s safety. Rod breaking would make the patient fall down and get injury. As we mentioned before, the most important requirement was safety. The patient must not be hurt when he was wearing the AFOs. In section 3.7, there were three ways to adjust the stiffness, so in this section we would analysis if the three ways would affect the stress. Aluminum alloy was the most suitable metal material for the AFOs, so we applied it for FEA.

![Von Mises stress and Yield Strength Aluminum 1061 Alloy](image)

**Figure 4-14 Von Mises stress and Yield Strength**
Figure 4-14 shows Von Mises stress and Yield strength of Aluminum 1060 Alloy. The yield strength of 1060 was very small, so the von mises stresses of different thickness were much bigger than yield strength. But, according to figure 4-14, when the thickness was increased, the von mises stress was decreased. As we considered the rod as a beam, according to the maximal stress function

$$\sigma_{max} = \frac{F \cdot y_{max}}{I_z}$$

$$I_z = \frac{bh^3}{12}$$

Hence, the maximal stress function was

$$\sigma_{max} = \frac{12F \cdot l \cdot y_{max}}{bh^3}$$

Because the Von Mises stress was involved with maximal stress, when the dimension of beam section increased, the von mises stress would decrease. But as we discussed in section 3.7, the increase of dimension would also increase the stiffness. Therefore, when we needed to decrease the von mises stress of the beam, we only increase the dimension of the easy broken area. In another word, avoiding to increase the stiffness too much, we just strengthen the weak area of the beam.
The other method to decrease the von mises stress was to change the shape of the beam.

![Beam section](image)

**Figure 4-15 Von Mises stress by SolidWorks**

Figure 4-15 shows that when the shape of section was changed like that, the von mises stress would decrease a lot. Because the draft design should be decided based on the torque requirement first, the stress analysis must be done after the deflection analysis.

When we selected a series of materials, we could select the one which had highest yield strength to guarantee the safety of the patient. For the Aluminum Alloy series, the yield strength of #2018 was 317.1Mpa. As conclusion, when we decided to apply some certain materials with a range of elastic modulus to meet the torque requirement, we could select the material which yield strength would also meet the requirement. For the safety of the patients, increasing the dimension of the beam section and changing the shape of the section would distribute the stress concentration and decrease the Von Mises stress.

### 4.7.3 Conclusion of FEA results

There were two conclusions could be drawn from the FEA results. One was about the material selection and the other one was about the structural design. Firstly, metal
material such as Aluminum Alloy and Titanium had good performance in terms of the strength. However, these materials were not flexible enough and also made the patients uncomfortable. Therefore, the flexible material should be a better choice. Organic/plastic materials were reasonable selection and most of this kind of materials was flexible enough to control subjects’ own kinematics by themselves. But, according to the FEA result, some organic/plastic materials were too flexible. When the normative torque profile with 25 unit force was applied, the deflection of AFOs structure was too large and AFOs was bended too much. In another word, the rod was too easy to break and it would hurt the patients. Therefore, the material should be selected in order to keep the balance between flexibility and strength. For some patients with heavy ankle disability, the material such as Aluminum Alloy and Carbon fiber might be suitable. For the patients who need a little torque, organic/plastic material would be better. Moreover, because of the custom-fit design for the base, the material of the base must be produced by the 3D-printer. SLS Nylon 11 was selected for fabricating the base. The 3D printed materials were limited by the 3D printed machine. We only could choose the material from the datasheet which was provide by the supplier.

Secondly, both of two types of structural designs were available. Because the spring could provide a little torque to assist the function rod, the spring was only used to adjust the stiffness when we needed to change the stiffness a little. The hook for extension spring connection on the rod was easy to break due to the stress concentration. To
increase the height of the hook could solve this problem. Even though the relocation solved the stress concentration, it would lead to some other problems such as the extension oscillation of the long spring. Overall, ‘complex’ structure with joint and spring were not reliable assembly method without more developments and simple AFOs structure was performed well in terms of the flexibility and strength.

4.8 CONCLUSION AND FUTURE WORK
The assembling AFOs design was developed based on the concept that AFOs was manufactured in three parts and then the separate parts could be assembled. Two aspects were considered when we designed the new AFOs, the structure and the material. Some new structure designs were tried to provide more options for material selection, but the new structure only could provide a little torque compared to the function rod. Therefore, to find a material which could provide enough torque and, at the same time, be flexible enough was the main problem for the material selection. For different patients, the torque requirement would be different. We only could provide a series of materials to meet the stress and reflection requirement. For the base part, it was considered as a rapid prototyping product, so the material of the base part was different from the rod and it didn’t need us to analyze. Through the finite element analysis, it confirmed that the base part was less affected by the stress concentration. Therefore, the material of the AFOs base was selected by considering the patient’s comfort and capability of the rapid prototyping. SLS Nylon 11 was selected as a reliable material of the base. Since the rod
was not only manufactured by rapid prototyping method, the rod materials had more choices. The variety of possible materials, i.e. aluminum, titanium, polypropylene, carbon fiber, etc., were examined by FEA with an applied normative torque within 25 unit force. In terms of both flexibility and strength, carbon fiber and aluminum were selected as reliable materials of the AFOs rod.

More materials of the structure should be investigated and analyzed by FEA software. There is a method to find better material which provides both strength and flexibility. Instead of arbitrary selection of possible materials, possible materials are selected in terms of its mechanical properties. Based on the applied torque and ankle rotational angle, the property of material can be approximated calculated. Using approximated parameters, possible materials are selected and examined in terms of its strength and safety factor.
5  Human Ankle Motion Simulated Testbed Design

5.1  Testbed Design Overview

After we built AFOs prototypes, the best way to test the new design was to find a patient and test it. That was called human subject test. But, we should not ask a patient to do the test, an unproved design might hurt the patient. Instead of doing human subject test, a testbed could avoid injuring the patients by accident during the test. The function of testbed was to simulate a patient’s ankle motion, so the AFOs was like being tested by a real patient. But there was a key point which we should paid attention was that the device was a testbed rather than a simulation machine. In other words, it was used to test the AFOs not to simulate a human ankle. Therefore, in terms of the main function of the AFOs, the testbed only needed to simulate one degree of ankle motion which were dorsiflexion and plantar flexion. In 3D Gait Model 2354 which was distributed by OpenSim project group, there were six degrees of freedom in ankle. In order to simplify the motion of testbed, we removed five degrees of freedom. Only dorsiflexion and plantar flexion were left. The testbed was designed as a 2D Gait simulation device. After completing some coordinate transform in X-Y coordinate system, several testbed designs would be provided and analyzed in the following sections. In order to simplify the design process, we only built the most effective design and then a suited control system was applied.

5.2  Lower Limb Motion Analysis

5.2.1  OpenSim Overview

“OpenSim is a freely available user extensible software system that lets users develop models of musculoskeletal structures and create dynamic simulations of movement.” [1] At the beginning of the research, a motion/gait recognition system was not necessary. OpenSim can provide a range of models for regular gait. It is easy to know the motion information of gait at any time point in a gait cycle. The best function of OpenSim is when a motion file is loaded, any
coordinate data can be obtained. The value of position and degree verse time can be plotted by plot tool “plotter”. Then export data to txt. file. The values of position and degree can be calculated by Matlab. The data was used to complete coordinate transform and design the structure of testbed.

5.2.2 3D Gait Model 2354 Overview

3D Gait Model 2354 was a freely available gait model shared by OpenSim. The model is three-dimensional, 23-degree-of-freedom computer model of the human musculoskeletal system. There are two gait models can be downloaded from the website, gait 2392 and gait 2354. “The Gait 2392 model features 92 musculotendon actuators to represent 76 muscles in the lower extremities and torso. For the Gait2354 model, the number of muscles was reduced by Anderson to improve simulation speed for demonstrations and educational purposes.” [9] Analyzing the gait model is to design the structure of testbed. Hence, computing the forces of muscles is not included in the research. The simulated speed for demonstrations of gait 2354 is better than gait 239, so we selected the gait 2354.

The inertial parameters for the body segments in model 2354 are adapted from a 10-segment, 23 degree-of-freedom model developed by Frank C. Anderson and Marcus G. Pandy. In the Anderson and Pandy model, mass and inertial properties for all segments, except the hind feet and toes, are based on average anthropometric data obtained from five subjects (age 26 +/- 3 years, height 177 +/- 3 cm, and weight 70.1 +/- 7.8 kg). All the data are recorded according to the method described by McConville. The lengths of the body segments are taken from the Delp model. [9]

For the hind foot and toes, the mass, position of the center of mass and moments of inertia are found by representing the volume of each segment by a set of interconnected vertices, the coordinates of which are derived from measuring the surface of a size-10 tennis shoe. Assuming
a uniform density of 1.1 g/cm³ for the feet, the density is numerically integrated over the volume of each segment to find the mass. [9]

All the inertial parameters for the model are scaled by a factor of 1.05626. [3]

Table 5-1 Inertial parameters for the body segments included in the model

<table>
<thead>
<tr>
<th>Body segment</th>
<th>Mass (Kg)</th>
<th>Moments of inertia</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>xx</td>
</tr>
<tr>
<td>Torso</td>
<td>34.2366</td>
<td>1.4745</td>
</tr>
<tr>
<td>Pelvis</td>
<td>11.777</td>
<td>0.1028</td>
</tr>
<tr>
<td>Right femur</td>
<td>9.3014</td>
<td>0.1339</td>
</tr>
<tr>
<td>Right tibia</td>
<td>3.7075</td>
<td>0.0504</td>
</tr>
<tr>
<td>Right patella</td>
<td>0.0862</td>
<td>0.00000287</td>
</tr>
<tr>
<td>Right talus</td>
<td>0.1000</td>
<td>0.0010</td>
</tr>
<tr>
<td>Right calcaneus</td>
<td>1.250</td>
<td>0.0014</td>
</tr>
<tr>
<td>Right toe</td>
<td>0.2166</td>
<td>0.0001</td>
</tr>
<tr>
<td>Left femur</td>
<td>9.3014</td>
<td>0.1339</td>
</tr>
<tr>
<td>Left tibia</td>
<td>3.7075</td>
<td>0.0504</td>
</tr>
<tr>
<td>Left patella</td>
<td>0.0862</td>
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<td>Left talus</td>
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<td>0.0010</td>
</tr>
<tr>
<td>Left calcaneus</td>
<td>1.250</td>
<td>0.0014</td>
</tr>
</tbody>
</table>
5.2.3 Gait Cycle Analysis

Our principle was to design a testbed which can simulate the motion of human ankle, so we did the gait cycle analysis at first. First of all, to simplify the analysis, we only concerned on the motion of right leg. According to figure 5-4, there were two phases in one gait cycle, stance phase and swing phase. Stance phase included five key points: heel strike, foot flat, midstance, heel off and toe off. Swing phase included three key points: toe off, mid swing and heel strike.

![Gait Cycle Diagram](image)

**Figure 5-1 Demos, Gait Cycle [11]**

Stance phase was the phase when the foot kept contacting with the ground. In stance phase, there were three key points, heel strike, foot flat and heel off. Heel strike was the beginning phase of contacting the ground. Foot flat was the phase which the whole foot contacted with the ground. At the same time, the leg, which the contacting foot belonged to, became the main support of the body weight. Heel off was the ending phase of contacting the ground. In other words, after heel off phase, the foot was totally away from the ground.

Swing phase was the phase when the foot was totally away from the ground. In the swing phase, there were three key points: toe off, midswing and heel strike. Toe off was the beginning phase of leaving the ground. Midswing was the phase when the leg swings above the ground and at this
time, the other leg supported the body weight. Heel strike phase was the last phase that the foot was away from the ground.

Therefore, when we analyzed the Gait model 2354, we also divided the gait cycle in two phases. Table 5-2 shows the two phases and seven key points in one gait cycle when we made heel strike be the beginning of the gait cycle.

**Table 5-2 Gait Cycle in OpenSim (Gait model 2354)**

<table>
<thead>
<tr>
<th>Heel Strike</th>
<th>Foot Flat</th>
<th>Midstance</th>
<th>Heel off</th>
<th>Toe off</th>
<th>Midswing</th>
<th>Heel Strike</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stance phase</td>
<td>Swing phase</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

To analyze the Gait model 2354 by phase, we need to match the phases to the corresponding period in Gait model 2354. After observing the gait cycle in OpenSim, we got the following conclusions.

- **Heel strike:** the time point when the knee joint begins to bend right after it goes to straight
- **Foot flat:** the time point when the tibia is vertical to the ground reference frame
- **Heel off:** the time point when the knee joint becomes straight after foot flat
- **Midstance:** the period from foot flat to heel off, we don’t detail this point.
- **Toe off:** the time point when the femur is vertical to the ground reference frame
• Midswing: the period from toe off to the next heel strike, we also don’t detail this point

In the next section, to simplify the gait cycle, we would define the key points.

5.3 Coordinate Transform

First of all, the definition of reference frame should be clarified. The following definition was made by the OpenSim group.

• Pelvis: The pelvic reference frame is fixed at the midpoint of the line connecting the two anterior superior iliac spines

• Femur: The femoral frame is fixed at the center of the femoral head

• Tibia: The tibial frame is located at the midpoint of the line between the medial and lateral femoral epicondyles

• Patella: The patellar frame is located at the most distal point of the patella

• Talus: The talar frame is located at the midpoint of the line between the apices of the medical and lateral malleoli

• Calcaneus: The calcaneal frame is located at the most interior, lateral point on the posterior surface of the calcaneus

• Toe: The toe frame is located at the base of the second metatarsal

![Figure 5-2 OpenSim model](image)
To simplify the motion of leg, the pelvis was assumed as fixed in the coordinate system. According to figure 5-6, pelvis tilt, pelvis list and pelvis rotation were limited in -5 degree to 6 degree. The variations were small enough to be ignored. For figure 5-7, the position of pelvis in x, y and z direction were almost consistent. Hence, the pelvis would be fixed in the simplified gait model.

Figure 5-3 Pelvis motion vs. time

Figure 5-4 Pelvis position vs. time

There were 3 directions of rotation represented the motion of hip. Hip flexion represented the rotation around Z axis. Hip adduction and hip rotation represented the rotation around X axis and Y axis. According to figure 5-8, the variations of hip adduction and hip rotation was small
enough to be ignored. In figure 5-9, hip flexion of right leg had notable change. To simplify the gait model, only the rotation around was maintained, which was hip flexion. In other words, the motion of hip was limited in X-Y plane.

Due to the skeleton structure of knee joint, there was only one direction which was Z direction of rotation. In addition, the knee angle variation of right leg (figure 5-10) was huge. It could not be ignored.
Figure 5-7 Knee flexion right leg vs. time

Gait model 2354 was a simplified model, so the rotation of ankle only had one direction which was Z direction. The variation of ankle angle in figure 5-11 was huge, it also could not be ignored.

Figure 5-8 Ankle flexion right leg vs. time

For subtalar and mtp angle of right leg, the variation was almost zero (figure 5-12).

Figure 5-9 Subtalar & mtp angle right leg vs. time
To sum up, the motion of right leg was limited in X-Y plane. Pelvis was considered as fixed, so the pelvis would be considered as the origin in the reference frame of right leg. Hip flexion, knee flexion and ankle flexion were the three main rotation around Z axis. In the following context, \( hip_{fr} \), \( knee_{fr} \), and \( ankle_{fr} \) would be used as the representation of variables in calculation.

In Gait model 2354, hip flexion referenced to the pelvis. Knee flexion referenced to the hip. Ankle flexion referenced to the knee. Hence the first step of coordinate transform was to get the flexion data referenced to origin. In the following calculation, \( hip_f \), \( knee_f \) and \( ankle_f \) would be the representation of variables which referenced the origin.

**Figure 5-10 Indicates of the leg**

Due to the pelvis was fixed, hip flexion didn’t need to be changed.

\[
hip_f = hip_{fr}
\]

For the \( knee_f \) and the \( ankle_f \), the transform equations were the following.

\[
knee_f = hip_f + knee_{fr} \\
ankle_f = knee_f + ankle_{fr}
\]

Hence, the flexion which referenced to the pelvis was in figure 5-14.
Due to the function of testbed for AFOs, a structure which could simulate the human ankle was necessary. A concept of simulated human ankle design would be provided first. Figure 5-15 shows the concept design of ankle foot simulated structure. According to the structure, pelvis, hip and femur should be reduced. If pelvis was not included in the simulated system, a replaced origin should be selected. There were two options for the new origin. Gait model 2354 was based on Lower limb Extremity Model 2010. The bone dimensions of a group of males around 170cm were provided by Gordon et al. (1989). [10] After some calculation and implication, we got the following parameters (table 5-3). In the following sections, different origin was applied to analyze which method was better.

Table 5-3 estimated values of the leg

<table>
<thead>
<tr>
<th></th>
<th>estimated value/mm</th>
<th>estimated value/in</th>
</tr>
</thead>
<tbody>
<tr>
<td>$femur_l$ (femur length)</td>
<td>577.3</td>
<td>22.73</td>
</tr>
<tr>
<td>$tibia_l$ (tibia length)</td>
<td>437.7</td>
<td>17.23</td>
</tr>
<tr>
<td>$footA_1$ (foot_a length)</td>
<td>73.7</td>
<td>2.90</td>
</tr>
<tr>
<td>$footB_1$ (foot_b length)</td>
<td>196</td>
<td>7.72</td>
</tr>
<tr>
<td>$foot_h$ (foot height)</td>
<td>67.1</td>
<td>2.64</td>
</tr>
</tbody>
</table>
5.3.1 Fix the pelvis segment

First of all, we got the position data of knee joint, ankle joint, point A and point B (figure 5-13) on the foot, when we fixed the pelvis segment.

\[
\begin{align*}
knee_x &= femur_l \cdot \sin(hip_f) \\
knee_y &= femur_l \cdot \cos(hip_f) \\
ankle_x &= knee_x + tibia_r \cdot \sin(knee_f) \\
ankle_y &= knee_y - tibia_r \cdot \cos(knee_f)
\end{align*}
\]

\[
\begin{align*}
footA_x &= ankle_x + \sqrt{(footA_l^2 + foot_h^2)} \cdot \sin\left(ankle_f - \tan^{-1}\frac{footA_l}{foot_h}\right) \\
footA_y &= ankle_y - \sqrt{(footA_l^2 + foot_h^2)} \cdot \cos\left(ankle_f - \tan^{-1}\frac{footA_l}{foot_h}\right) \\
footB_x &= ankle_x + \sqrt{(footB_l^2 + foot_h^2)} \cdot \sin\left(ankle_f + \tan^{-1}\frac{footB_l}{foot_h}\right) \\
footB_y &= ankle_y - \sqrt{(footB_l^2 + foot_h^2)} \cdot \cos\left(ankle_f + \tan^{-1}\frac{footB_l}{foot_h}\right)
\end{align*}
\]

Figure 5-16 shows the locus of knee joint, ankle joint, point A and point B on foot. Origin point (0, 0) was the pelvis. Obviously, if we built a structure to simulate the whole leg of a human, the
range of x direction should be from -600mm to 500mm. It was too huge for a structure to test an AFOs. Therefore, we decided to reduce the pelvis segment and hip joint and fixed the knee joint as the origin in order to make the testbed be smaller size.

![Joint locus (pelvis fixed)](image)

**Figure 5-13 Joint locus (pelvis fixed)**

### 5.3.2 Fix the knee joint

To fix the knee joint, we need to reduce the offset of knee joint. Based on the position data in section 4.3.1, we made position of knee joint, ankle joint, point A and point B on foot minus position of knee joint. Then, the knee joint was fixed.

\[
\text{position1}_x = \text{position}_x - \text{knee}_x
\]

\[
\text{position1}_y = \text{position}_y - \text{knee}_y
\]

![Joint locus (knee fixed)](image)
Figure 5-14 Joint locus (knee fixed)

Figure 5-17 shows that the range of x variable is from -500mm to 350mm, and y is from -530mm to -150mm. The overlapping range of x variable for footA and footB is from -390mm to 100mm.

5.3.3 Fix point B on the foot

To fix point B on the foot, as what we did in section 4.3.2, we reduced the offset of point B. Also, based on the position data in section 4.3.1, we made the position of knee joint, ankle joint, point A and point B on foot minus position of point B. Then, point B was fixed.

\[
\begin{align*}
\text{position2}_x &= \text{position}_x - \text{footB}_x \\
\text{position2}_y &= \text{position}_y - \text{footB}_y
\end{align*}
\]

Figure 5-15 Joint locus (point B fixed)

5.4 Ankle Foot Orthosis analysis

As mentioned in section 4.2.3 Gait cycle analysis, there were two phases and seven key points in one gait cycle. We also used the same definition method to analyze the Gait model 2354. And then, in section 4.3 Coordinate transform, reference frame was defined. In order to match the phases to the corresponding period in Gait model 2354, we defined the seven key point in gait cycle based on the motion data in Gait model 2354 as the following method (pelvis fixed reference frame).
• Heel strike: the time point corresponding to zero value of $knee_{fr}$, when the value varies from negative to positive
• Foot flat: the time point when the value of $knee_f$ equals to zero
• Heel off: the time point when the value of $knee_{fr}$ varies from increase to decrease after foot flat
• Midstance: same with the formal definition, the period from foot flat to heel off
• Toe off: the time point when the value of $ankle_{fr}$ equals to zero
• Midswing: same with the formal definition, the period from toe off to the next heel strike

To make the heel strike be the beginning of the gait cycle in motion data, some adjustment of data was completed.

**Table 5-4 Gait cycle time line**

<table>
<thead>
<tr>
<th>Time/s</th>
<th>Heel strike</th>
<th>Foot flat</th>
<th>Midstance</th>
<th>Heel off</th>
<th>Toe off</th>
<th>Midswing</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0.566667</td>
<td>0.733333</td>
<td></td>
<td>1.16667</td>
<td>1.533333</td>
<td></td>
</tr>
</tbody>
</table>

Firstly, we should analyze that how the AFOs works in the stance phase. We assume a patient who wearing the AFOs begins the gait cycle at heel strike. In the period from heel strike to foot flat, the bottom surface of heel was the supporting point of AFOs. The point was considered as a fixed point on the ground. According to the motion data provided by Gait model 2354, in the period from 0.56667s to 0.733333s, $ankle_{fr}$ varied no more than 3 degree. Table 4-5 showed the simplified structure of AFOs which point A was fixed on the ground.

**Table 5-5 three foot status**
Heel strike | Foot flat/mid stance | Heel off/toe off

The brace was tied to the leg, and the base was tied to the foot. Actually the whole AFOs would function like a torque spring. To simplify the structure, we reduced the flexion feature of base and brace. Then the base and brace were assumed rigid. When the point A was fixed on the ground, only ankle could bend the torque spring. Meanwhile, maybe the weight of foot would contribute to the bending a little. As the ankle flexion changing a little, we could say the AFOs didn't begin to function in this period.

For the period from foot flat to heel off, the whole base could be considered as fixing on the ground. The ankle joint was like a pivot fixed on the ground. The whole body would bend the torque spring in AFOs. In this period, the AFOs began to function. For the AFOs, this period was a process of storing energy. Especially for the patient who could not provide enough torque by ankle, body weight helped a lot on bending the AFOs. In other words, the AFOs was transforming gravity potential energy of body weight to elastic potential energy.

After completing the storage of energy, the period from heel off to toe off was the process for the AFOs to release the energy. We considered point B was fixed on the ground. The torque spring
would help the ankle joint to rotate back to the neutral position which the \( ankle_{fr} \) value equaled to zero.

In conclusion, we assumed that in the period from heel strike to foot flat the AFOs didn't work. The period from foot flat to toe off was the function period for AFOs.

For the swing phase, the condition was similar with heel strike. In this period, the AFOs helped to prevent the foot drop. For the whole gait cycle, we concluded that the period from foot flat to toe off was the main process while the AFOs functioned. We reduced some motion data to suit the testbed design.

![Joint locus (reduced) (footB fixed)](image)

**Figure 5-16 Joint locus**

The reduced locus figure 5-19 shows that the curve of ankle position was almost an offset of curve of the footA position. It means that the footA didn't have relative movement with ankle joint. To simplify the test, we assumed that the AFOs structure was like the one shown in table 4-5. Based on the assumption, we ignored the deformation of base so that point footA was fixed and only ankle flexion was left. If we didn't apply the simplified structure of AFOs, we need to release the footA point. In the following section, a testbed for the two concepts was designed. We could test the AFOs based on two concepts and compare the results.
5.5 Testbed structure design
To simulate the motion of ankle, two structures, ankle structure and motion structure, were designed. Ankle structure was the structure for wearing the AFOs like a real patient’s foot. Motion structure was to work as the ground and provide a torque for bending the AFOs.

5.5.1 Ankle structure design

Figure 5-17 Ankle structure design
Figure 5-20 presented the rigid base structure design. In the structure, part A worked like the tibia of patient. Part B was like a human’s lower leg so that the AFOs could be tied to the tibia. Part C was bearing and shaft system which worked like ankle joint. Part D was to simulate the foot. It could fix the AFOs on the “ground” with high pressure to guarantee the full contact between AFOs and the “ground”. The potentiometer in part C was used for providing angle data of ankle flexion.
5.5.2 Motion structure design

Figure 5-18 Motion structure design

Figure 5-21 presented the motion structure design. The “ground” was the base part of the whole testbed. The “arm” with parallelogram structure transformed the torque which provided by a motor to a horizontal force. The parallelogram structure could make the calculation of torque which applied on the AFOs easy and accurate. The load cell at the end of “arm” was used for getting force data. Due to the small values of the velocity and acceleration of the arm, we could assume the testbed and AFOs system was at equilibrium state at any time. Therefore, the force value measuring by the load cell multiplied the height of “arm” was the torque which applied on the AFOs.

5.5.3 Testbed structure design conclusion

Figure 5-19 Testbed structure
Figure 5-22 showed the testbed structure. The AFOs was fixed on the “ground”, so the test for the gait cycle would be midstance period. The rod worked as a tibia needed have a smooth surface to reduce the friction, so the “arm” could keep more horizontal. The height of the arm was difficult to decide. Due to the height of the AFOs, if the arm height was too low, the angle of ankle would be limited in a small range. If the arm height was too high, the backlash would be large. It would make the result imprecise. Due to the big stiffness of AFOs, a DC motor with huge torque output was applied. The limited current of the motor was 3.6A and it was enough for the output. Two mechanical stops were applied to avoid breaking the AFOs.

5.6 Control system design
5.6.1 Overview
Control system design was involved with the human being’s walking pattern. Gait cycle analysis provided the evidence that a person was always trying to keep a gait cycle. It means that in normal condition (the person was not tired), a person would maintain the same motion of joints instead of force and torque. The testbed to simulate the motion of ankle should follow the same walking pattern with human. Therefore, the position control was applied to the control system. The experimental data which the hospital provided included angle data and torque data. During the test, if the torque control was needed, we would match the angle to torque so that the torque control was transformed to position control.
5.6.2 Equations of Motion

Figure 5-20 Testbed parameters

Figure 5-20 showed the testbed parameters. The testbed was divided by two parts, A and B. A part was considered as a motor and B part was considered as AFOs. Therefore, the Aluminum frame was considered as a part of the motor. When we did the calculation, the motor’s moment of inertia $J_m$ would include the inertia of the motor itself and the inertia of the frame.

$$T_m = J_m \ddot{\theta}_m + B \dot{\theta}_m + T_{AFO} \quad (5.1)$$

$$B = 0.2$$

Equation 5.1 was the main function of the whole system. $T_m$ was the output of the motor. $T_{AFO}$ was the input of AFOs. $J_m$ was the moment of inertia of the motor system. $B$ was the damping of the system.

$$T_m \cdot K_i = i_a \quad (5.2)$$

$$K_i = 30A/Nm$$
Equation 5.2 was the relationship between $T_m$ and $i_a$. The control type of the motor was current control, so we need to import current of motor $i_a$ to our control model. $K_t$ was the current constant we got from the motor datasheet.

$$T_{AFO} = F_{AFO} \cdot r$$

There was a load cell at the top of the frame. It could provide force data for pulling and pushing the AFOs. $r$ was the height of the load cell. The AFOs part was simplified, so we could use a simple equation to present the torque of bending the AFOs.

$$F_{AFO} = \theta_{AFO} \cdot k + b$$

A simple test was did by manual. We wanted to get a certain relationship between $F_{AFOs}$ and $\theta_{AFOs}$. Figure 5-21 shows the Curve fit result of matlab. We drew the frame by manual. When the force reached 9lb, the frame was hold on a while to record the data.

![Figure 5-21 Curve fit of manual test](image)

$$y = 0.01415x - 0.01046$$

The test results showed that the linearity of the system was good. After fitting, the $R$-square was 0.9419. It meant the linear curve fitting was credible.

We assumed that when the $\theta = 0^\circ$, the force of AFOs should be 0 N. Hence, $b$ should be zero.
\[ F_{AFO} = k \cdot \theta_{AFO} \quad (5.3) \]
\[ k = 23.6 \]

After some simplification and unit transformation, equation 5.3 is the final equation between the load cell and the angle of AFOs. \( r \) is the height of the load cell. To simplify the control system, we assumed \( \theta_m \) was small.

\[ r = L \cdot \cos(\theta_m) + 12.7\text{mm} \]

Due to the \( \theta_m \) was small,

\[ r = L + 12.7\text{mm} \]

\[ L = 16\text{in} = 406.4\text{mm} \]

Equation 5.4 was the relationship between \( \theta_{AFO} \) and \( \theta_m \). Due to the structure of the testbed, the load cell and the top of the AFOs had the same displacement.

\[ L_{AFO} \cdot \sin(\theta_{AFO}) = L \cdot \sin(\theta_m) \quad (5.4) \]

Due to the \( \theta_m \) and \( \theta_m \) are small,

\[ L_{AFO} \cdot \theta_{AFO} = L \cdot \theta_m \]

\[ L_{AFO} = 14\text{in} = 355.6\text{mm} \]

The moment of inertia of motor system was estimated by SolidWorks.

\[ J_m = 40000\text{kg} \cdot \text{mm}^2 \]

The final equation of motion was

\[ \frac{i_a}{K_t} = J_m \cdot \frac{L_{AFO}}{L} \ddot{\theta}_{AFO} + B \cdot \frac{L_{AFO}}{L} \dot{\theta}_{AFO} + k \cdot \theta_{AFO} \cdot r \]

### 5.6.3 Laplace Transform and Transfer Function

The Laplace transform for the current and angle of AFOs were

\[ \mathcal{L}(i_a) = I(s) \]
\[ \mathcal{L}(\ddot{\theta}_{AFO}) = s^2 \theta(s) \]
\[ L(\dot{\theta}_{AFO}) = s\theta(s) \]
\[ L(\theta_{AFO}) = \theta(s) \]

Hence, the Laplace transform of equation of motion was

\[
\frac{I(s)}{K_t} = I_m \cdot \frac{L_{AFO}}{L} s^2 \theta(s) + B \cdot \frac{L_{AFO}}{L} s\theta(s) + k \cdot \theta(s) \cdot r
\]

Then the open loop transfer function of the control system was

\[
\frac{\theta(s)}{I(s)} = \frac{1}{30 \cdot (35000 \cdot s^2 + 0.175s + 9890.76)}
\]

5.6.4 PID control tuning

The control type of the testbed is PID control. When we got the open loop transfer function of the control system, \(K_p\), \(K_i\) and \(K_d\) could be estimated by PID Tuner in Matlab.

![PID control tuning result](image)

**Figure 5-22 PD control tuning result**

PD control tuning result by PID Tuner is shown in figure 4-22. The response time was set by 0.3 second. \(K_p = 2.5 \times 10^7\) and \(K_d = 7.6 \times 10^6\).
PID control tuning result by PID Tuner is shown in figure 4-23. The response time was set by 0.3 second. $K_p = 2.5 \times 10^7, K_i = 3.0 \times 10^6$ and $K_d = 7.6 \times 10^6$.

For the two tuning results, $K_p, K_i$ and $K_d$ have huge order of magnitude. That’s because the moment of inertia was an estimated value. When we built the control system by Labview, the value for the three parameters could not be too huge. According to the tuning results, $0.3K_p \approx K_d$ and $K_p \approx K_i$. Therefore, when we tuned the control system, we used the relationship between $K_p, K_i$ and $K_d$ as a reference.

### 6 RP-AFOs test

#### 6.1 Overview

RP-AFOs test was used for proving the test design to be eligible. A gait cycle provided by OpenSim was applied as the input gait data. Our principle was when we input the gait cycle data, the AFOs’ ankle angle would behave referenced to the gait cycle which we input. Therefore, we did a lot of tuning work on the PID control system to make the gait cycle data which the testbed system behave close to the input data. Because the open loop control result was close to the final result, we tuned the PID control system based on the open loop system.
6.2 Labview program and PD control result

Figure 6-1 was the PID part of the Labview program. After tuning, we set $K_p = 0.44, K_i = 0$ and $K_d = 0.04$.

![Figure 6-1 Labview program](image)

Real-time result was shown in figure 6-2. The smooth curve was the input gait cycle and the other curve was the AFOs ankle angle measured by sensor. According to the result, the real-time result had 0.6 second delay. For the testbed, the delay would not affect the test result. We didn’t need the AFOs ankle to be same as the input gait at the same time. Our goal was to make the AFOs be bended like a real patient. Although, the control system had delay, the ankle angle’s change fitted the shape of the input curve well.

![Figure 6-2 real-time result](image)

After removing the delay for analysis, figure 6-3 shows the accuracy of the result. In the result most segments of the curve fitted the input gait well except the circled segments. The big error
was caused by two factors. The first factor was the backlash of the structure. Especially when the motor changed the direction in short time. The swing of the input curve would make the motor change its rotation direction several times in a short time. It would cause the shake of the structure, so the real-time curve would be like it in the red circle. The second factor was avoiding to break the AFOs, we didn’t apply a high current input. Therefore, when the system needed a high current input, the power could not provide enough current.

![Figure 6-3 matched results](image)

To solve the problem, more precise and strong mechanical parts, such as thicker shaft, bigger coupling and more precise bearing were needed. And another way to get a better result is to apply an advanced control method.
7 Conclusions and future work

The RP AFOs was a type of customized AFOs which can lower the cost and reduce the building time. Especially, 3D scanning data can be stored and emailed through the internet, so that whenever and wherever a patient needs a new AFOs, he would get it from a nearby 3D printer. That’s the most benefit of the RP AFOs. Assembling AFOs actually is an optimized design of RP AFOs. Reducing the 3D printed column and applying more standard parts can help lower the cost further and make the replacement of the AFOs much easier. As we discussed, we only can provide a method to build a customized AFOs rather than a standard design for most of the patients. As the applying of the 3D scanning and 3D printing method, the process of building the AFOs was changed. In the future, we hope we can collect more experience from the orthotist and conclude a standard process to build the AFOs by RP method. For the fatigue feature of the AFOs, we were trying to insert a low cost sensors, C-ship, into the AFOs, so that the patients and the doctors can even monitor the rehabilitation status at real time. C-ship is type of low-cost design option for interaction sensing as well as device self-diagnostics which was injected a conductive elastic gel into the specially designing of voids and cavities of a device. [4]

Therefore, in the future, the AFOs with sensor can provide accuracy indicate that when do they need to replace the AFOs. For the testbed, optimized control system will be applied. If it was needed, more degree of freedom motion will be added. If possible, a Kinect sensor could be used as a real-time gait cycle data collector, so that an ankle foot simulated system will be more perfect.
REFERENCE


