Proceedings of the ASME 2016 International Design Engineering Technical Conferences and
Computers and Information in Engineering Conference
IDETC/CIE 2016
August 21-24, 2016, Charlotte, North Carolina

DETC2016-60565

DESIGN AND PRELIMINARY TESTING OF A PROTOTYPE FOR EVALUATING LOWER LEG TRAJECTORY ERROR AS AN OPTIMIZATION METRIC FOR PROSTHETIC FEET

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ABSTRACT

This work presents the design and preliminary testing of a prosthetic foot prototype intended for evaluating a novel design objective for passive prosthetic feet, the Lower Leg Trajectory Error (LLTE). Thus far, all work regarding LLTE has been purely theoretical. The next step is to perform extensive clinical testing. An initial prototype consisting of rotational ankle and metatarsal joints with constant rotational stiffness was optimized and built, but at 2 kg it proved too heavy to use in clinical testing. A new conceptual foot architecture intended to reduce the weight of the final prototype is presented and optimized for LLTE. This foot consists of a rotational ankle joint with constant stiffness of 6.1 N·m/deg, a rigid structure extending 0.08 m from the ankle-knee axis, and a cantilever beam forefoot with bending stiffness 5.4 N·m^2. A prototype was built using machined delrin for the rigid structure, three parallel extension springs offset along a constant radius cam from a pin joint ankle, and machined nylon as the beam forefoot. In preliminary testing, it was determined that, despite efforts to minimize weight and size, this particular design was still too heavy and bulky as a result of the extension springs to be used in extensive clinical testing. Future work will focus on reducing the weight further by replacing linear extension springs with flexural elements before commencing with the clinical study.

INTRODUCTION

Bhagwan Mahaveer Viklang Sahayata Samiti (BMVSS), a distributor of prosthetic devices based in India, produces a low cost prosthetic foot called the Jaipur Foot that exceeds the performance of most prosthetic feet commonly used in the developing world, and even some used in developed countries [1, 2]. However, the Jaipur Foot is handmade, which results in variable quality consistency between feet and higher cost of production than for mass-manufactured feet. The motivation of this work is to design a new prosthetic foot that maintains the performance and cost of the original Jaipur Foot but is mass-manufacturable. In order to do so, it is necessary to understand what it is about the mechanical design of the Jaipur Foot that yields its high biomechanical performance.

Despite many studies comparing different prosthetic feet, multiple literature reviews have reached the same conclusion: how the mechanical behaviour of a passive prosthetic foot affects the biomechanical functionality is not well understood [3–6]. Despite being ten years old, these literature reviews still represent the state of the science, possibly because the focus of prosthetic design in academia has shifted from passive prosthetics to robotic prosthetics.

One metric, roll-over geometry, has stood out as a prominent design objective for passive prosthetic feet over the past decade. Roll-over geometry is defined as the path of the center of pressure

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along the bottom of the foot from heel strike to opposite heel strike in the moving reference frame defined by the ankle and the knee (the ankle-knee reference frame) [7]. Physiological roll-over geometries are similar for persons with a given leg length. These shapes have been shown to remain unchanged as walking speed, shoe heel height, and added torso weight are varied [8–10]. Some studies have suggested that prosthetic feet with roll-over geometries that mimic physiological roll-over geometries result in more symmetric gait [7] and higher metabolic efficiency while walking [11, 12].

However, our recent theoretical work has shown that roll-over geometry is not sufficient in characterizing prosthetic feet, as it does not fully describe lower leg kinematics. As a result, it is possible for two different prosthetic feet to have identical roll-over geometries, but very different lower leg kinematics during walking. In previous work, we proposed a novel prosthetic foot design objective, the Lower Leg Trajectory Error (LLTE) [13]. This metric incorporates both the roll-over geometry of the foot and the orientation of the lower leg segment in the laboratory reference frame throughout a step, thus fully describing the lower leg kinematics.

Thus far, all work regarding LLTE has been purely theoretical. The next step in moving towards using LLTE in the design of commercial prosthetic limbs is to clinically test the validity of LLTE as a design objective by building a prototype optimized for LLTE. An initial prototype was built in 2015 [14], has been built. This prototype consisted of rotational ankle and metatarsal joints with constant rotational stiffnesses optimized for LLTE. The joint stiffnesses were provided by linear extension springs offset from the ankle joint and compression springs offset from the metatarsal joint. While theoretically a very simple design, the resulting prototype had a mass of 2.06 kg, twice that of the Jaipur Foot, which is already heavier than most existing prosthetic feet.

The goal of this work is to produce a prototype prosthetic foot that can be used in a large scale gait analysis study to test the viability of LLTE as a design objective for prosthetic feet. To do so, a new conceptual prosthetic foot architecture consisting of a rotational ankle joint and a cantilever beam forefoot is presented. The design variables of this new architecture, namely the rotational stiffness of the ankle and the bending stiffness of the forefoot, are optimized for LLTE. The considerations in building a physical prototype based on this theoretical design are discussed, and the resulting prototype is presented. Qualitative feedback from preliminary testing is reported and discussed.

PROTOTYPE CONCEPT AND OPTIMIZATION

The conceptual architecture is similar to the previous prototype with rotational joints at the ankle and the metatarsal, but rather than a metatarsal joint, the new prototype has a compliant cantilever beam forefoot, which eliminates the need for the linear compression springs at the metatarsal joint and consequently reduces the weight (Fig. 1). The design variables available to be optimized for LLTE are the ankle joint rotational stiffness, k_{ank}, the length of the rigid structure extending from the ankle-knee axis, d_{rigid}, and the forefoot beam bending stiffness, EI. In this work, the height of the ankle rotational joint, h, is a parameter fixed at 0.08 m. This height was chosen to best approximate the center of rotation of a physiological ankle joint based on published gait data [15].

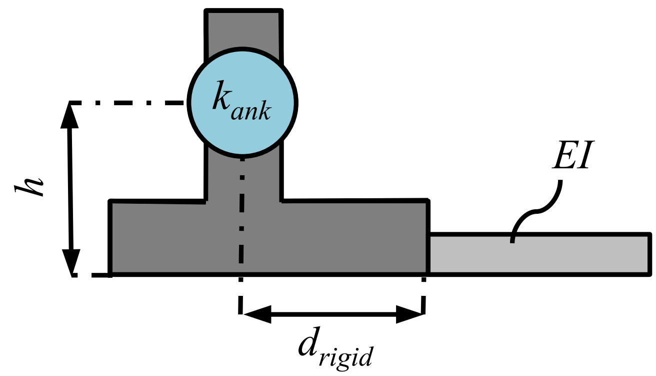


FIGURE 1: CONCEPTUAL PROSTHETIC FOOT ARCHITECTURE.

Calculation of LLTE

In order to calculate the LLTE, a set of representative physiological gait data is required. The vertical and horizontal ground reaction forces, GRF_y and GRF_x respectively, and the instantaneous center of pressure along the ground, d_{cp}, are used as inputs to calculate the deformed shape of the foot/ankle complex througout stance phase. The center of pressure and all other spatial coordinates in the laboratory reference frame used in this work are measured from a reference point directly below the ankle when the bottom of the foot is in contact with the ground.

The LLTE measures how well the resulting modeled lower leg kinematics match target physiological kinematic data. For this study, a set of published physiological gait data for a subject of body mass 56.7 kg was used. [15] The LLTE is a root-mean-square error comparing the trajectory of the lower leg segment of the modeled prototype to a target physiological trajectory, defined as

$$LLTE \equiv \left[\frac{1}{N} \sum_{n=1}^{N} \left\{ \left(\frac{x_n - \hat{x}_n}{\hat{x}_{max} - \hat{x}_{min}} \right)^2 + \left(\frac{y_n - \hat{y}_n}{\hat{y}_{max} - \hat{y}_{min}} \right)^2 + \left(\frac{\theta_n - \hat{\theta}_n}{\hat{\theta}_{max} - \hat{\theta}_{min}} \right)^2 \right\} \right]^{\frac{1}{2}}, \quad (1)$$

where x and y are the horizontal and vertical positions of the ankle joint respectively and θ is the orientation of the ankle-knee axis with respect to the vertical. The variables $\hat{x}$, $\hat{y}$ and $\hat{\theta}$ refer to the physiological data. The error in each coordinate is normalized by the range of that coordinate in the physiological data over the portion of the step included in the analysis. The subscript n refers to each time interval, with total number of time intervals N. [13]

For a given set of design variables (that is, ankle stiffness and forefoot bending stiffness), the coordinates x, y and θ were calculated for each time interval using the published ground reaction forces and instantaneous centers of pressure as inputs. When the center of pressure is along the rigid structure, that is, $d_{cp} < d_{rigid}$ (Fig. 2), the moment at the ankle joint is given by

$$M_{ank} = GRF_y \cdot d_{cp} + GRF_x \cdot h. \quad (2)$$

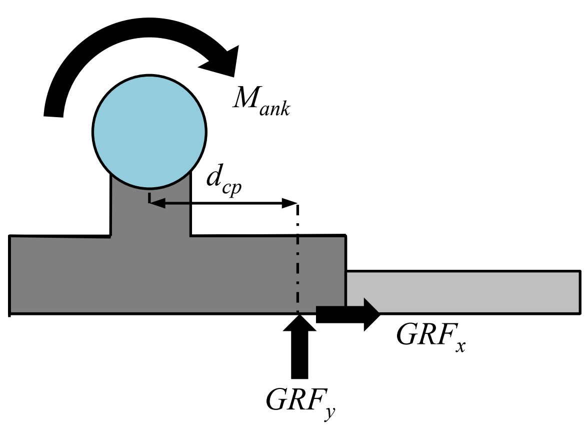


FIGURE 2: FREE BODY DIAGRAM SHOWING CALCULATION OF ANKLE MOMENT FROM GROUND REACTION FORCES WHEN CENTER OF PRESSURE ACTS ON RIGID STRUCTURE.

For a given ankle joint stiffness, k_{ank}, the resulting rotation at the ankle joint is

$$\theta_{ank} = \frac{M_{ank}}{k_{ank}}. \quad (3)$$

Because the bottom of the rigid structure of the foot must be in contact with the ground in order to not contradict the center of pressure location used as an input, the angle of the lower leg segment, θ, is equal to the ankle angle, θ_{ank}. The horizontal and vertical positions of the ankle are $x = 0$ and $y = h$ respectively for all times when the center of pressure is along the rigid portion of the foot.

When the center of pressure progresses beyond the rigid structure to the compliant beam, this calculation becomes more complex. In this case, the angle of deflection of the beam forefoot, or θ_{foot}, must also be calculated (Fig. 3. To find θ_{foot}, the magnitude of the force acting transverse to the beam, F_{trans} , must be found. However, since the inputs to the model are the ground reaction forces in the laboratory reference frame, the magnitude of the transverse force cannot be found without knowing θ_{foot}. Hence the deformed shape of the beam under the ground reaction forces was calculated iteratively.

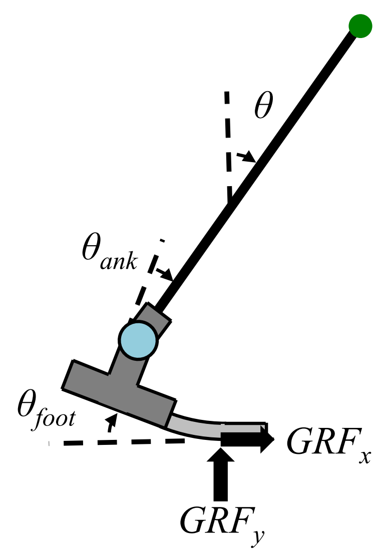


FIGURE 3: FREE BODY DIAGRAM FOR GROUND REACTION FOCES ACTING ON COMPLIANT BEAM FOREFOOT.

First, the load transverse to the beam was calculated assuming the beam did not deform at all, or $\theta_{foot} = 0$. Then

$$F_{trans} = GRF_y. \tag{4}$$

This transverse load was then used to calculate a second iterative value for θ_{foot}, by

$$\theta_{foot} = \frac{F_{trans}\left(d_{cp} - d_{rigid}\right)^2}{2EI}. \tag{5}$$

The new transverse load was then found with

$$F_{trans} = GRF_y \cdot \cos\theta_{foot} + GRF_x \cdot \sin\theta_{foot}. \tag{6}$$

These calculations of θ_{foot} and F_{trans} were repeated using Eqns. (5) and (6) until subsequent values of θ_{foot} differed by less than 0.5 degrees.

It should be noted that eqn. (5) is only valid for small deflections for which $\theta_{foot} \approx \tan\theta_{foot}$. For particularly small beam bending stiffness values, this equation no longer accurately represents the physical beam when the ground reaction force acts at the end of the toe. However, in the range of bending stiffnesses, beam lengths, and transverse forces considered here, deflections are small and eqn. (5) is appropriate.

The moment about the ankle was then calculated with

$$M_{ank} = F_{trans} \cdot d_{cp} + F_{axial} \cdot h \tag{7}$$

where

$$F_{axial} = -GRF_y \cdot \sin\theta_{foot} + GRF_x \cdot \cos\theta_{foot}. \tag{8}$$

Equation (3) was used to find the ankle angle, θ_{ank}. The lower leg angle, θ, was given by

$$\theta = \theta_{ank} + \theta_{foot}. \tag{9}$$

The horizontal and vertical coordinates of the ankle were calculated as

$$x = d_{cp} \cdot \left(1 - \cos\theta_{foot}\right) \tag{10}$$

and

$$y = d_{cp} \cdot \sin\theta_{foot} + h \cdot \cos\theta_{foot}. \tag{11}$$

Through eqns. (2) through (11), x, y, and θ were calculated for each time interval from foot flat to late stance. Using these coordinates, the LLTE was calculated for the given set of design variable values.

Optimization

The design variables, k_{ank}, d_{rigid}, and EI, were optimized heuristically through grid sampling. Each design variable was systematically varied over a range of reasonable values. The LLTE value was calculated for each possible combination of design variables. The set of design variables giving the lowest LLTE value was taken to be the optimal design.

The minimum LLTE value, 0.159, was calculated for k_{ank} = 6.1 N·m/deg, d_{rigid} = 0.08 m, and EI = 5.4 N·m^2. To depict the dependence of the LLTE value on each of the design variables, Fig. 4 shows the LLTE values found for a slice of the design space for which d_{rigid} is held constant at the optimal value of 0.08 m.

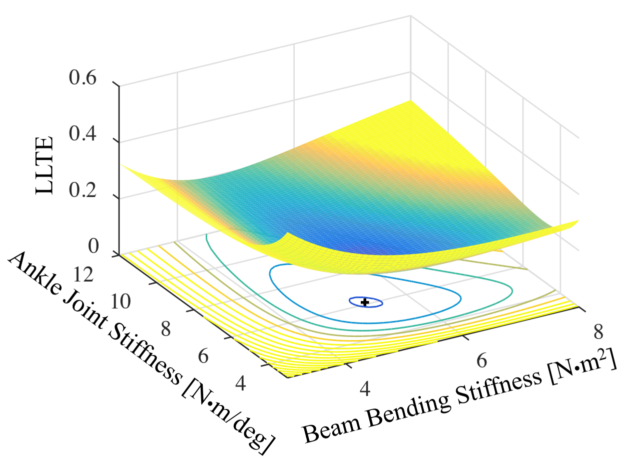
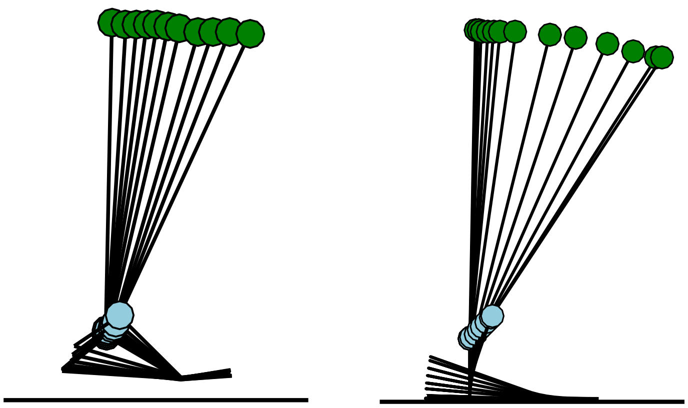


FIGURE 4: LLTE VALUES FOR SLICE OF DESIGN SPACE FOR WHICH d_{rigid} IS HELD CONSTANT AT d_{rigid} = 0.08 m.

The resulting lower leg trajectory is graphically compared to the target physiological lower leg trajectory in Fig. 5. Each of the individual kinematic coordinates, x, y, and θ are plotted against the corresponding physiological coordinates in Fig. 6.



(a) Physiological Trajectory (b) Optimal Model Trajectory

FIGURE 5: GRAPHICAL COMPARISON OF OPTIMAL FOOT DESIGN LOWER LEG TRAJECTORY TO PHYSIOLOGICAL LOWER LEG TRAJECTORY.

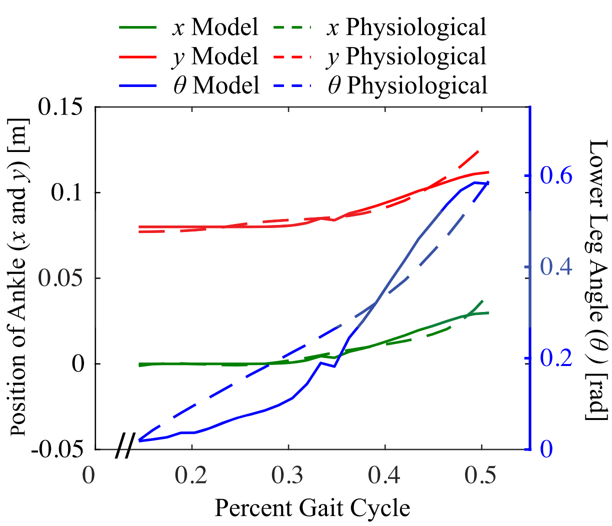


FIGURE 6: INDIVIDUAL SPATIAL COORDINATES OF OPTIMAL FOOT DESIGN COMPARED TO PHYSIOLOGICAL TARGET VALUES THROUGHOUT STANCE PHASE.

MECHANICAL DESIGN

In order to clinically validate the theoretical work suggesting the LLTE value as a design objective for prosthetic feet, it is necessary to design, build and test a prosthetic foot based on the optimal design found in the previous section. The goal is to design a proof-of-concept prototype as quickly as possible without spending time on details that are irrelevant in early-stage design, such as appearance, long-term durability, and mass-manufacturability. A solid model of the prototype designed for this purpose is shown in Fig. 7. The rigid structural components were machined from delrin. The ankle joint rotates about a steel pin. Extension springs offset behind the ankle joint at a constant radius provide the ankle joint rotational stiffness. The compliant beam forefoot was made from nylon and was fixed to the rigid delrin structure with machine screws fastened directly into tapped holes in the delrin. At BMVSS, the standard method of attaching the prosthetic socket to the Jaipur Foot is to heat the plastic exoskeletal socket until it becomes pliable, then slide the socket over the ankle, which consists of a wooden block inside of the rubber exterior of the foot. Four radial wood screws secure the socket to the ankle. To allow technicians at BMVSS to use this same method of attachment in prototype testing, a wooden ankle block of similar size and shape to the Jaipur Foot ankle was mounted to the top of the delrin structure. After a wooden ankle block is used, it can be replaced so no cracking occurs around mounting holes from previous tests. The prototype as built has a mass of 1.24 g, which is approximately 40% less than the mass of the previous prototype. This reduction in mass is due to the new architecture, which no longer requires metal compression springs at the metatarsal joint nor a rigid toe structure.

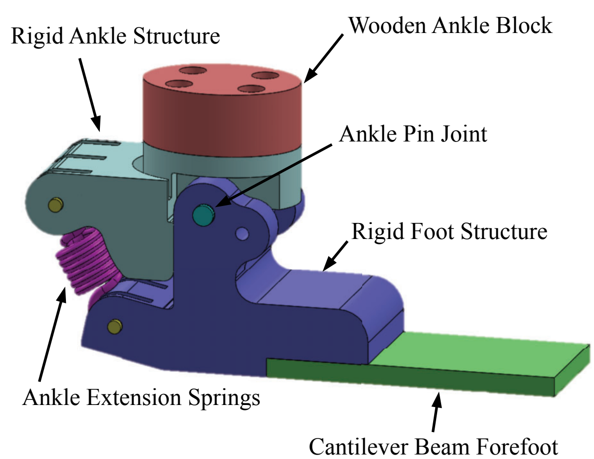


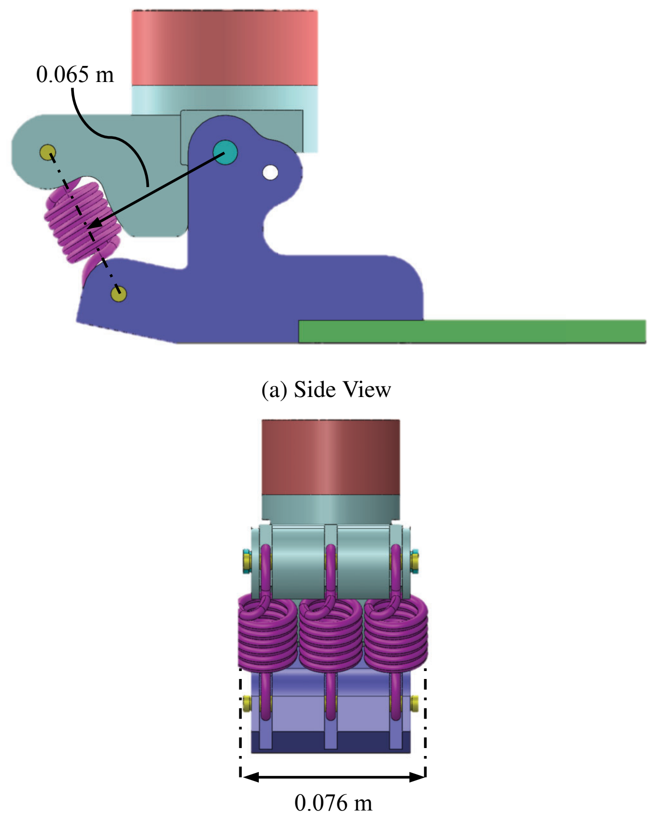
FIGURE 7: SOLID MODEL OF PROTOTYPE DESIGNED BASED ON LLTE OPTIMIZATION.

Spring Selection and Considerations

For speed and ease of design, off-the-shelf springs were used to provide the ankle joint rotational stiffness. Based on the above analysis, the ankle joint required a rotational stiffness of 6.1 N·m/deg and at least 10 degrees of rotation before yield. Additionally, the entire mechanism needed to be as compact and light weight as possible such that it did not interfere with gait nor add significant mass not accounted for in the analysis. These requirements immediately preclude the use of torsion springs, as those springs of sufficient stiffness were far too bulky to fit within the approximate size and shape of a prosthetic foot. Linear compression springs were also considered, as they can be small and very stiff, but constraining the ends of the compression springs in such a way as to achieve constant rotational stiffness about the ankle joint throughout large rotations proved problematic. Thus linear extension springs were chosen.

The extension springs were mounted using pins passing through hooks at either end, so as the ankle joint dorsiflexed, the hooks were free to rotate on the pin joints to avoid being over-constrained. The side of the springs rested along a constant radius cam. In this way, the extension force of the spring had a constant moment arm about the ankle joint even over large rotations, resulting in a constant rotational stiffness. The final spring configuration was selected to maximize range of motion with minimal total mass. Ultimately, three springs, each of linear stiffness 27846 N/m, were used in parallel, offset from the ankle joint by a radius of 0.065 m (Fig. 8). In this configuration, the ankle could dorsiflex 14.8 degrees before reaching the manufacturer's recommended maximum extension. The total mass and width of all three springs were 144.8 g and 0.076 m respectively.

The springs were mounted at an angle rather than vertically, as was done in the earlier prototype, to reduce the total volume and, consequently, mass, of the structure required to support them.



(a) Side View

(b) Rear View

FIGURE 8: LINEAR EXTENSION SPRING CONFIGURATION USED TO PRODUCE CONSTANT ANKLE JOINT ROTATIONAL STIFFNESS OF k_{ank} = 6.1 N$\cdot$ m/deg.

Cantilever Beam Forefoot Design

To replicate approximate physiological foot geometry, the beam forefoot was chosen to be 0.064 m wide and 0.07 m long. Materials considered were Delrin, ABS, nylon, polycarbonate, aluminum and steel. To produce the specified beam bending stiffness $EI = 5.4$ N$\cdot$m^2, the required thickness of the beam for each material was calculated. For those thicknesses, the maximum force that could be applied to the tip of the beam before yielding occured was calculated. The nylon beam could withstand the highest load before yielding. This result was not surprising, as nylon's high ratio of yield strength to Young's modulus makes it a particularly good flexural material. Thus the beam forefoot was constructed of nylon with thickness 0.008 m.

Preliminary Testing

The prototype was brought to India for an initial round of testing with our partners at BMVSS. The purpose of this testing was not yet to validate the theoretical LLTE work, but rather to determine the suitability of this prototype for use in a larger-scale gait analysis study.

The prototype was fitted on three male subjects with unilateral transtibial amputations who primarily use the Jaipur Foot. Apart from the amputations, the subjects had no further pathologies. The subjects were asked to walk on flat ground using the prototype until they felt comfortable with it, at which point they were asked to walk up and down stairs and ramps. After 30 minutes to an hour using the prototype, the subjects were asked qualitatively what they liked and disliked about the prototype. The subjects liked the energy storage and return aspects of the foot relative to the Jaipur Foot, which is purely dissipative. Dislikes primarily focused on the appearance and the weight of the prototype. Despite a 40% reduction in mass relative to the previous prototype, the prototype was still too heavy, particularly in the region posterior to the ankle, to claim that the weight of the prototype did not negatively affect the user's gait, potentially negating the benefits of the optimized LLTE. Because need for improvement was identified from the first three subjects, no further testing was necessary with this particular prototype. Further reduction in weight is required before proceeding with a larger scale gait analysis study to determine the validity of the theoretical work.

DISCUSSION AND CONCLUSION

As mentioned throughout this work, LLTE remains a theoretical objective function for prosthetic foot design. Clinical validation is required to show whether it provides a means of connecting the mechanical behaviour of a prosthetic foot to biomechanical functionality when used in practice.

Calculation of LLTE requires a set of ground reaction force and instantaneous center of pressure data as inputs and corresponding kinematic gait data as target outputs. In this study, a set of published physiological gait data for typical, unimpaired walking was used. The ground reaction forces used as inputs here are therefore different than what would be expected on the prototype, as it is well known that persons with amputations exhibit different gait characteristics than persons without amputations. However, it is also known that the gait of a person using a lower limb prosthesis is affected by the prosthesis itself. Thus rather than start with input data collected from persons with amputations that necessarily caries with it attributes of a different prosthetic foot, the physiological, unimpaired data is used as a starting point. The design of the prototype can then be refined through an iterative process. Once a foot is made based on unimpaired data, it can be tested on a group of individuals with amputations and the ground reaction forces measured. These measured ground reaction forces can then be used as inputs with the same set of target gait kinematic outputs to re-design the foot. This process can be repeated until the input ground reaction forces used to design the foot and the ground reaction forces observed in testing the foot converge.

Similarly, because the published gait data came from a single subject with body mass 56.7 kg, the optimal foot design is only optimal for persons of similar body mass. In order to validate LLTE as a design objective for prosthetic feet, the prototype will have to be customized to fit the body mass and foot length of the subjects in the study. This customization can be done using the same method as described here with a different set of gait data as inputs.

This paper presented the theoretical optimization, physical design, and preliminary testing of a prototype prosthetic foot to evaluate the effectiveness of Lower Leg Trajectory Error as a design objective. A conceptual foot architecture intended to reduce the weight of a previous prototype was presented. The calculation of the LLTE was described, and the design variables, that is, the ankle stiffness, the length of the rigid structure, and the bending stiffness of the compliant beam forefoot, were optimized. The design with the minimum LLTE value, 0.159, was found for k_{ank} = 6.1 N·m/deg, d_{rigid} = 0.08 m, and EI = 5.4 N·m^2. A physical prototype that meets these specifications was presented. Preliminary testing revealed that, despite a significant weight reduction from the previous prototype, the new prototype is still too heavy to be used in a large-scale study to validate LLTE as a design objective. Future work will focus on eliminating the use of heavy linear extension springs by designing integrated geometries that exhibit similar behavior through flexural elements.

ACKNOWLEDGMENT

Funding for this project was provided by the MIT Tata Center for Technology and Design and the MIT Department of Mechanical Engineering. The authors would like to thank Mr. DR Mehta, Dr. Pooja Mukul, and Dr. MK Mathur at BMVSS for their continued support of this work.

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