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DESIGN AND TESTING OF A PROSTHETIC FOOT PROTOTYPE WITH INTERCHANGEABLE CUSTOM ROTATIONAL SPRINGS TO ADJUST ANKLE STIFFNESS FOR EVALUATING LOWER LEG TRAJECTORY ERROR, AN OPTIMIZATION METRIC FOR PROSTHETIC FEET

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ABSTRACT

A prosthetic foot prototype intended for evaluating a novel design objective for passive prosthetic feet, the Lower Leg Trajectory Error (LLTE), is presented. This metric enables the optimization of prosthetic feet by modeling the trajectory of the lower leg segment throughout a step for a given prosthetic foot and selecting design variables to minimize the error between this trajectory and target physiological lower leg kinematics. Thus far, previous work on the LLTE has mainly focused on optimizing conceptual foot architectures. To further study this metric, extensive clinical testing on prototypes optimized using this method has to be performed. Initial prototypes replicating the LLTE-optimal designs in previous work were optimized and built, but at 1.3 to 2.1 kg they proved too heavy and bulky to be considered for testing. A new, fully-characterized foot design reducing the weight of the final prototype while enabling ankle stiffness to be varied is presented and optimized for LLTE.

The novel merits of this foot are that it can replicate a similar quasi-stiffness and range of motion of a physiological ankle, and be tested with variable ankle stiffnesses to test their effect on LLTE. The foot consists of a rotational ankle joint with interchangeable U-shaped constant stiffness springs ranging from 1.5 Nm/deg to 16 Nm/deg, a rigid structure extending 0.093 m from the ankle-knee axis, and a cantilever beam forefoot with a bending stiffness of 16 Nm². The prototype was built using machined acetal resin for the rigid structure, custom nylon springs for the ankle, and a nylon beam forefoot. In preliminary testing, this design performed as predicted and its modularity allowed us to rapidly change the springs to vary the ankle stiffness of the foot. Qualitative feedback from preliminary testing showed that this design is ready to be used in larger-scale studies. In future work, extensive clinical studies with testing different ankle stiffnesses will be conducted to validate the optimization method using the LLTE as a design objective.

INTRODUCTION

Despite many studies comparing different prosthetic feet, multiple literature reviews have reached the same conclusion: there is little understanding of how a passive prosthetic foot design affects the gait of an amputee [1–4]. In previous work by the authors, a novel prosthetic foot design objective was proposed, the Lower Leg Trajectory Error (LLTE) [5]. This metric enables the optimization of prosthetic feet by modeling the trajectory of the lower leg segment throughout a step for a given prosthetic foot and selecting design variables values to minimize the error between this trajectory and target physiological lower leg kinematics. This method was previously used to optimize simple analytical prosthetic foot models including (i) a pinned ankle and metatarsal joint with constant rotational stiffnesses as design variables, and (ii) a pinned ankle joint and cantilever beam forefoot, where rotational ankle stiffness and the forefoot bending stiffness where varied [6].

Thus far, all work regarding LLTE has been purely theoretical. The next step in moving towards using LLTE to

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design commercial prosthetic limbs is to clinically test the validity of LLTE as a design objective for prosthetic feet. These tests will have to ensure that the model accurately predicts the lower leg kinematics of a subject using a fully characterized prosthetic foot. Validation would imply that an LLTE-optimal foot does indeed allow a user to walk with close to able-bodied kinematics.

The goal of the present study is to create a prototype prosthetic foot that can be used for gait analysis study to test the clinical viability of LLTE. A prototype prosthetic foot must be built that is:

- Light enough that the weight of the foot does not affect the gait kinematics over the duration of the test
- Fully mechanically characterized, such that the deformation of the foot under a given load can be calculated, thereby allowing evaluation of the LLTE value for the foot
- Modular so that at least one design variable can be altered during testing in order to compare gait kinematics across a range of values of that design variable eg. ankle stiffness or forefoot bending stiffness.

Our previous prototypes were built using commercially available springs. These feet proved to be too heavy, large and did not allow spring interchangeability [7,8]. A new modular prosthetic foot that consists of a rotational ankle joint with interchangeable springs and a cantilever beam has been designed and built. The design variables of the architecture – the rotational stiffness of the ankle and the bending stiffness of the forefoot – were chosen according to the LLTE optimized foot. The considerations in building a physical prototype based on this theoretical design are discussed, and the resulting prototype is presented. Mechanical testing of the foot, showing that the intended design specifications have been satisfied, is presented. Qualitative feedback from preliminary user testing is also reported and discussed.

LLTE DESIGN OPTIMIZATION METHOD

The conceptual architecture consists of a rotational ankle joint with constant stiffness k_{ank} and a cantilever beam forefoot with a bending stiffness k_{met} (Fig. 1), as presented in previous LLTE work [5,8]. The geometry of the rotational ankle, beam forefoot foot were selected to replicate the articulation of the physiological foot-ankle complex from a set of published gait data, with h = 8 cm and d_{rigid} = 9.3 cm [9]. The rigid structure length, d_{rigid}, was chosen such that during late stance, the effective rotational joint of the pseudo-rigid-body model of the cantilever beam forefoot would be approximately at the center of rotation of the metatarsal joint for the physiological data. The pseudo-rigid-body model approximates a cantilever beam with a vertical end load as a rigid link and a rotational joint with stiffness related to the beam bending stiffness [10].

The design variables, k_{ank}, and k_{met}, were previously optimized using our LLTE-based design method [6]. The LLTE design method works by imposing physiological ground reaction forces through a step on a model prosthetic foot with given stiffness and geometry. The resulting deflection, and thus the trajectory of the lower leg (shank) are compared to physiological kinematics. The stiffness of the ankle and forefoot can then be tuned to reduce the lower leg trajectory error (LLTE) [5]. For this study, Winter's gait data for a subject of body mass 56.7 kg [9] was used as inputs into the LLTE model. The set of design variables giving the lowest value for LLET was taken to be the optimal design. The minimum LLTE value, 0.222, was calculated for k_{ank}= 3.7 Nm/deg and k_{met} = 16.0 Nm².

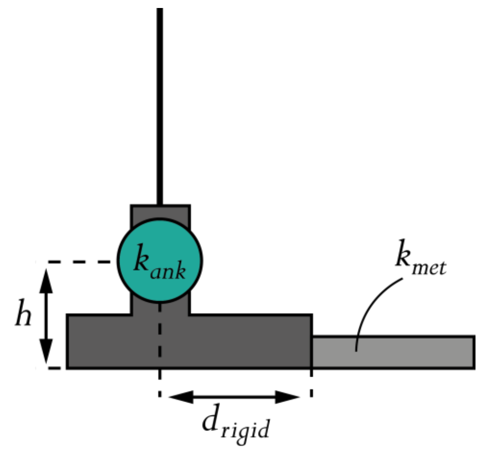

Figure 1: Foot Architecture, comprising of an ankle joint and a forefoot cantilevered beam. The position of the ankle joint and the forefoot beam have been chosen to replicate the articulation of the physiological foot-ankle complex.

MECHANICAL DESIGN OF FOOT PROTOTYPE

In order to validate LLTE as a design metric, it is necessary to design, build, and test a set of prosthetic feet based on the optimal design presented in the previous section and determine that the LLTE-optimal foot does indeed allow a user to walk with close to able-bodied kinematics. It is also important to understand the sensitivity of the design parameters on a foot's performance.

Looking at the dependence of the LLTE value on each of the design variables, Figs. 2-3 show that the LLTE value is much more sensitive to the ankle stiffness than the forefoot beam stiffness. When the foot is clinically tested, ankle stiffnesses that vary from 1.5 to 16 Nm/deg will be tested for comparison to gait kinematics across a wide range of values. The predicted LLTE values for an ankle rotational stiffness of 1.5 Nm/deg and 16 Nm/deg are 1.96 and 1.14, respectively. These LLTE values are nearly an order of magnitude different from the optimal design, therefore it is expected that they will greatly affect gait kinematics. Also, this range of rotational stiffness spans a similar range as ankle quasi-stiffness data from normal walking, which have been estimated as roughly 1.5-6.3 Nm/deg [11], 3.5–17.3 Nm/deg [12] or 3.5–24.4 Nm/deg [13] during different phases of gait.

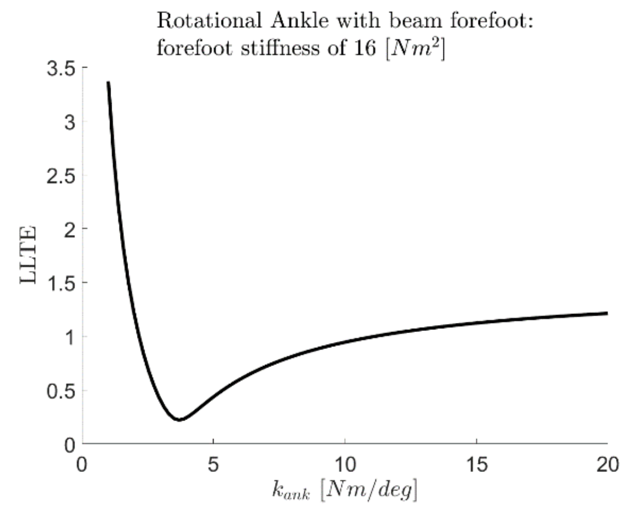

Figure 2: Dependence of the LLTE value on the ankle rotational stiffness for $\boldsymbol{k_{met}}$ = 16.0 Nm². The minimum LLTE value is achieved for $\boldsymbol{k_{ank}}$ = 3.7 Nm².

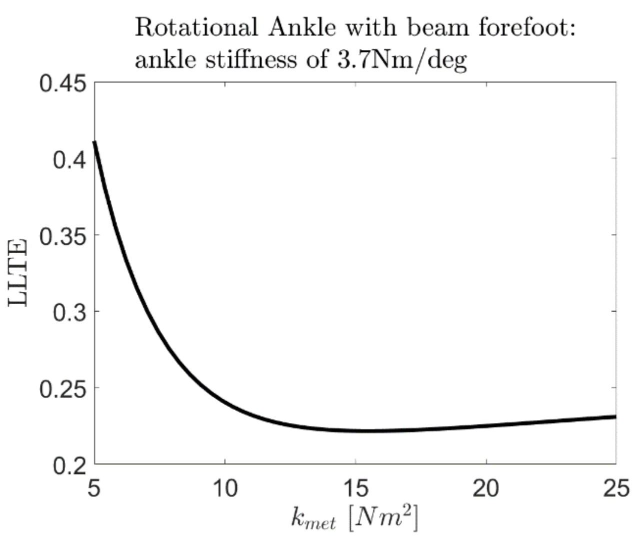

Figure 3: Dependence of the LLTE value on the forefoot beam stiffness for $\boldsymbol{k_{ank}}$ = 3.7 Nm/deg. The minimum LLTE value is achieved for $\boldsymbol{k_{met}}$ = 16.0 Nm².

The objective in this work is to design a proof-of-concept foot prototype that can accommodate our specified wide range of ankle stiffnesses with interchangeable springs. A solid model of this prototype is shown in Fig. 4. The rigid structural components were machined from acetal resin. The ankle joint rotates about a steel pin. Custom machined nylon springs fitted in aluminum mounts provide the ankle joint rotational stiffness. The compliant beam forefoot was made from nylon and was fixed to the rigid acetal resin structure with machine screws fastened directly into tapped holes in the acetal resin (Fig. 4a.). As built, the prototype has a mass of 0.980g, which is approximately 45% less than the mass of the previous prototypes [8]. This reduction in mass was achieved by replacing the metal coil springs with custom nylon flexural springs in the new architecture.

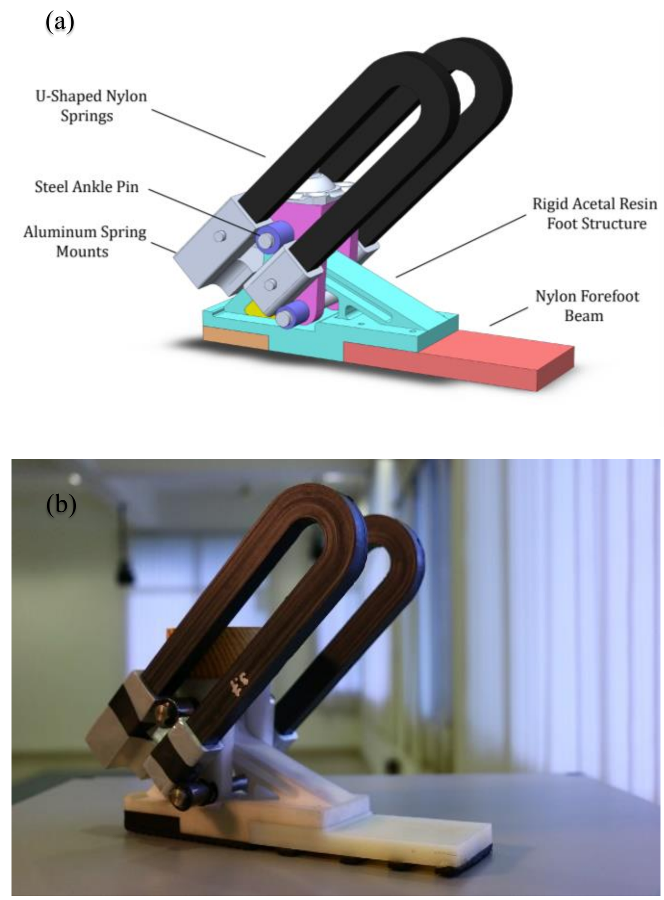

Figure 4: Solid model (a) and photograph (b) of the prosthetic foot prototype with a constant rotational stiffness at the ankle of $\boldsymbol{k_{ank}}$ = 3.7 Nm/deg and a forefoot beam stiffness of $\boldsymbol{k_{met}}$ = 16.0 Nm².

Spring Design: Requirements

To test sensitivity, a range of ankle joint rotational stiffnesses from 1.5 Nm/deg to 16 Nm/deg were selected. The springs for these range of stiffnesses had to undergo a moment of 105 Nm before yield, corresponding to the case in which a 56.7kg user applies their body weight on the tip of the prosthesis toe. Additionally, the entire mechanism needed to be as compact and lightweight so that it did not interfere with the gait, as well as modular to enable fast interchangeable springs to alter ankle joint rotational stiffness values during testing.

These requirements immediately precluded the use of commercially available coil springs, as existing coil springs of sufficient stiffness and range of motion were too heavy and bulky to allow interchangeability.

To meet these requirements, the material showing a high yield strain ($\epsilon_{yield} = \frac{\sigma_y}{E}$, where σ_y and E are the yield strength and elastic modulus of the material, respectively) combined with a high strain energy density ($u = \frac{\sigma_y^2}{E}$) was selected. Nylon 6/6 exhibited the best characteristics for a readily available, easy to machine material, with a strain energy density of 1.77 kJ/kg and a yield strain of 0.034.

Maximization of Strain Energy

The stiffness and range of motion requirements for the ankle spring exceeded the possible values for most common springs, even flexural springs, which would commonly be used for a device of this size. Therefore, it was necessary to consider how to best maximize the strain energy stored in a bending beam. The U-shape ankle spring design was thus inspired by the idea of maximizing the strain energy stored in a bending beam.

The material will yield under a stress σ_y, corresponding to a maximum moment M_y under which the beam can be loaded. In a typical cantilevered beam bending scenario (Fig. 5) the moment varies linearly from the tip to the base of the beam. The maximum moment occurs only at a single location, where the beam is constrained. Strain energy density in the beam is proportional to the elastic modulus times the square of moment in the beam (Eq. 1).

$$u \sim \frac{\sigma^2}{E} \sim \frac{(My)^2}{EI^2} \quad \textbf{(1)}$$

where y is the distance from the neutral axis.

In the case of the cantilevered beam, most of the strain energy is stored at the base of the beam. No strain energy is stored at the tip. To maximize the strain energy stored in a bending beam, it has to experience a constant maximum moment M_y across the entire length. To achieve that, a four-point beam bending scenario with rigid extremities was considered (Fig. 6). A beam loaded in this manner is able to store four times more elastic energy than a cantilevered beam of the same size.

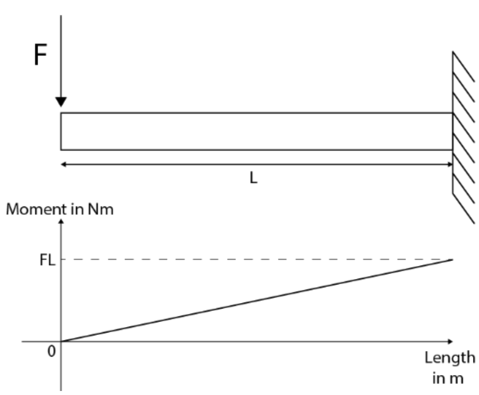

Figure 5: Schematic of a beam of length $\boldsymbol{L}$ under a load $\boldsymbol{F}$, and the corresponding moment in the beam.

Packaging and Fabrication

To package this spring in the prosthetic foot design while keeping the same characteristics, the four-point beam was packaged into a U-shape. The springs are held by aluminum mounts that act as the rigid extremities and impose a rotation on the ends of the beam. These mounts also enable interchangeability of springs (Fig. 4). Changing the overall length or the width of the beam varies the rotational stiffness of the ankle (Fig. 8).

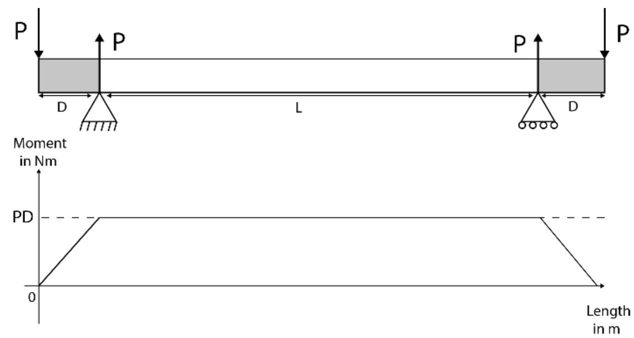

Figure 6: Schematic of a beam of length $\boldsymbol{L}$ with rigid extremities of length $\boldsymbol{D}$, under a load $\boldsymbol{P}$, and the corresponding moment in the beam.

To design these U-springs, first order calculations were performed using Euler-Bernoulli beam bending theory, with b the thickness of the beam, w the width of the beam and L its length. A relation between the rotational stiffness of the beam k_{beam}, its length, thickness, width, Young's Modulus E, yield stress σ_y was derived using Eqs. 2-4.

The maximum moment M_y under which the beam was loaded was derived from the yield stress of Nylon 6/6 with a safety factor of 1.2 (Eq. 2). Then, the maximum end slope of the beam was calculated from the moment under which the beam was loaded, the Young's Modulus of Nylon 6/6 and the beam geometry (Eq. 3). The end slope corresponded to half of the ankle angle θ_{ankle} , since in the ankle reference, one of the ends of the beam remains still. The rotational stiffness was then calculated as the moment divided by the ankle angle (Eq. 4).

$$M_y = \frac{2I\sigma_y}{b} \quad \textbf{(2)}$$

$$\theta_{max} = \frac{M_y L}{2EI} = \frac{\theta_{ankle\ max}}{2} \quad \textbf{(3)}$$

$$k_{ankle} = \frac{M}{\theta_{ankle}} = \frac{Ewb^3}{12L} \quad \textbf{(4)}$$

Using these relations (Eqs. 1-3), a first estimate of the beam geometries was calculated to achieve the desired rotational stiffness with an applied moment of 105 Nm before yield. Because the radius of curvature of the beam at the curve is on the same order of magnitude of the thickness of the beam, the U-shaped beam is stiffer than a straight beam of the same geometry. Therefore, FEA was performed to adjust the length of the U spring from the Euler-Bernoulli Solution to achieve the desired rotational stiffness (Fig. 7).

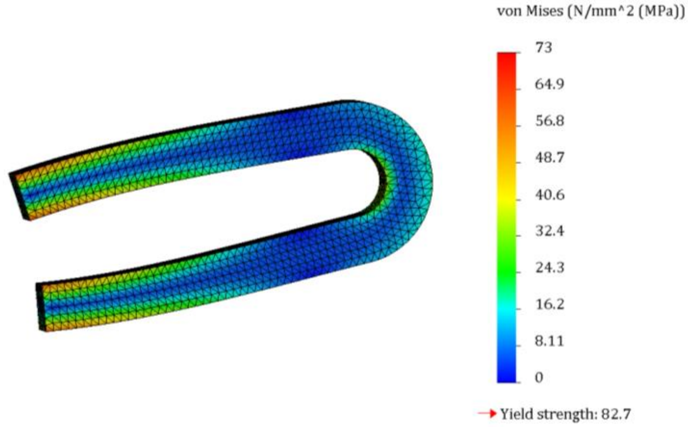

Figure 7: FEA analysis of the U shaped spring undergoing a moment of 52.5 Nm.

For U-shaped springs that yield the optimal ankle stiffness of 3.7 Nm/deg, the beams have a width of 18.24 mm, a thickness of 14 mm and a length of 160 mm. The length of the beams was varied to achieve the desired range of ankle stiffness (Fig. 8). The total mass of a pair of nylon U-shaped springs was 80g to 400g, with the optimal 3.7 Nm/deg springs weighing 225g. The springs were mounted at an angle rather than vertically to reduce the total foot volume and mass of the structure required to support them.

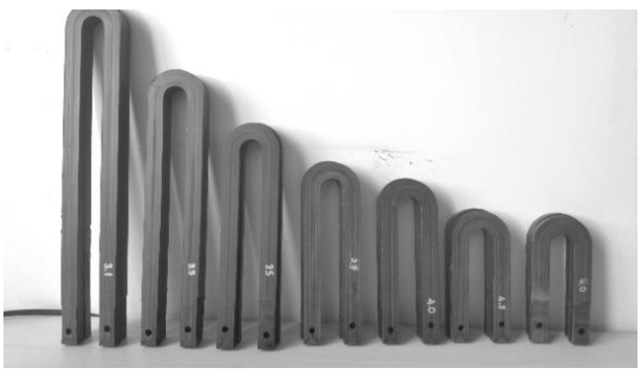

Figure 8: Set of springs with different rotational stiffness values. The longer the spring, the more compliant it is.

Cantilever Beam Forefoot Design

The geometry of the beam forefoot foot was selected to replicate the articulation of the physiological foot. Thus the width and length were respectively w_b = 0.058 m and l_b = 0.07 m, such that the total length of the foot was 21 cm. To achieve the beam bending stiffness of k_{met} = 16 Nm², several materials were considered such as acetal resin, nylon, polycarbonate, aluminum and steel. The beam thickness h_b and maximum force F_{max} that can be applied to the tip of the beam were derived from their Young Modulus E and yield stress σ_y using Eqs.5-6.

$$k_{met} = \frac{E w_b h_b^3}{12} \quad \textbf{(5)}$$

$$F_{max} = \frac{\sigma_y h_b^2 w_b}{6L} \quad \textbf{(6)}$$

From the beam thickness and maximum force values, nylon could withstand the highest load before yielding. Thus the beam forefoot was machined out of nylon with a thickness of h_b=11.1 mm.

Experimental Validation

The ankle rotational stiffnesses were measured using an Instron load testing machine. The experimental setup consisted of a jig constraining the prototype while the Instron loaded the pylon of the foot, thus applying a moment on the ankle joint (Fig. 9). The foot was loaded at a constant rate of 200 mm/s until a moment of approximately 105 Nm was applied on the ankle. The load and displacement were recorded at a rate of 15 Hz.

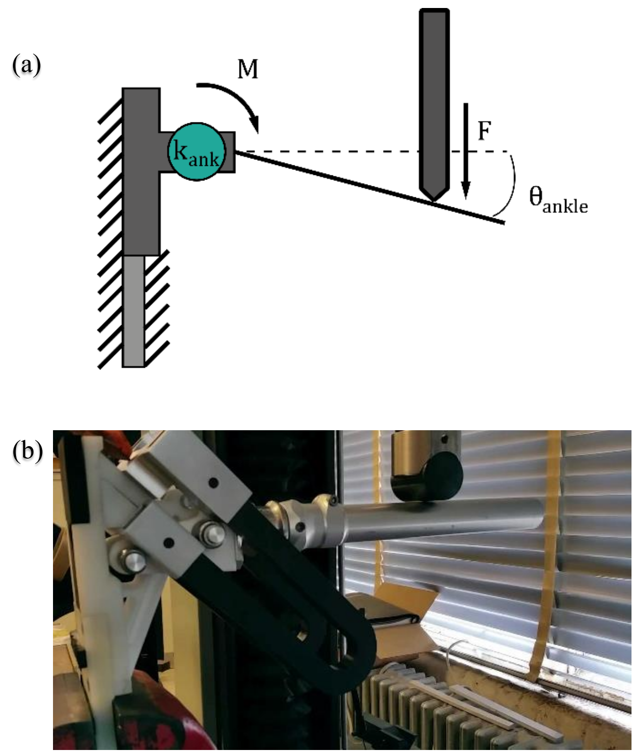

Figure 9: Experimental setup schematic (a) and photograph (b) measuring the ankle rotational stiffness $\boldsymbol{k_{ank}}$, with $\boldsymbol{F}$ the applied load on the shank from the Instron with the foot constrained in a vice, $\boldsymbol{M}$ the resulting moment on the ankle and $\boldsymbol{\theta_{ankle}}$ the measured ankle angle.

The load and displacement data were then converted using geometric relations into ankle moment and angle data. The U-shaped springs all exhibited constant linear stiffnesses ranging from 1.5 to 16 Nm/deg, as desired. The U-spring experimental data are plotted in Fig. 10 showing rotational stiffnesses of 1.5, 3.7, 5, 6 and 16 Nm/deg. The linear fits of the experimental data agree with the finite element analysis for the rotational stiffness values with a 3% error and a R^2 correlation value of 0.995.

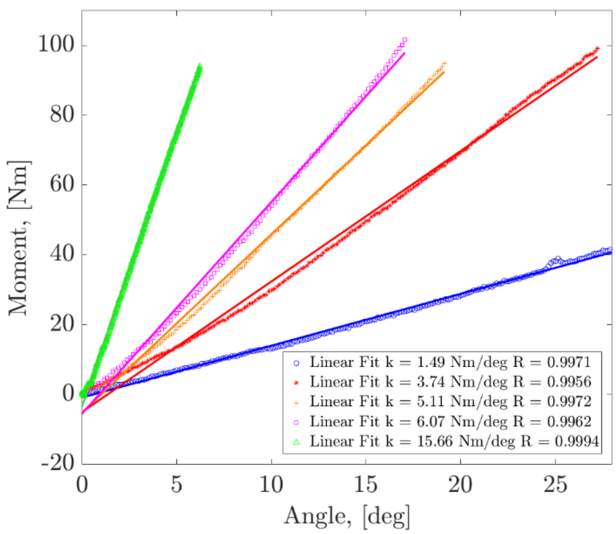

Figure 10: Experimental Data from testing the set of springs with corresponding rotational stiffness of 1.5, 3.7, 5, 6 and 16 Nm/deg. Linear fits verifying the rotational stiffness value of the springs, which agree with the FEA predicted values, are also shown.

PRELIMINARY TESTING

The prototype was tested using pseudo-prosthesis boots (Fig. 11) to ensure that both the compliant elements and the foot could withstand the typical loads experienced during flat ground walking. The prototype with different U-shaped springs was then brought for a round of testing with below-knee amputees in India (Fig. 12). Initial qualitative user testing in India to analyze comfort, functionality, spring interchangeability, reliability, and structural integrity were performed to determine the suitability of this prototype for use in a larger-scale gait analysis study. Our goal is that the technology resulting from this work will manifest in a high-performance, low-cost prosthetic foot appropriate for India and other developing countries. The prototype was fitted on three male subjects with unilateral transtibial amputations who have been long time users of the Jaipur Foot, a common Indian prosthetic foot. The subjects had body masses ranging from 55kg to 65kg. 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.

Figure 11: Pseudo-prosthesis boots mounted with the prosthetic foot prototype for preliminary testing

Figure 12: Subject with below knee amputation testing the prototype

The prototype withstood 30 min to an hour of testing on multiple subjects with multiple ankle springs with no mechanical issues, the springs could be interchanged in a matter of minutes without removing the foot from the socket. The weight of the prosthesis was not a concern for the users and no additional issues were raised during testing. The subjects then filled a survey describing qualitatively what they liked and disliked about the prototype. Subjects liked the energy storage and return of the prototype and the increased walking speed. Dislikes were mainly focused on the aesthetics of the foot. This positive feedback from preliminary testing is compelling enough to warrant further refinement of the foot design and its use for clinical studies.

DISCUSSION AND CONCLUSION

This paper presents the physical design, mechanical characterization, and preliminary user testing of a prototype prosthetic foot to evaluate the effectiveness of Lower Leg Trajectory Error (LLTE) as a design objective. The unique merits of this foot is that it enables a wide range of ankle stiffnesses to be tested over a large range of motion, similar to the quasi-stiffness and range of motion of physiological ankles. A conceptual foot architecture with a rotational ankle joint with constant stiffness U-shaped beam and a cantilever beam forefoot with a bending stiffness was considered.

The LLTE value, and thus the performance of the foot architecture, was most sensitive to the ankle rotational stiffness, which was varied in the foot design. A physical prototype design reducing the overall weight of the prosthesis while achieving a forefoot bending stiffness of 16 Nm^2 and set of interchangeable springs allowing us to change the rotational ankle stiffness from 1.5 Nm/deg to 16 Nm/deg was presented. This prototype enables testing of an LLTE-optimal foot with an optimal rotational ankle stiffness of 3.7 Nm/deg along with similar feet with higher LLTE values in order to investigate the sensitivity of ankle stiffness on walking kinematics.

Preliminary testing showed that the presented foot design reduced weight compared to previous prototypes, maintained structural integrity, and allowed fast interchange of ankle springs. The next step in this research will be to perform clinical testing and characterization of the foot and determine the viability and sensitivity of LLTE as a design objective.

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