Design and Qualitative Testing of a Prosthetic Foot With Rotational Ankle and Metatarsal Joints to Mimic Physiological Roll-Over Shape

Proceedings of the ASME 2015 International Design Engineering Technical Conferences &
Computers and Information in Engineering Conference
IDETC/CIE 2015
August 2-5, 2015, Boston, Massachusetts, USA

DETC2015-46518

DESIGN AND QUALITATIVE TESTING OF A PROSTHETIC FOOT WITH ROTATIONAL ANKLE AND METATARSAL JOINTS TO MIMIC PHYSIOLOGICAL ROLL-OVER SHAPE

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ABSTRACT

This paper presents the analysis, design, and preliminary testing of a prototype prosthetic foot for use in India. A concept consisting of a rigid structure with rotational joints at the ankle and metatarsal with rotational stiffnesses provided by springs is discussed. Because literature suggests that prosthetic feet that exhibit roll-over shapes similar to that of physiological feet allow more symmetric gait, the joint stiffnesses were optimized to obtain the best fit between the roll-over shape of the prototype and of a physiological foot. Using a set of published gait data for a 56.7 kg subject, the optimal stiffness values for roll-over shape that also permit the motion required for natural gait were found to be 9.3 N·m/deg at the ankle and 2.0 N·m/deg at the metatarsal. The resulting roll-over shape has an R^2 value of 0.81 when compared with the physiological roll-over shape. The prototype was built and tested in Jaipur, India. Preliminary qualitative feedback from testing was positive enough to warrant further development of this design concept.

INTRODUCTION

Bhagwan Mahaveer Viklang Sahayata Samiti (BMVSS), based in Jaipur, India, is one of the world's largest distributors of assitive devices [1]. In 2013, they distributed 24,000 of their prosthetic feet, the Jaipur Foot. The Jaipur Foot was designed in 1968 to meet the specific needs of persons with lower limb amputations living in India: it lasts 3-5 years in the field, can be used barefoot, allows users to squat, and costs approximately $10 USD [2]. A study comparing the Jaipur Foot to two different prosthetic feet available in the western market found that the Jaipur Foot allowed a the most natural gait [3]. However, the current foot is handmade, which is relatively costly in terms of both time and money, and causes quality to vary from foot to foot. The goal of this work is to create an upgraded replacement to the Jaipur Foot that meets the needs of the nearly one million persons with lower limb amputations living in India to replace the original Jaipur Foot [4]. Before manufacturability can be addressed, first the mechanism must be designed, tested, and iterated via proof of concept prototypes.

Most prosthetic feet used in developing countries are solid ankle cushioned heel, or SACH, feet [5]. The SACH foot consists of a rigid structure, or keel, and a cushioned heel to provide shock absorption at heel strike. While inexpensive and robust, the SACH foot does not meet the needs of persons with lower limb amputations, particularly in India. The original motivation behind the design of the Jaipur Foot was that the solid ankle of SACH-type feet does not allow squatting, a critical requirement for most people in India [2].

In the past two decades, energy storage and return, or ESAR, feet have become a popular alternative to SACH feet in the western world. The human ankle is a net power generator over the

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course of a step. At the end of stance phase, the muscles around the ankle provide a power input to aid in push-off. Powered prosthetic ankle/feet have successfully lowered the metabolic cost of walking compared to passive prosthetic feet by replicating this power input with onboard actuators, sensors, and batteries [6]. However, these robotic prostheses are expensive and would not withstand the sand, mud and water in which prosthetic feet are commonly used in developing countries. ESAR feet may be able to capture some of the benefit of providing energy input during push-off while still meeting the cost and durability constraints of developing countries.

Most, if not all, commercially available ESAR feet consist of one or more compliant beams of varying geometry. As the foot is loaded, the beams deflect, which both mimics dorsiflexion in physiological walking, and stores strain energy, which is released in late stance to assist in push-off. These types of feet do not permit squatting, which requires purely rotational motion at the ankle joint (Fig. 1). Numerous studies have compared various ESAR feet and other types of feet using mechanical testing, biomechanical gait analysis, and subjective feedback. Comprehensive reviews of this literature come to the conclusions that there is insufficient evidence to prove that any particular ESAR foot is superior to any other foot, including the SACH foot [7], and that the connection between mechanical properties of prosthetic feet on human function are not yet understood [8].

While the complete relationship between mechanical design and prosthetic functionality is unclear, one property stands out in prosthetic foot literature as a promising design objective: the roll-over shape. The roll-over shape is defined as the path of the center of pressure from heel strike to toe off in the ankle knee reference frame [9]. Roll-over shapes vary little for people of similar leg lengths. The roll-over shape has been found to be invariant to walking speed [10], added weight [11], and shoe heel height [12]. Studies suggest that prosthetic feet with roll-over shapes similar to physiological roll-over shapes result in higher symmetry in loading between prosthetic and non-prosthetic sides [9] and higher metabolic efficiency while walking [13, 14].

This paper presents the design and preliminary testing of a proof of concept prototype which provides energy storage and return, replicates a physiological roll-over shape, and allows rotational motion at the ankle joint such that it can be adapted to allow squatting more easily than compliant beam-type prosthetic feet. The analytical optimization, mechanical implementation of the prototype, and preliminary qualitative feedback are reported.

FIGURE 1: SUBJECT SQUATS WITH THE JAIPUR FOOT. SQUATTING REQUIRES PURE ROTATIONAL MOTION AT THE ANKLE JOINT THAT THE COMPLIANT BEAM-TYPE FEET AVAILABLE ON THE WESTERN MARKET DO NOT PERMIT.

Biomechanical Gait Data

Throughout this paper, the adjective "typical" is used to describe data measured from persons with no amputations or other physical impairments and under no special conditions. The set of gait data published in Winter's *Biomechanics and Motor Control of Human Movement* is used in this study as an example of typical gait kinematics and ground reaction forces. These data were collected over a single step for a subject of body mass 56.7 kg [15]. The ground reaction forces and the location of the center of pressure from the published data were considered typical loading the prosthetic foot might experience during walking. The center of pressure and joint kinematic data were used to obtain a roll-over shape for Winter's subject, referred to as the physiological roll-over shape. This physiological roll-over shape served as a basis for comparison for the roll-over shapes found for the prototype, as described in the subsequent section.

Because Winter's subject had body mass 56.7 kg, the prototype was optimized for a subject of a similar body mass. As this prototype further progresses toward a commercial product, there is potential to optimize it for various body mass ranges, as is sometimes done with prosthetic feet in the U.S. However, in order to reduce cost, the collaborators at BMVSS would prefer feet available in different lengths, but all optimized for the body mass of the average Jaipur Foot user, which is approximately 60 kg.

While it is known that persons with lower limb amputations have slower self-selected walking speeds [16–18], increased non-prosthetic side leg loading [19], and decreased gait symmetry [19–23] compared to persons with no physical impairments, typical, unimpaired gait data were used in calculating the roll-over shape of the prototype in this study. Typical loading is often assumed in measuring the roll-over shape of prosthetic feet through mechanical testing [5, 9, 24]. Also, powered prostheses designed to reproduce the ankle angle versus moment curve as measured during typical walking have been shown to lower the metabolic cost of walking relative to passive prostheses [6]. For these reasons, the authors believe that using typical, unimpaired gait data in calculating the roll-over shapes of this prototype is the best option.

PROTOTYPE CONCEPT AND OPTIMIZATION

The prototype consists of a rigid structure with pin joints at the ankle and metatarsal, with springs providing rotational stiffness at both joints. The dimensions of the prototype are based on the anatomy of the subject of Winter's published gait data, as shown in Fig. 2. The rotational stiffnesses, k_{ank} and k_{met}, are defined as

$$k_{ank} = \frac{M_{ank}}{\theta_{ank}} \tag{1}$$

and

$$k_{met} = \frac{M_{met}}{\theta_{met}} \tag{2}$$

where M_{ank} and M_{met} are the moments about the ankle and metatarsal joints respectively, and θ_{ank} and θ_{met} are the angular rotations (i.e. dorsiflexions) at each of these joints.

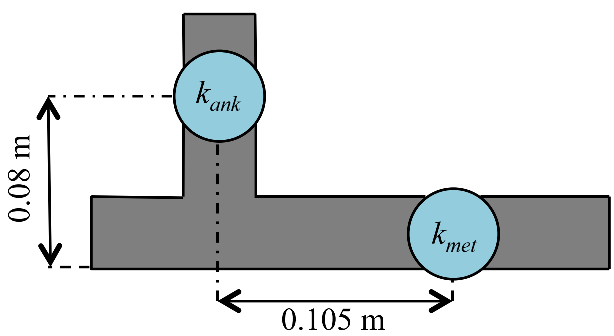

FIGURE 2: CONCEPTUAL ARCHITECTURE OF THE PROTOTYPE. DIMENSIONS ARE BASED ON THE JOINT CENTER OF ROTATIONS FOR WINTER'S PUBLISHED GAIT DATA [15]

When unloaded, the prototype is in a neutral position. Both joints allow dorsiflexion when the foot is loaded, but the geometry does not permit plantarflexion in either joint. The springs that provide the rotational stiffnesses store energy when the foot is loaded during early stance. The stored energy is then released during late stance to aid in push-off.

For a given set of stiffness values, k_{ank} and k_{met}, the roll-over shape of the prototype can be found by assuming typical loading and calculating the resulting deformation of the prototype. These values were optimized for the best fit between the roll-over shape of the prototype and of a physiological foot using the following procedure.

Rotational Stiffness Optimization for Roll-Over Shape

At a given time during a step, the horizontal and vertical components of the ground reaction force (GRF_h and GRF_v) and the horizontal position of the center of pressure relative to the ankle (CoP_x) can all be found for typical walking from published gait analysis data. This loading condition was applied to the prototype analytically (Fig. 3). Using quasistatic moment and force balances, the resulting moment about the ankle joint is given by

$$M_{ank} = CoP_x \cdot GRF_v + 0.08\,\mathrm{m} \cdot GRF_h \tag{3}$$

When the center of pressure is posterior to the metatarsal joint, the moment about the metatarsal joint is zero. When the center of pressure is anterior to the metatarsal joint, the moment about the metatarsal joint is given by

$$M_{met} = (CoP_x - 0.105\,\mathrm{m}) \cdot GRF_v \tag{4}$$

Equations (3) and (4) are valid only under quasistatic loading. This assumption is often used in analyzing prosthetic feet as the loading frequency is one to two orders of magnitude smaller than the fundamental frequency for typical prosthetic feet [5, 25, 26]. Physiological roll-over shapes measured quasistatically vary little from those measured during walking at typical speeds [9].

For specified joint stiffnesses k_{ank} and k_{met}, the deflection at both joints under the applied moments, θ_{ank} and θ_{met}, were found using Eq. (1) and Eq. (2). Together, θ_{ank} and θ_{met} define the deformed shape of the foot at time t, as shown in Fig. 4. The position of the center of pressure on the deformed foot is a single point corresponding to time t on the roll-over shape for specified values of k_{ank} and k_{met}. This was then repeated for all times from foot flat through opposite heel strike to yield the complete roll-over shape. Because the prototype does not allow plantarflexion, no motion occurs in the prototype when the center of pressure is posterior to the ankle. Consequently the roll-over shape posterior to the ankle is a straight line.

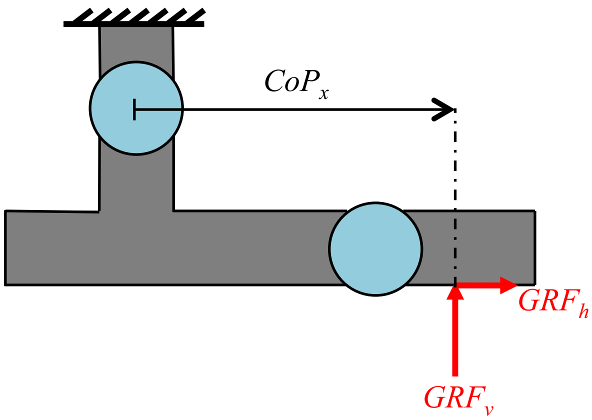

FIGURE 3: FREE BODY DIAGRAM OF LOADING APPLIED ANALYTICALLY TO PROTOTYPE CORRESPONDING TO A PARTICULAR INSTANTANEOUS TIME DURING THE STEP. THE VALUES OF THE VERTICAL GROUND REACTION FORCE (GRF_v), THE HORIZONTAL GROUND REACTION FORCE (GRF_h), AND THE HORIZONTAL POSITION OF THE CENTER OF PRESSURE (CoP_x) WERE TAKEN FROM WINTER'S PUBLISHED GAIT DATA [15].

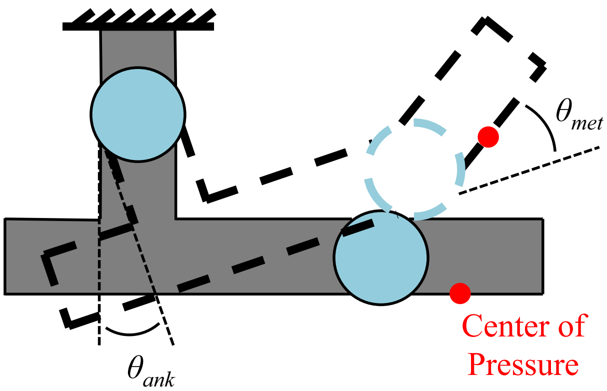

FIGURE 4: DEFORMED SHAPE OF THE FOOT CORRESPONDING TO A PARTICULAR INSTANTANEOUS TIME DURING THE STEP

Note that equations (3) and (4) provide the joint moments assuming the undeformed geometry. In the deformed state, the moment arm for the vertical ground reaction force increases while the moment arm for the horizontal ground reaction force decreases. The net resulting error on the moment calculated using undeformed geometry versus deformed geometry is very small for deformations of the size required to reproduce the physiological roll-over shape. For example, if the center of pressure in the deformed state of the model were to fall exactly on the physiological roll-over shape at the point of maximum deflection, the moment calculated using the deformed geometry differs from that calculated using the undeformed geometry by less than 0.03%. Thus calculating the joint moments using the undeformed geometry is a reasonable simplification.

Once the roll-over shape for the foot model was obtained, an R^2 value was calculated to measure the goodness of fit between the roll-over shape of the model and points interpolated along the physiological roll-over shape from Winter's gait data. The joint stiffnesses, k_{ank} and k_{met}, were varied across a range of values to find the best roll-over shape fit between the analytical models and the physiological foot. These calculations were all performed using a MATLAB (The MathWorks, Inc.) script written by the researchers.

Preliminary Optimization Results. The best roll-over shape fit was found for k_{ank} = 7.1 N·m/deg with k_{met} approaching infinity. That is, the best fit for roll-over shape occured for a rigid foot with a single rotational degree-of-freedom at the ankle joint. The resulting roll-over shape, shown in Fig. 5 had an R^2 value of 0.94 compared with the physiological roll-over shape.

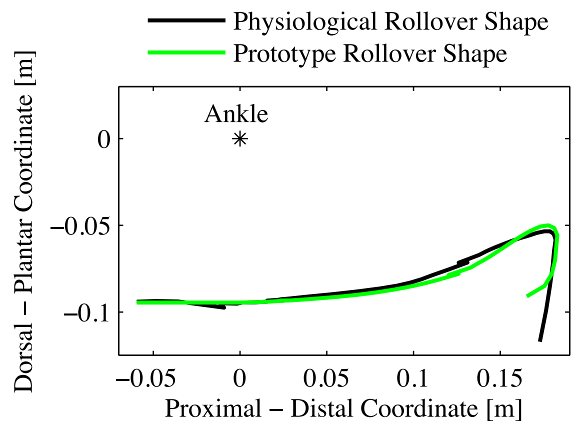

FIGURE 5: THE BEST ROLL-OVER SHAPE FIT (R^2 = 0.94) WAS FOUND FOR A PROTOTYPE WITH ANKLE STIFFNESS 7.1N·m/deg AND METATARSAL STIFFNESS APPROACHING INFINITY.

Adjustment for Metatarsal Motion. While the analysis showed that the best fit to the physiological roll-over shape came from a rigid foot with a single degree-of-freedom rotational ankle joint, such a foot would not allow the motion required for typical walking. When the heel lifts off the ground during late stance in typical walking, the foot pivots about the contact point between the ground and the metatarsal joint. When the heel lifts off the ground with a single degree-of-freedom foot as described above, the foot must pivot about the contact point between the toe and the ground (Fig. 6a). The ankle must consequently lift much higher than if the prototype had an articulated metatarsal joint (Fig. 6b).

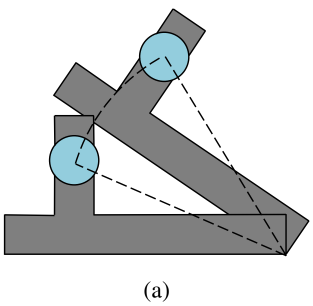

FIGURE 6: WHILE THE BEST ROLL-OVER SHAPE FIT WAS FOUND FOR A RIGID PROSTHETIC FOOT WITH A SINGLE DEGREE-OF-FREEDOM ANKLE JOINT, SUCH A FOOT DOES NOT PERMIT NATURAL MOTION (a). THE ANKLE MUST LIFT MUCH HIGHER DURING LATE STANCE THAN FOR A SIMILAR FOOT WITH AN ARTICULATED METATARSAL JOINT (b).

The rigid foot with only an ankle joint forces an unnatural walking motion. Since the roll-over shape analysis assumed typical loading, the analytical results from the simplified model would most likely not be replicated in vivo.

To allow natural motion, which is more likely to result in ground reaction forces similar to those used in the analysis and consequently validate the analytically calculated roll-over shapes, a metatarsal joint was added. In the published gait data, the metatarsal joint flexes a maximum of 30° [15]. As k_{met} decreases from infinity, the prototype permits more motion at the metatarsal, but the roll-over shape fit becomes worse. To balance between permitting natural motion and replicating the physiological roll-over shape, a metatarsal joint stiffness of k_{met} = 2.0 N·m/deg was selected such that the metatarsal joint in the analytical model reached a maximum angle of 15° under the applied loads.

With k_{met} = 2.0 N·m/deg, k_{ank} was varied to find the best fit between the roll-over shapes of the model and of the physiological foot using the method previously described. The resulting roll-over shape R^2 values for a range of ankle stiffnesses are shown in Fig. 7. The best roll-over shape fit with the prescribed k_{met} = 2.0 N·m/deg occured at k_{ank} = 9.3 N·m/deg, with R^2 = 0.81 The roll-over shape calculated for these stiffnesses is shown in Fig. 8.

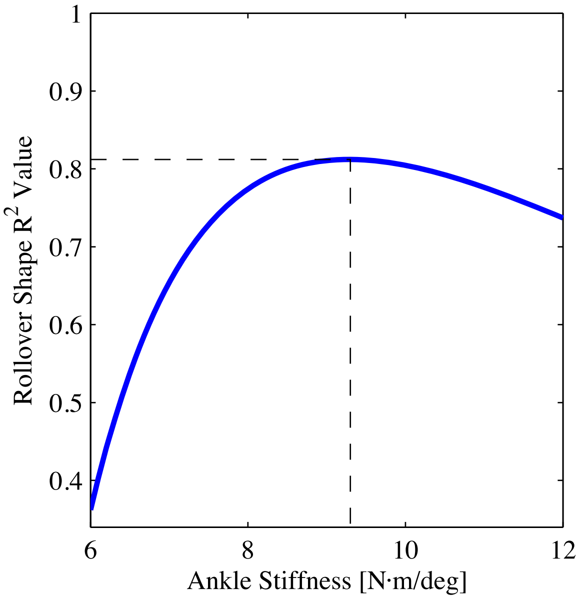

FIGURE 7: R-SQUARED VALUES COMPARING ROLL-OVER SHAPES OF PHYSIOLOGICAL FOOT TO PROTOTYPE FEET WITH METATARSAL STIFFNESS 2.0 N·m/deg FOR A RANGE OF ANKLE STIFFNESS VALUES. THE MAXIMUM R^2 VALUE IS 0.81 FOR ANKLE STIFFNESS 9.3 N·m/deg, SHOWN BY DOTTED LINES.

MECHANICAL DESIGN

Based on the analysis above, a mechanical prototype consisting of rotational ankle and metatarsal joints connected with rigid structural components was built. Because torsion springs of stiffnesses in the range of the optimal values require custom manufacturing and are prohibitively large, linear compression and extension springs were offset from the pin joints to provide the rotational joint stiffnesses. The geometry of the foot assures that the spring forces act at a constant radius from each joint, resulting in a constant rotational stiffness.

A solid model of the final design of the foot is shown in Fig. 9. Off-the-shelf springs with appropriate linear stiffnesses in the smallest form factors were selected. The remaining foot geometry was determined by these springs. Two extension springs, each with a linear stiffness of 51.8 N/cm, provided the ankle stiffness. These were positioned such that the moment arm from the spring force about the ankle was 7 cm, resulting in a nominal torsional ankle stiffness of 4.4 N·m/deg per spring, or 8.8

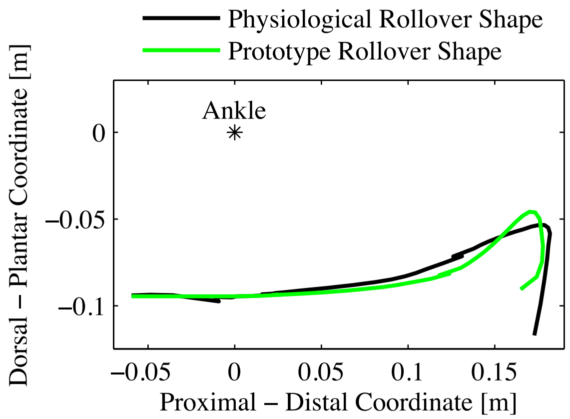

FIGURE 8: BEST ROLL-OVER SHAPE FIT ($R^2 = 0.81$) WITH ADJUSTMENT FOR NATURAL METATARSAL MOTION OCCURED FOR PROTOTYPE WITH METATARSAL STIFFNESS 2 N·m/deg AND ANKLE STIFFNESS 9.3 N·m/deg.

N·m/deg total. This is slightly lower than the optimal ankle stiffness of 9.3 N·m/deg. Constraints due to the availability of off-the-shelf springs and the overall size and weight of the prototype precluded the exact replication of the optimal value.

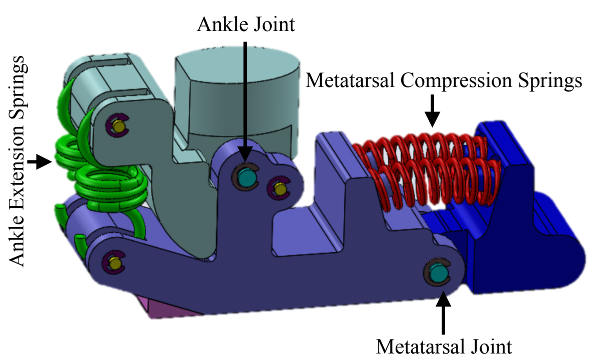

FIGURE 9: SOLID MODEL OF THE FINAL FOOT PROTOTYPE

Similarly, two compression springs of stiffness 210 N/cm provided the metatarsal joint stiffness. The springs were positioned such that the moment arm from the spring force about the metatarsal joint was 5 cm, resulting in a nominal metatarsal torsional stiffness of 0.92 N·m/deg per spring, or 1.8 N·m/deg for the joint as a whole. As with the ankle, the exact optimal metatarsal stiffness of 2.0 N·m/deg could not be achieved with off-the-shelf springs within the limits of reasonable prosthetic foot geometry.

The rigid components linking the joints were machined from delrin and sized such that deflection within these components was negligible compared to motion about the ankle and metatarsal joints, and the factor of safety for all foreseeable failure modes under expected loading was greater than two. A pin provided a mechanical stop to prevent any possible overloading at the ankle from occuring. A rubber heel wedge served to absorb some shock during heel strike. Rubber strips were epoxied along the bottom of the foot to increase traction.

The Jaipur Foot is typically attached to plastic sockets fitted at BMVSS by heating the bottom of the sockets, sliding them over the ankle block of the foot, and securing the foot in place with four radial wood screws. The ankle block of the prototype was dimensioned such that this same method could be used to attach the prototype to the sockets of the test subjects with no adaptations. The final prototype as tested is shown in Fig. 10.

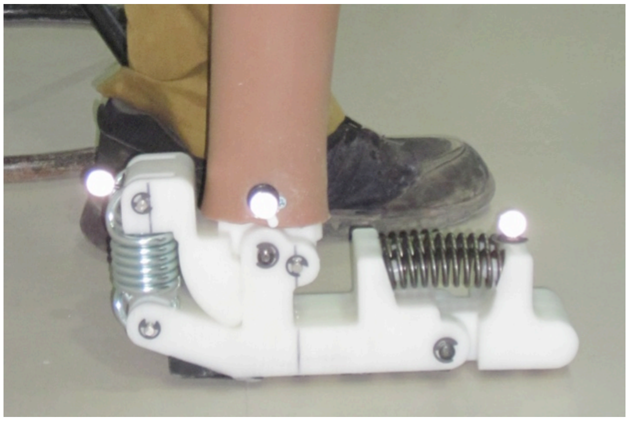

FIGURE 10: PICTURE OF THE FINAL PROTOTYPE AS TESTED

PRELIMINARY TESTING RESULTS AND DISCUSSION

The prototype was tested in accordance with an MIT Committee on the Use of Humans as Experimental Subjects (COUHES)-approved protocol to get early feedback on the viability of the design concept. Six male subjects, all experienced Jaipur Foot users with unilateral transtibial amputations and no other physical impairments, were fitted with the prototype by prosthetists at BMVSS. The subjects had body masses ranging from 45 kg to 80 kg. The subjects wore the prototype for between 30 minutes and an hour while walking around the BMVSS facility. After this time, they were asked qualitative questions about the prototype with the help of a translator.

Despite the large mass of the finished prototype, at 2 kg as compared to the 0.8 kg Jaipur Foot, the prototype was positively received. Five of the subjects liked the energy storage and return aspect of the prototype, with several of them stating that they felt like they could run or jump with the prototype. Three of the subjects commented that although the foot was noticably heavier than the Jaipur Foot, it did not feel very heavy when they were wearing it. Two of the subjects said that the weight negatively affected their movement while wearing the foot, but they would be very happy with the foot if it weighed less.

Nearly all of the subjects commented that the springs felt too stiff, or that the prototype did not provide enough dorsiflexion, with subjects of lower body mass disliking the stiffness more than the subjects of higher body mass. Based on observations, the subjects almost always favored their sound limb more with the prototype than they did with the Jaipur Foot, adopting a slight limp to keep the majority of their weight on their sound limb. This means that the loads on the prosthetic foot were less than the typical loading that was assumed in the analysis and stiffness optimization, which would result in less than the intended dorsiflexion. Two possible explanations for this are that 1) the subjects were not given adequate time to get comfortable wearing the prototype foot, and 2) the fact that the foot looks like an experimental prototype rather than a commercial product could have made the subjects wary of the durability of the prototype. One subject verbalized this latter sentiment. If subjects were given more time to acclimate to the prototype, they would likely become more comfortable loading the prototype with their full body weight. The loads would then approach typical loading, and the amount of dorsiflexion would increase. Further testing for longer durations is required to determine whether the spring stiffnesses are indeed too high.

The prototype as built did not allow squatting, which requires a lower ankle rotational stiffness than is optimal for walking. During squatting, the moment at the ankle produced by a 60 kg user can be up to a maximum value of approximately 53 N·m if the weight is distributed equally between his or her legs and the center of pressure is under the toes of the foot. With the prototype's nominal ankle stiffness of 8.8 N·m/deg, this load would result in 6° of dorsiflexion. Between 15° and 30° are required for squatting. However, because the ankle joint permits purely rotational motion, the prototype can be adapted to allow squatting more easily than compliant beam-type feet. For example, the ankle joint stiffness can be optimized for walking with a mechanism that disengages the spring to allow free motion during squatting, a mechanism can be designed that has a bi-modal stiffness, where after the ankle reaches a certain angle, the stiffness drops significantly, or the ankle stiffness can be reduced to find a compromise that may not be the optimal value for either squatting or walking, but permits both. Once the analysis is validated and the prototype is optimized for natural walking, the prototype will be adapted to permit squatting while maintaining the best possible performance for walking.

While qualitative feedback from six subjects is insufficient to conclusively compare this prototype to the Jaipur Foot or any other prosthetic foot on the market, the positive responses suggest that this design concept merits further development and testing. As the design progresses, more rigorous testing, including longer duration, quantitative gait analysis, and activities beyond level-ground walking will be used to further refine the design of the foot.

CONCLUSION

The goal of this work is to design a prototype prosthetic foot that meets the needs of persons with lower limb amputations living in India. A prototype that consisted of a rigid structure with rotational joints at the ankle and metatarsal was designed. The rotational stiffnesses at each of these joints were optimized such that the prototype roll-over shape, calculated analytically using typical loading from published gait data, best fit the physiological roll-over shape from that same published gait data.

The best roll-over shape fit was found for a prototype with an ankle stiffness of 7.1 N·m/deg and metatarsal stiffness approaching infinity, with $R^2 = 0.94$. This corresponds to a rigid foot with a single degree-of-freedom rotational ankle joint. However, such a foot does not permit a natural walking motion, as the ankle must be lifted higher during late stance than for a similar foot with an articulated metatarsal joint. The ankle stiffness optimization process was repeated with the metatarsal stiffness set to 2.0 N·m/deg, such that the metatarsal joint reached a maximum angle of 15° under the maximum load. With this constraint, the best roll-over shape fit was found for ankle stiffness 9.3 N·m/deg, with $R^2 = 0.81$.

A prototype was built using pin joints to produce the ankle and metatarsal joint motion and off-the-shelf linear compression and extension springs to provide the joint stiffnesses. The final prototype as built had nominal ankle stiffness 8.8 N·m/deg and metatarsal stiffness 1.8 N·m/deg. The availability of off-the-shelf springs and geometric constraints limited how closely the prototype joint stiffnesses could match the optimal values.

The prototype was tested in India on six male subjects with unilateral transtibial amputations. Despite weighing more than twice as much as the original Jaipur Foot, the prototype received mostly positive feedback. Several subjects commented that the springs were too stiff. Further testing for longer durations and with qualitative gait analysis is required to determine whether this comment is a consequence of the subjects having insufficient time using the prototype to become comfortable with it.

The generally positive response to the foot is sufficient to warrant further refinement of this prototype. Future work will focus on 1) obtaining quantitative biomechanical gait data with the current prototype to validate the analysis, 2) improving the optimization method, with a particular focus on developing a quantitative design objective that accounts for whether the prototype allows natural motion, and 3) designing a new mechanism that can achieve the same type of motion as this prototype, but fits in a smaller, lighter package and is mass-manufacturable. Once the mechanism is satisfactory, the final step will be to incorporate a foam cosmesis that makes the prosthetic foot look like a biological foot while simultaneously protecting the internal mechanism from the environment.

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 in participating in design reviews and testing prototypes in India.

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