Lower Leg Trajectory Error: A Novel Optimization Parameter for Designing Passive Prosthetic Feet

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***Abstract*—Roll-over geometry, a leading design objective for passive prosthetic feet, does not provide enough information to find the lower leg segment orientation in the laboratory reference frame. The physical behavior of a prosthetic foot adds a constraint which makes finding this orientation possible. A novel optimization parameter for prosthetic feet that incorporates both the roll-over geometry and the orientation of the lower leg, the Lower Leg Trajectory Error, is introduced. Ground reaction forces and locations of the center of pressure from published gait data for physiological, able-bodied walking are applied to a biologically inspired conceptual prosthetic foot to calculate the resulting lower leg trajectory from foot flat to toe off. The stiffnesses of this conceptual foot are optimized using the Lower Leg Trajectory Error. To further investigate the role of roll-over geometry and physical behavior, the lower leg trajectories of the optimized foot, a rigid foot with roll-over geometry identical to this optimized foot, and a rigid foot with physiological roll-over geometry are compared.**

I. Introduction

There is substantial evidence to suggest that the mechanical function of passive below-knee prostheses affects walking mechanics and efficiency of users [1]–[9]. However, how the mechanical features of a passive prosthesis affects the functionality is not fully understood [10]. Without this knowledge, passive prosthetic feet cannot be optimized for peak performance.

Prosthetic foot design for both passive and active mechanisms has focused on replicating the functionality of the physiological foot-ankle complex to maximize functional mobility for the user [1], [11]–[13]. However, the physiological foot-ankle complex is a complicated system capable of feedback, active control, and power generation. A passive prosthesis cannot reproduce all of these functions. One simplified parameter that a passive prosthesis can replicate is the physiological roll-over geometry. The roll-over geometry is defined as the path of the center of pressure along the foot from heel strike to toe off in the ankle-knee reference frame [14]. Studies suggest that prosthetic feet that replicate roll-over geometry result in increased metabolic efficiency, more symmetric gait, and higher subjective preference [3], [4], [14]–[16]. These studies focused on either commercially available prosthetic feet or on prototypes that could be altered to vary the roll-over radius or arc length.

This paper demonstrates that the orientation of the lower leg, an important parameter for both gait kinematics and joint reaction forces and moments, is not included in the roll-over geometry, but is constrained by the physical geometry of a prosthetic foot. A novel optimization parameter that incorporates both the roll-over geometry and the lower leg orientation, the Lower Leg Trajectory Error, is introduced. This parameter is used to optimize the stiffness for a biologically-inspired conceptual model foot.

II. Roll-Over Geometry Omits Orientation

As previously stated, the goal of a passive prosthesis is to replicate biological limb functionality with a relatively simple mechanical structure. For a passive mechanical prosthesis, a given loading scenario will produce a specific deformation. The relationship between the loading and the deformation, or the stiffness, can be non-linear and/or vary in different parts of the structure, but even in these complex situations, the deformation resulting from a specific load can always be calculated.

The roll-over geometry of a prosthetic foot is a spatial measure of stiffness. When the center of pressure is at a certain position along the foot, the roll-over geometry shows what the vertical deflection of that point will be. The roll-over geometry also serves to simplify the many variables that can be measured during a biological step into a single curve that can be used as a design objective.

While the roll-over geometry compiles a lot of information into a single curve, it does not provide any information regarding the orientation of the lower leg segment in the laboratory reference frame (Fig. 1). When the center of pressure is located at a particular point along the roll-over geometry, that single point does not constrain the angular orientation of the foot-ankle-knee complex. More information is needed about the physical construction of the foot and how it interacts with the ground to fully define the position of the system.

A person with a transtibial amputation interfaces with the prosthesis throught the socket. Ideally there is no relative motion between the user and the socket, so the position and orientation of the socket dictates the position and orientation of the user's residual limb. Both the socket and the residual limb make up the lower leg. The socket also transmits forces and moments to the user (Fig. 2). The orientation of the

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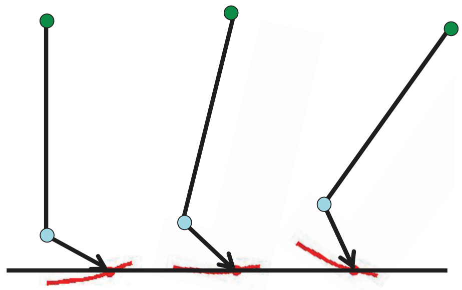


Fig. 1: For a below-knee prosthesis (shown here with the green dot representing the user's knee, blue dot representing the ankle) for which all that is known is the roll-over geometry (red curve), when the center of pressure is at a particular location (red dot), the orientation of the lower leg segment is indeterminable.

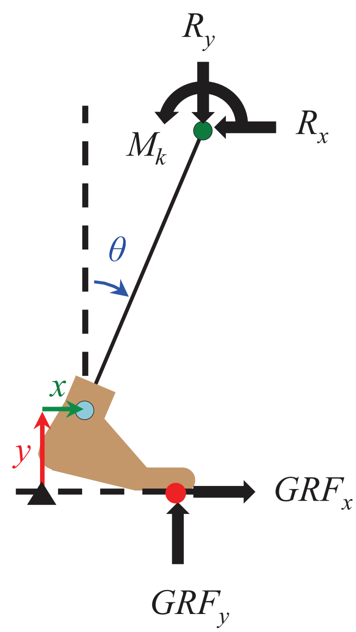


Fig. 2: Free-body diagram of foot-ankle-knee system in the saggital plane. The system is acted on by the ground reaction forces (GRF_x and GRF_y) and the reaction loads (R_x and R_y) and moment (M_k) at the knee. The position and orientation of the lower leg segment is fully defined by three variables: the horizontal and vertical position of the ankle (x and y respectively) and the angle of the lower leg with respect to vertical (θ). The orientation of the lower limb affects not only the gait kinematics of the user, but also the reaction moments on his or her residual limb and at her knee.

socket defines the moment arm from the ground reaction forces to the user's residual limb and knee. Therefore any variation in the orientation of the lower leg affects both the gait kinematics and the loading at the user's knee.

The physical geometry of a prosthetic foot introduces an additional constraint that prescribes the orientation of the lower limb. For a particular prosthetic foot, the geometry can be included to optimize the design not only for roll-over geometry, but also for the orientation of the roll-over geometry in the laboratory reference frame, and thus the trajectory of the lower leg.

III. Lower Leg Trajectory Error

To define the position of the lower leg in 2D space, three parameters, x, y, and θ here, are needed (Fig. 2). These were then compared to the target values taken from published physiological gait data, $\hat{x}$, $\hat{y}$, and $\hat{\theta}$. As the lower leg moves throughout a step, each of these variables are functions of time. The net error between the lower leg trajectories of a prosthetic foot model and of the physiological step was taken to be the root-mean-square error of each of these variables normalized to the physiological range of values, or

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

,

where the subscript n refers to the n^{th} time interval (that is, the n^{th} point along the roll-over geometry), and N is the total number of time intervals considered. The subscripts max and min refer to the maximum and minimum values of the parameter over the course of the step, or over the portion of the step included in the optimization. Lower values of $LLTE$ represent a better fit with the able-bodied ankle-knee trajectory; a model that fit the data exactly would result in an $LLTE$ value of zero.

IV. Design Optimization Using Lower Leg Trajectory Error

A. Model Foot Architecture

To demonstrate the usefulness of the Lower Leg Trajectory Error as a design optimization parameter, an example model foot was optimized for lower leg trajectory. Because the goal of this work is to create prostheses that replicate physiological kinematics as closely as possible, a model prosthetic foot that has rotational joints at the ankle and metatarsal, which replicate the articulated joints of biological feet, was considered (Fig. 3). These joints have rotational stiffnesses k_{ank} and k_{met} respectively. The links connecting these joints are perfectly rigid.

B. Lower Leg Trajectory Calculation and Optimization

To find the predicted lower leg trajectory for this analytical model, the horizontal and vertical components of the ground reaction forces and the position of the center of pressure along the ground were used as inputs. For the

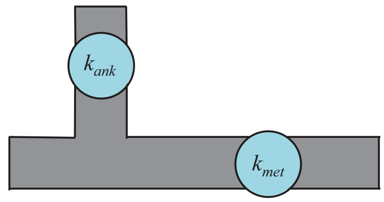


Fig. 3: Analytical model of prosthetic foot consisting of rotating joints at the ankle and metatarsal with constant rotational stiffnesses, k_{ank} and k_{met}, respectively.

initial optimization, these inputs were taken from published gait data for typical, able-bodied walking [17].

At each instant during a step, the ground reaction forces were applied to the analytical foot model at the center of pressure, which was found on the foot by taking the distance the center of pressure had traveled along the floor from heel strike to the instantaneous position and assuming no slip between the ground and the bottom of the foot. The ground reaction forces produced moments at the metatarsal and ankle joints. For specified rotational stiffnesses, the instantaneous deformed shape of the foot (Fig. 4a) can be found. This calculation was performed assuming that the foot can be modelled quasistatically, which is a common assumption in analyzing feet, as the natural frequencies of most prosthetic feet are one to two orders of magnitude larger than the loading rate during walking [18]–[20]. The position of the center of pressure in the deformed shape provides a single point on the roll-over geometry for a foot with specified joint stiffnesses.

For the jointed foot architecture in Fig. 3, when the center of pressure is in a given location, the ground must be in contact with the segment of the foot containing the center of pressure (Fig. 4b). This fact, together with the deformed shape of the foot, dictate how the ground must be oriented relative to the foot in the ankle-knee reference frame, and, consequently, the position and orientation of the lower leg in the laboratory reference frame (Fig. 4c).

To obtain the full lower leg trajectory for a particular set of stiffnesses, the position and orientation were found for each time interval from foot flat to toe off. This calculation was repeated over a range of joint stiffness values. The lower leg trajectory error was found for each set of stiffnesses to find optimal values.

C. Optimization Results

The Lower Leg Trajectory Errors for the range of stiffnesses considered are shown in Fig. 5. The optimal joint stiffnesses are those with the lowest value of $LLTE$, or in this case, 6.7 N·m/deg for the ankle joint and 1.8 N·m/deg for the metatarsal joint, with $LLTE$ = 0.17.

The Lower Leg Trajectory Error can be used to determine how closely different conceptual foot models replicate the

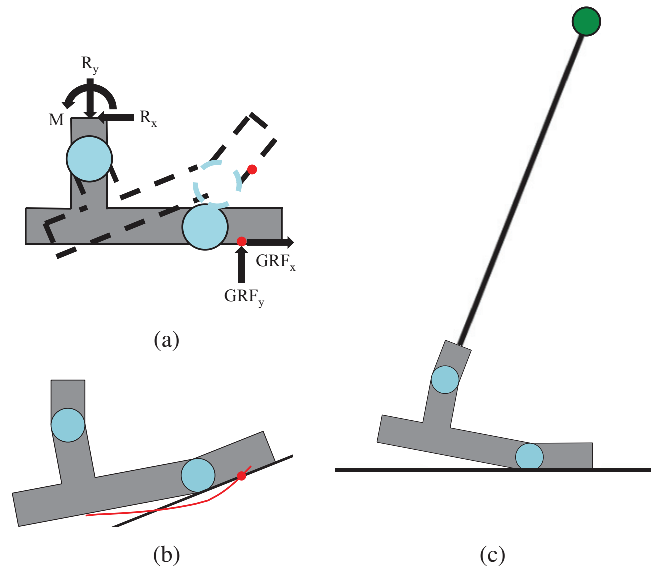


Fig. 4: To find the lower leg position and orientation for the jointed model foot, the instantaneous deformed shape of the foot for given joint stiffness values is found by applying ground reaction forces at the center of pressure (a). In the ankle-knee reference frame, the physical geometry of the foot constrains the ground to be in contact with the segment containing the center of pressure (b). This relative orientation of the ground and the deformed foot defines the orientation of the lower leg in the laboratory reference frame (c).

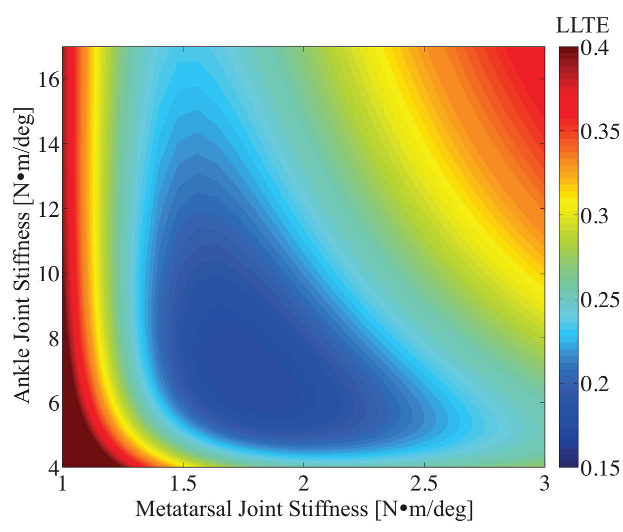
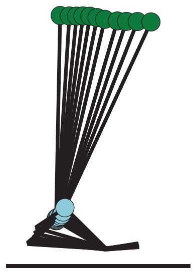
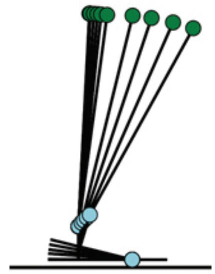


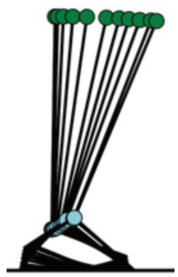
Fig. 5: $LLTE$ values calculated for the jointed foot model in Fig. 3 over a range of metatarsal and ankle joint stiffnesses. The optimal stiffness values are those which produce the minimum $LLTE$. In this case the minimum $LLTE$ value is 0.17 for ankle stiffness of 6.7 N·m/deg and metatarsal stiffness of 1.8 N·m/deg.



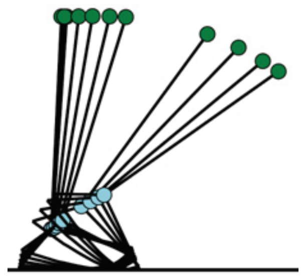
(a) Physiological Foot



(b) Jointed Foot



(c) Rigid Foot with Physiological Roll-Over Geometry



(d) Rigid Foot with Jointed Foot Roll-Over Geometry

Fig. 6: Comparison of physiological lower leg trajectory from able-bodied gait data (a) with the optimized jointed foot model ($LLTE$ = 0.17) (b), a rigid foot with physiological roll-over geometry ($LLTE$ = 1.32) (c) and a rigid foot with roll-over geometry exactly matching the roll-over geometry of the jointed foot ($LLTE$ = 2.09 (d).

physiological lower leg trajectory. For instance, consider a rigid foot, such as one cut from a solid piece of wood. Because no deformation occurs within the foot, the shape of the bottom of the foot determines the roll-over geometry, which makes this foot model a useful tool to investigate different roll-over geometries [15], [16]. The rigid nature of the foot constrains the ground to be tangent to the bottom of the foot at the instantaneous center of pressure. Using this constraint, the lower leg trajectory can be calculated for rigid feet of different roll-over geometries.

Comparing the jointed foot model optimized for lower leg trajectory to a rigid foot with an identical roll-over geometry, as well as a rigid foot with physiological roll-over geometry, demonstrates the importance of both the roll-over geometry and the physical behavior of feet for the resulting lower leg trajectory. The $LLTE$ value quantifies what is shown graphically in Fig. 6: the jointed foot model most closely replicates the physiological lower leg trajectory ($LLTE$ = 0.17), followed by the rigid foot with physiological roll-over geometry ($LLTE$ = 1.32). The rigid foot with roll-over geometry identical to the jointed foot does not replicate the physiological foot lower leg trajectory well, with $LLTE$ = 2.09. The fact that two feet can have such different lower leg trajectories with identical roll-over geometries verifies that roll-over geometry alone is insufficient to design or evaluate prosthetic feet.

V. Discussion

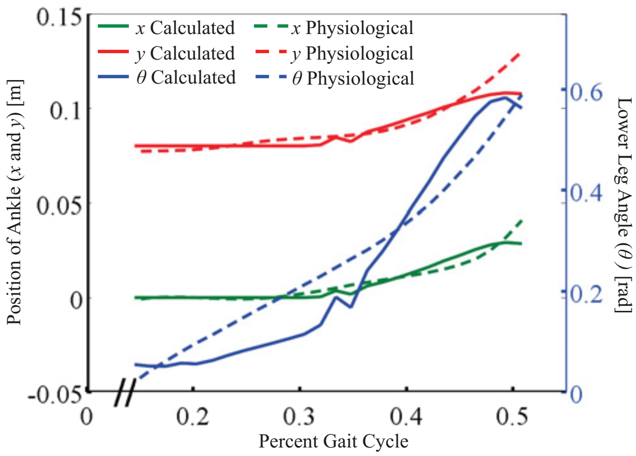
Examining the variables defining the lower leg trajectory (x, y, and θ) individually provides a deeper understanding of the $LLTE$ for the optimized jointed foot and the rigid foot with physiological roll-over geometry (Fig. 7). Note that the calculations for the rigid foot do not extend as far as the jointed foot. This is because the physiological roll-over geometry changes curvature before toe off. If the rigid foot were shaped such that it also contained this change in curvature, there would be a portion of the bottom of the foot which could never contact the ground, which would impede the replication of the physiological roll-over geometry. The calculation of $LLTE$ for the rigid shape thus excludes the last 5% of stance phase.

Figures 6 and 7 show that the $LLTE$ for the rigid foot with physiological roll-over geometry is primarily from discrepencies in the horizontal progression of the ankle, or x. Because the foot is rigid, as the center of pressure progresses along the bottom, the ankle joint both rises and moves forward. In contrast, the ankle in the jointed foot model remains stationary until the center of pressure progresses beyond the metatarsal joint, at which point the heel lifts off the ground and the ankle begins to move. The translation of the ankle for the jointed model matches the physiological ankle translation much more closely than for the rigid foot.

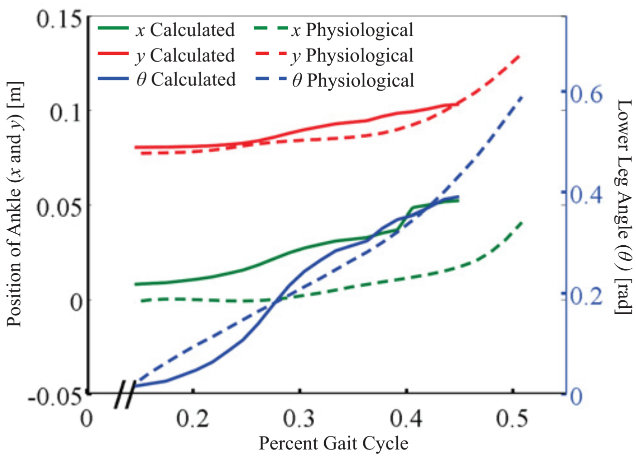
It is also clear from both the spacing in Fig. 6 and from the slope of the θ curves in Fig 7 that the knee progresses at different rates for the jointed foot and the rigid foot with physiological roll-over geometry. During foot flat, the knee progresses very slowly for the jointed model, as the moment arm from the ground reaction force about the ankle is small, which means that the ankle joint does not rotate much. Later in stance, as the moment becomes larger, both joints deflect more and the knee progresses rapidly. The knee in the rigid foot also progresses slowly both in the beginning of foot flat and just before toe off, but advances quickly in between, as the foot rocks forward with no resistance.

The discrepancies between the lower leg trajectory for the physiological system and the rigid foot with the physiological roll-over geometry demonstrate an important implication of the underdefined nature of the roll-over geometry: though the roll-over geometry is calculated from a single set of physiological, able-bodied gait data, when the loading from that same set of gait data is applied to the roll-over geometry, the lower leg trajectory output does not match the physiological trajectory. This again demonstrates that roll-over geometry alone is insufficient as a design objective, as matching the roll-over geometry does not guarantee that the lower leg kinematics will similarly match target gait kinematics.

The lower leg trajectory error incorporates the roll-over geometry, which is a function of the lower leg position and orientation (x, y and θ) and the position of the center of pressure, which was used as an input. However, the trajectory error encompasses more information than the roll-over geometry alone, namely the orientation of the lower leg. Consequently, the jointed foot model optimized here for lower leg trajectory has a worse fit with the physiological



(a) Jointed Foot



(b) Rigid Foot with Physiological Roll-Over Geometry

Fig. 7: A comparison of x, y, and θ for the jointed foot optimized for lower leg trajectory (a) and a rigid foot with physiological roll-over geometry (b). Values of these parameters for the rigid foot could not be calculated for the end of stance phase because the physiological roll-over geometry changes curvature, which cannot be replicated with a rigid foot without impeding the roll-over geometry earlier in stance.

roll-over geometry than it would if the joint stiffnesses had instead been optimized solely for roll-over geometry (Fig. 8). The R^2 value comparing the roll-over geometry of the model optimized for lower leg trajectory error is 0.53.

Because the $LLTE$ compares modeled values to physical values at each time interval during a step, the $LLTE$ includes a temporal optimization element not present in the roll-over geometry. Most roll-over geometry investigations focus only on the shape itself or certain attributes of the shape, such as radius [3], [4], [15] or arc length [16], [21]. While it is possible to include temporal effects in roll-over geometry by evaluating the rate of progression of the center of pressure, it is not typically done.

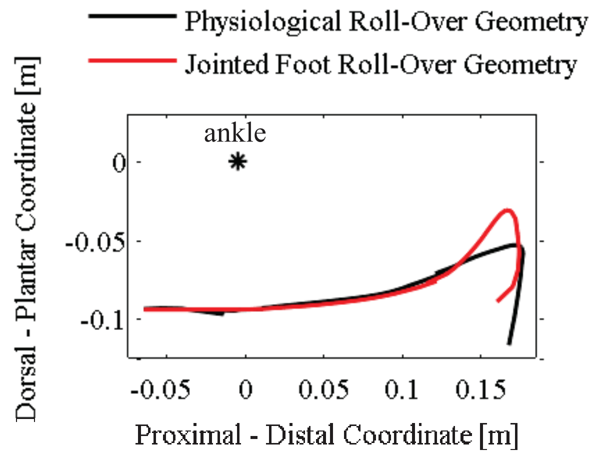


Fig. 8: A comparison of the roll-over geometry of the conceptual jointed foot model optimized for lower leg trajectory and the physiological roll-over geometry (R^2 = 0.52).

This analysis was performed using inputs from published able-bodied gait data. As previously mentioned, there are differences between the gait of persons with lower limb amputations and able-bodied persons. Additionally, the design of a particular prosthetic foot affects how a user walks. When a prosthetic foot is optimized for able-bodied gait data and then built, there will certainly be differences between the loads actually applied to the prosthesis and the able-bodied loads for which the foot was designed, which will consequently lead to a response of the foot different from that predicted in the model. This problem may be ameliorated by performing iterations of this analysis. Once an initial prototype is built based on able-bodied optimization, testing can be done to measure the ground reaction forces and center of pressure progression by a group of users. These can then be used as inputs while still targeting able-bodied outputs to refine the design. The hope is that gradually the input ground reaction forces and center of pressure progression used in optimization converges to that measured when the prosthesis is worn by human users.

As initially defined, the error in each of the three variables comprising the lower leg trajectory, x, y and θ, are weighted equally in the definition of the optimization parameter $LLTE$. As the analysis is purely theoretical, there is no reason to suspect that any one of these is more important than the others. In moving forward, testing should be done to determine whether this is truly the case when a human user is involved.

The proposed optimization parameter, $LLTE$, only addresses mid-stance kinematics, from foot flat to toe off. The heel strike to foot flat phase of stance can be investigated independently. Many commercially available prosthetic feet already decouple the response of the foot during these two separate phases by using one cantilever beam extending forward from the ankle and either a second cantilever beam extending backward or a foam cushion at the heel. The purpose of the heel portion of these prosthetic feet is primarily to provide shock absorption.

VI. Conclusion

The analysis presented here shows that the roll-over geometry, a design objective currently used in passive prosthetic feet, omits the orientation of the lower leg, an imporant parameter for both gait kinematics and joint reaction forces and moments. Consequently, it is possible for a prosthetic foot to exactly mimic the physiological roll-over shape, but greatly differ from physiological lower leg orientation. A novel design optimization parameter that incorporates both lower leg orientation and roll-over geometry, the Lower Leg Trajectory Error, was introduced. The stiffness of a biologically-inspired conceptual prosthetic foot was optimized using the $LLTE$ value.

The Lower Leg Trajectory Error is a new metric that can be used to evaluate how closely a prosthesis mimics biological functionality. It also provides a basis to assess prosthetic feet analytically in early stage design. While further testing is required to validate the clinical effectiveness of the Lower Leg Trajectory Error, incorporating more information into the design of passive prostheses will facilitate improved replication of physiological gait.

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