Compliant Mechanism Optimization of a Passive Prosthetic Foot Using Lower Leg Trajectory

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Introduction

This work presents the application of compliant mechanism optimization to design a light weight, low cost, single part passive prosthetic foot that best replicates physiological lower leg kinematics under typical ground reaction forces. Despite plenty of literature comparing various mechanical characteristics of passive prosthetic feet, the relationship between their mechanical properties and biomechanical functionality is not fully understood [1]. One widely used metric is the roll-over geometry, which is defined as the path of the center of pressure during stance phase as measured in the ankle-knee reference frame [2]. Roll-over geometry offers advantages over other metrics in that it can be evaluated for typical physiological walking, providing a target design shape, as well as mechanically for prosthetic feet without the inherent variability of human subjects. However, our previous work has demonstrated that, because roll-over geometry is measured in the ankle-knee reference frame without including any information regarding the orientation of the ankle-knee reference frame relative to the global reference frame, it is possible for two different prosthetic feet to have identical roll-over geometries but exhibit very different lower leg kinematics during gait [3]. Therefore, roll-over geometry is insufficient as a design objective.

We have developed a novel approach to designing and optimizing prosthetic feet by aiming to replicate the trajectory of the lower leg segment during physiological walking under typical ground reaction forces (GRFs). This approach is implemented by calculating the deformed shape of a given prosthetic foot under the GRFs at each instant during a step, using those deformed shapes to find the position of the lower leg segment at each time, then comparing those positions to the target physiological data set using a root-mean-square error over the course of the step, a metric that we have termed the Lower Leg Trajectory Error (LLTE). The optimal design is then the design that results in the minimum LLTE, that is, the design that best replicates the physiological lower leg kinematics under the corresponding kinetics.

This approach has previously been used to optimize the design of three simple prosthetic foot architectures, each with two design variables: a rigid circular foot with the radius and horizontal position of the center of the circle as design variables, a foot with pin joints at the ankle and metatarsal with rotational stiffness of each joint as design variables, and a foot with a pin joint at the ankle and a compliant cantilever beam forefoot, with ankle stiffness and forefoot beam bending stiffness as design variables. Prototypes have been built based on these models and clinical testing is currently underway to validate this work, with promising preliminary results.

While these architectures were quick to optimize, as the deformation in response to loads could be calculated analytically, the resulting prototypes are heavy, at 980 g after multiple design iterations intended to reduce weight, and complicated to manufacture, requiring pinned joints, springs, multiple fasteners, and bulky structural components.

The goal of this work was to apply a compliant mechanism optimization method to design a single-part prosthetic foot that replicates physiological lower leg trajectories but is lighter weight and easier to fabricate than the previously considered simple architectures with two design variables.

Methods

The shape and size of the prosthetic foot keel was defined as a parametric wide Bezier curve [4], characterized by a set of nine design variables. These design variables described the height of the prosthetic foot, the shape of the curve connecting the ankle to the bottom surface of the foot, and the width of the shape along the length of the curve. The thickness of the foot into the plane was fixed at 6.0 cm. The analysis was performed using nylon 6/6 as the material, as one goal of this work is to produce a

low cost, mass-manufacturable foot suitable for use in developing countries.

To evaluate the lower leg trajectory for a given design, finite element analysis (FEA) was performed on the design three times, each time with a different load scenario representing a different part of stance phase. The GRF and center of pressure data were obtained from a set of published gait data. The deformed shape of the foot mechanism in each scenario was found from the FEA results. Based on this deformation, the relative angle between the foot and the ground was calculated, then used to determine the angle of the lower leg segment relative to vertical in the global reference frame. Assuming that no slipping occurs between the foot and the ground, the horizontal and vertical positions of the knee were computed. These three variables were then compared with the kinematic data from the same published data set to obtain the LLTE for that particular set of design variables.

The optimization was performed using MATLAB's built-in genetic algorithm function to find the set of design variables with the minimum LLTE value. Constraints were applied to ensure that only physically meaningful shapes were considered and that the maximum factor of safety in the structure did not exceed two.

After the optimal keel design was determined, a flexural heel member was added such that when a user of similar body mass to that for which the foot was designed places all of his or her weight at the end of the heel, the minimum factor of safety in the structure was greater than two.

Results

The optimal foot size and shape is shown in Fig. 1. The maximum stress in the structure at the instant in stance immediately before push-off, when loads are highest, was σ_{max} = 39.9 MPa. The yield strength of nylon is 82.7 MPa, so the factor of safety was 2.07 and the constraint was satisfied.

Discussion

The wide curve foot designed here has an LLTE value of 0.091, which is similar to the best of the previously considered two design variable foot architectures at 0.099. However, this foot is nearly half the mass, at 469 g compared to 980 g for the lightest of several design iterations based on the two

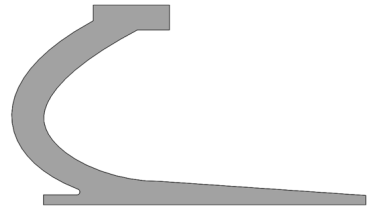

Figure 1. Optimal prosthetic foot shape and size

design variable architectures, and significantly easier to manufacture, as it consists of a single part that can be injection molded at large quantities, or 3D printed or machined at small quantities. Additionally, the compliant foot designed here provides a greater capacity for energy storage and return than our previous prototypes. FEA results showed that the optimal foot stored 23.6 J of elastic energy under the loads immediately prior to push-off, almost twice as much as the 12.0 J stored in the optimal two design variable architecture under the same loads.

Future work will expand on this method by including more loading scenarios to improve upon matching target kinematics throughout all of stance phase and adding design variables to increase the complexity of the potential designs considered. Other materials, such as carbon fiber, will also evaluated. The optimal design resulting from this work will be built and clinically tested in the coming months.

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