EFFECT OF A FOOT PAD ON THE PUSH-OFF AND JOINT TRAJECTORIES FOR THE POWERED TRANSFEMORAL PROSTHESIS: A PILOT STUDY

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INTRODUCTION

A powered transfemoral prosthesis is an assistive device that has two actuators for ankle and knee joints to provide stable walking to the lower extremity amputees (above-knee). Transfemoral amputees commonly have gait abnormalities that can lead to long term problems, such as fatigue, arthritis and scoliosis [1]. Reducing these abnormalities could lead to a reduction in some of the long-term issues.

Human walking consists of several events: heel-strike, foot-drop, heel-off, push-off, and toe-off [2]. The entire stance phase comprised approximately 60% of the gait cycle [2]. Between heel-off and push-off, positive work is required to insert energy into the moving body and to transition from the stance phase to the swing phase [3]. Adding a pad to the rigid foot can modulate the stiffness and damping between the rigid foot and the ground, which can simulate the natural human foot during the walking [4]. However, no systematic studies have investigated the effects of the foot pads in the gait characteristics of powered transfemoral prosthesis.

In this study, we investigated the effects of the foot pad on the gait kinetics and kinematics of the powered transfemoral prosthesis. Specifically, joint trajectory profiles and the push-off force at the toe are examined while foot pad condition was varied. It is hypothesized that providing a foot pad will enhance the gait kinetics and kinematics of the powered transfemoral prosthesis compared with the rigid foot.

METHODS

A custom-built fully actuated powered transfemoral prosthesis (AMPRO II) was used. AMPRO II is the second generation of a powered prosthesis developed at Texas A&M University. AMPRO II utilizes feedback from the prosthetic leg and force sensors at the prosthetic foot to synthesize the control signals for the knee and ankle joints of AMPRO II. The device has a height of 470mm and weight of 4.5kg. The control framework of the prosthesis consists of two different strategies: impedance control and trajectory tracking. The impedance control was utilized during the stance phase, and the desired human joint angle trajectories were tracked by PD control during the swing phase.

Figure 1: A foot considering a toe joint with spring steel for providing required push-off force. For this study, we constrained the toe joint with a rigid bar such that the stiffness of the foot was infinite. Please see the foot pad was attached to the existing foot of the prosthesis.

The experiment was performed on one subject (age: 29 years, height: 175cm, weight: 75kg, healthy non-amputee male) who walked on a level-ground treadmill. We tested 2 different conditions: no foot pad vs. foot pad. A treadmill speed was selected by user's preference (0.5 m/s). For each condition, the subject walked 20 gait cycles. Kinematic data (i.e., knee and ankle joint angles of the prosthesis via the embedded encoder) and kinetic data (i.e., force at the toe) were measured. The subject used an adapter to simulate an amputee gait. In order to measure toe force data, 2 force sensors (FlexiForce, Phidgets Inc. Calgary, AB, Canada) were embedded at the prosthetic foot. A qualitative analysis for both the toe

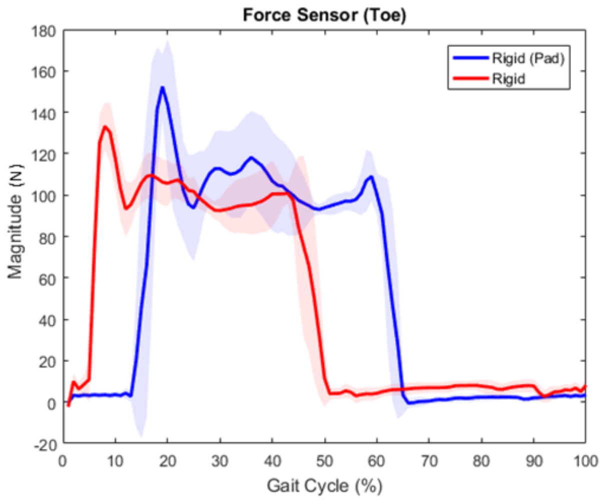

Figure 2: Force sensor (toe) data from 2 different cases: rigid foot with a pad vs. rigid foot without the pad. The bold lines are indicated the mean value of the sensor data, and the shaded regions are indicated the range within one standard deviation.

force profile and joint trajectories are performed without statistical analysis for this case study.

RESULTS AND DISCUSSION

Figure 2 shows the toe force profiles during the gait cycle with various foot pad conditions. For the rigid foot with no pad, the foot-drop happened around 10% of the gait cycle whereas for the rigid foot with a pad, the first peak happened around 20%. For the normal human gait, the first peak happens between 10% and 20% [5]. It is also clearly observed that the onset of the second peak was delayed from 43% to 60% for a rigid foot with a pad compared with a rigid foot without a pad. The delayed onset of the first peak is possibly due to the delayed transfer of the force to the force sensors due to the soft material of the pad. By varying the softness of the pad, the onset of the first and the second peaks seems to be modulated.

It is interesting to note that the second peak was delayed to 60%, suggesting that push-off may have occurred late compared with the rigid foot without a pad. Figure 3 supports this argument. When a foot pad existed, the push-off happened around 60% whereas when the pad was not used the push-off happened around 42% (Figure 3a). This was also supported by the ankle torque profile (Figure 3c). It was found from the literature that push-off usually happens between 50-60% of the gait cycle [5]. More

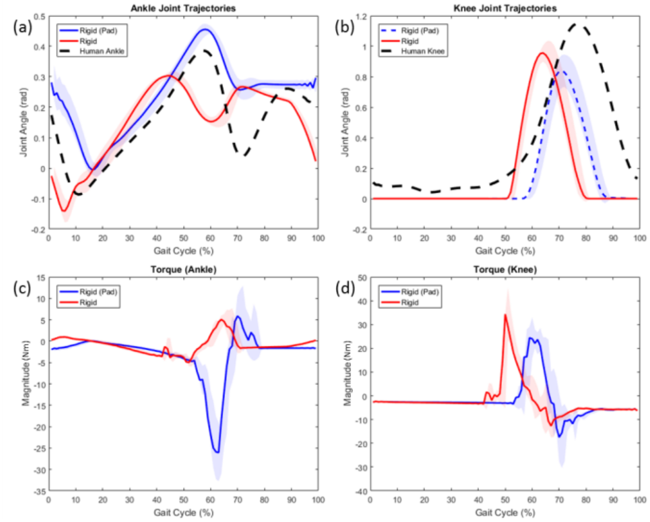

Figure 3: Ankle joint trajectory (a), knee joint trajectory (b), ankle torque (c), and knee torque (d) from 2 different cases: rigid foot with a pad vs. rigid foot without the pad. Dashed black line indicates the joint trajectory of the normal healthy gait.

interestingly, when a foot pad was used, the ankle and knee joint trajectories resembled the normal human walking with higher Pearson correlation (0.89 vs. 0.56 for ankle and 0.81 vs. 0.38 for knee).

CONCLUSIONS

In this pilot study, it was observed that the existence of the foot pad with appropriate softness enhanced both the onset timing of the push-off and the joint trajectory profiles for both ankle and knee. We are now planning to conduct a systematic experiment with more number of subjects, more conditions of the softness of the pad, motion capture system and force plates. We anticipate that optimal softness of the foot pad will provide optimal push-off with more normative joint trajectories for the powered transfemoral prosthesis.

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ACKNOWLEDGEMENTS

The authors thank to Namita Anil Kumar for the assistance to design a new prosthetic foot.