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A Preliminary Investigation of Powered Prostheses for Improved Walking Biomechanics in Bilateral Transfemoral Amputees

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Abstract—**The authors conducted a preliminary investigation of the extent to which a pair of powered prostheses can provide improved gait biomechanics in bilateral transfemoral amputee walking. Specifically, a finite state-based impedance controller for level ground walking was implemented in a pair of powered knee and ankle prostheses. The efficacy of the powered prostheses and impedance-based controllers was tested on a healthy subject using able-body adapters. Motion capture data was collected while the subject performed treadmill walking with the powered prostheses. This kinematic data is compared to that of healthy subjects, and also to previously published data for bilateral transfemoral amputee gait with passive prostheses. The comparison indicates that the powered prostheses are able to provide a walking gait that is considerably more representative of healthy biomechanical gait relative to passive prostheses.**

I. INTRODUCTION

Bilateral transfemoral amputees constitute a relatively small proportion of lower limb amputees (see, for example, [3, 4]). However, a bilateral transfemoral amputation is a much more significant disability than a unilateral transfemoral amputation. Specifically, in the case of the latter, an amputee is able to compensate substantially with his or her sound side to address biomechanical deficiencies in the prosthetic limb. Relative to healthy individuals, unilateral amputees rely disproportionately on their sound side leg for providing the net power output required for stair ascent, slope ascent, and for sit-to-stand transitions, all of which require net positive power at the knee and/or ankle joints [5-7]. Unilateral transfemoral amputees also provide significant compensatory effort with their sound side leg in activities which do not necessarily require net positive power, such as level walking, slope and stair descent, standing (particularly on uneven terrain), and stand-to-sit transitions [5-8]. For example, “heel hiking” is a common compensatory action observed during sound-side stance in level walking. Heel hiking elevates the amputee’s center of mass in order to increase the swing-side clearance between the prosthetic foot and ground, thus decreasing the likelihood of scuffing or stumbling with the prosthetic leg. Unlike unilateral amputees, bilateral transfemoral amputees lack a sound-side limb to provide compensation for deficiencies in gait biomechanics on the prosthetic side.

Recently, a number of powered lower limb prostheses have begun to emerge. Like the biomechanically intact lower limb, such prostheses are capable of generating net positive power at the knee and/or ankle joints. Recent publications indicate that such prostheses can provide improved kinematics, energetics and stability for unilateral lower limb amputees [10-12].

The authors have previously developed a powered transfemoral prosthesis and demonstrated improved kinetics for a variety of activities in unilateral amputees [13-15]. Given the inability of bilateral transfemoral amputees to compensate with a contralateral limb for the lack of power in passive prostheses, powered prostheses can potentially provide substantial benefit to the bilateral transfemoral amputee population.

This paper describes a preliminary investigation into the feasibility of bilateral transfemoral walking with powered prostheses. In particular, this paper presents an implementation of a pair of powered prostheses for level walking, controlled respectively by finite state-based impedance controllers. In order to assess efficacy, these prostheses are implemented on a healthy subject using a pair of able-body adapters. The kinematics of the knee and ankle joints with the powered prostheses during level treadmill walking are compared to those of healthy walking, and also compared to a bilateral transfemoral amputee walking with passive prostheses. This comparison indicates that the powered prostheses are able to provide considerably improved gait kinematics relative to the passive devices.

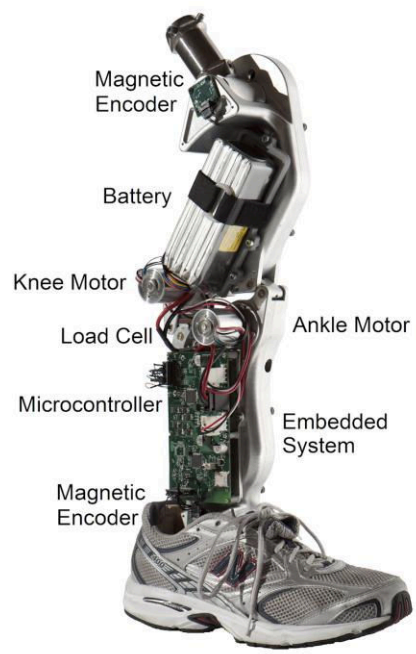

Figure 1. One of the two powered prostheses used in the experiments.

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II. Methods

A. Powered Prostheses

Two copies of a powered knee and ankle prosthesis similar to the design presented in [13] were used for this study (see Fig. 1). The prostheses are actuated by brushless DC motors and can provide up to 90 Nm of torque at the knee and 100 Nm at the ankle. Each prosthesis is powered by an onboard 3300 mA·hr lithium-polymer battery. All control and servo-electronics are embedded on the device, and the main control element is a 32-bit microcontroller from Microchip Technology, Inc. Sensors are sampled at 500 Hz and include knee and ankle joint angles and velocities, the axial load in the shank, and a 6-axis inertial measurement unit.

B. Walking Controller

The control of each prosthesis is governed by a finite state-based impedance model similar to the one presented in [13]. The state chart that describes the controller is shown in Fig. 2. State transitions are listed in Table I. The main deviation from the previously published controller is the addition of a high impedance state called pre-landing at the end of swing. The purpose of this state is to halt any remaining kinetic energy in the knee and prepare the prosthesis for the loading that takes place during heel strike. Since heel strike is detected by a load threshold, there is a small degree of latency between the actual moment of ground contact and the corresponding state transition. The higher impedance tuned for the pre-landing state helps prevent excessive flexion during this initial moment of loading.

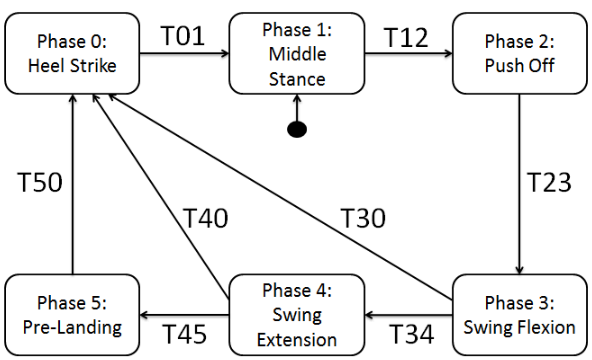

Figure 2. Updated walking state machine.

TABLE I. State Transitions of the Walking Controller

Transition	Description	Thresholds
T01	Foot has fully landed in stance.	↑ Ankle Torque
T12	Ankle begins actively pushing off.	↑ Ankle Angle
T23	Foot leaves the ground.	↓ Shank Load
T34	Maximum knee flexed is reached.	↓ Knee Velocity
T45	Knee straightens before heel strike.	↓ Knee Angle
T50/40/30	Ground contact is detected.	↑ Shank Load

The state behavior for each joint is governed by the passive control law,

$$\tau = k(\theta - \theta_{eq}) + b\omega\,, \tag{1}$$

where τ denotes the torque reference for the joint, and θ and ω denote the angular position and velocity of the joint, respectively. The three remaining parameters, k, b, and θ_{eq}, can be recognized as effective stiffness, damping, and set point terms, making (1) the equation for a (virtual) rotary spring and damper. Since there are two joints per prosthesis, three impedance parameters per joint, and six states in the walking controller, there are 36 impedance parameters per device and 72 total impedance parameters to select for level walking in a bilateral application.

C. Testing

In order to test the efficacy of the powered prostheses and associated controllers for bilateral transfemoral applications, a pair of able-body adapters was constructed to allow a healthy subject to don and use the powered prostheses. The adapters, fabricated from plastic and aluminum, immobilized the user's knee joints at approximately 100 degrees of flexion. A standard pyramid connector was installed below each of the subject's intact anatomical knees in order to mount the respective powered prostheses. The powered prostheses were configured to a height of 40 cm (15.75 in) from knee center to ankle center. This configuration added approximately 15.25 cm (6 in) to the subject's normal height. The subject's effective thigh length with the able-body adapters in place was approximately 59 cm (23.23 in).

The subject was allowed to practice over-ground walking at a self-selected speed for several hours using an overhead harness and forearm crutches. Once the subject was able to stand comfortably on the devices and initiate and cease walking on command, he began practicing walking on a treadmill with handrails for support. After the subject was comfortable with treadmill walking at a speed of 1.0 m/s, motion capture data was collected for the lower limbs. Six trials were recorded at 1.0 m/s. Data from a consistent and representative trial was used for gait evaluation.

Markers for the hips were placed at the top of the iliac crest on each side of the subject. Knee and ankle markers were placed directly over the prosthetic joint axes. Additional markers were used to define rigid bodies for the thighs, shanks and feet. Marker positions were tracked by 12 Optitrack S250e infrared cameras running at 120 Hz. The data were recorded using the NaturalPoint ARENA software environment, resampled to 100 Hz, and exported as BVH files. Sagittal plane joint angles were extracted from the BVH files in Matlab and up-sampled to 1000 Hz. Each powered prosthesis also logs time-stamped control signals at 250 Hz during normal operation. These data were also imported into Matlab, resampled to 1000 Hz, and synchronized with one another and the motion capture data by minimizing the sum of the squared error between the sagittal plane ankle joint angles. It should be noted that the measures of the sagittal plane knee angles deviated somewhat from the angles measured by the prostheses due to the compliance of the able-body adapters causing a deformation of the thigh (modeled as a rigid body).

Once the data were synchronized in Matlab, the internal state of the right prosthesis was used to parse the data into individual strides. Each full stride (as measured by complete transitions through the state model from heel strike to heel strike) was normalized in terms of percentage of stride.

III. Results

The joint angles for the knee and ankle from 20 consecutive strides are plotted in Fig. 3. These plots are referenced from heel strike on the right side and contain data from both sides. Although peaks in knee flexion for both stance and swing are slightly smaller than seen in the healthy subject population, the trends are quite similar given the physical differences arising from the use of the able body adapters.

The mode transitions from each leg are represented by the gray bands in the center of Fig. 3. The swing phases of each leg are highlighted in lighter gray. From this plot, it can be seen that double stance takes place predominantly during the push-off phase of the trailing leg (Phase 2) and the heel strike phase of the leading leg (Phase 0). The beginning of each phase is depicted visually in Fig. 4 from both video and motion capture.

Fig. 5 shows the mean stride for the right side powered prosthesis in comparison to healthy subjects and an amputee with bilateral knee disarticulations from the literature [1, 9]. Most notable are the improvement in stance knee flexion seen in the knee plot and the evidence of powered push-off indicated in the ankle plot by the presence of a large plantarflexion at approximately 60% of the stride.

Several common temporal characteristics of level gait are reprinted from [2] in Table II. These data are from a sample of men with ages of 19-32 years. This particular set was chosen because the healthy subject used in this experiment would fit this population if he were using his natural limbs. The same temporal characteristics were extracted from the

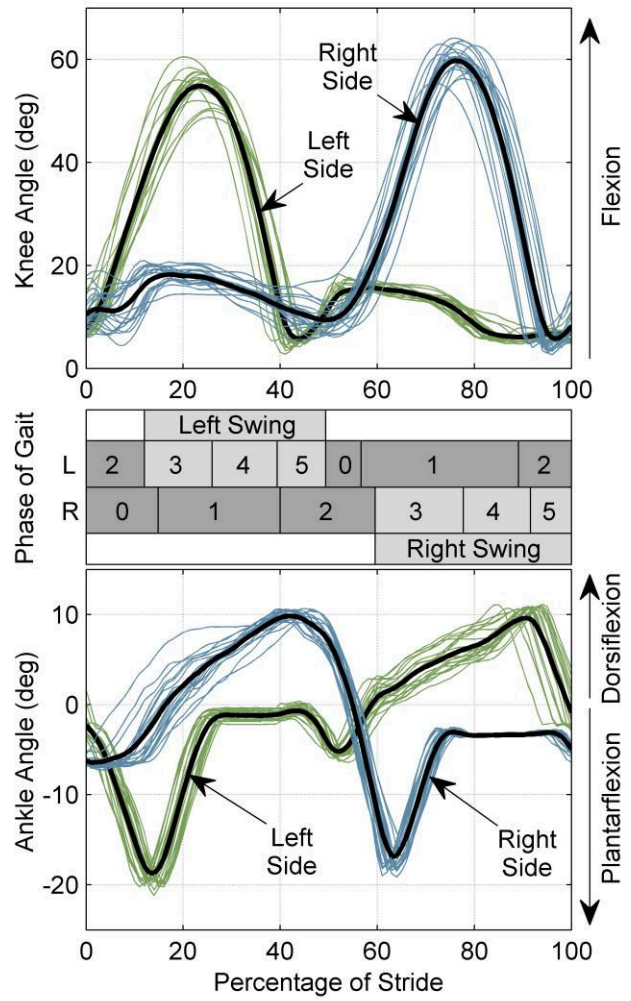

Figure 3. Lower limb kinematics for 20 consecutive strides with bilateral powered prostheses as executed by a healthy subject wearing able-body adapters. The mean of all strides is represented as a thick black line for each side.

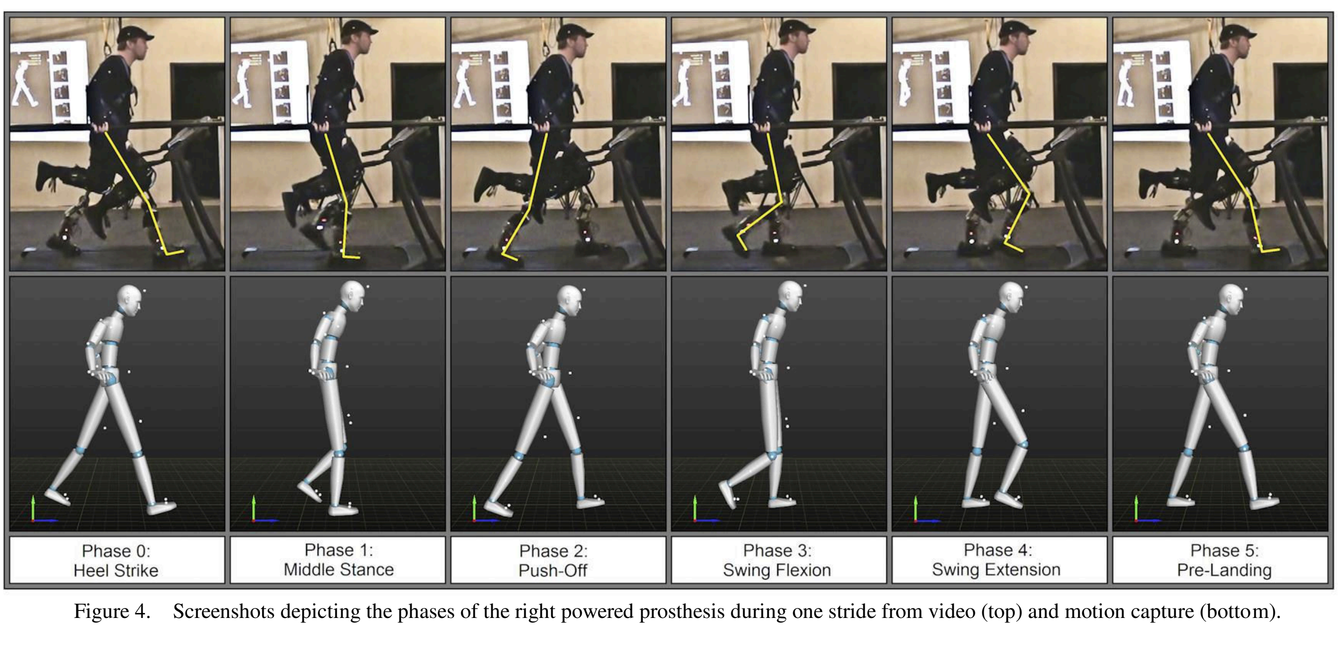

Figure 4. Screenshots depicting the phases of the right powered prosthesis during one stride from video (top) and motion capture (bottom).

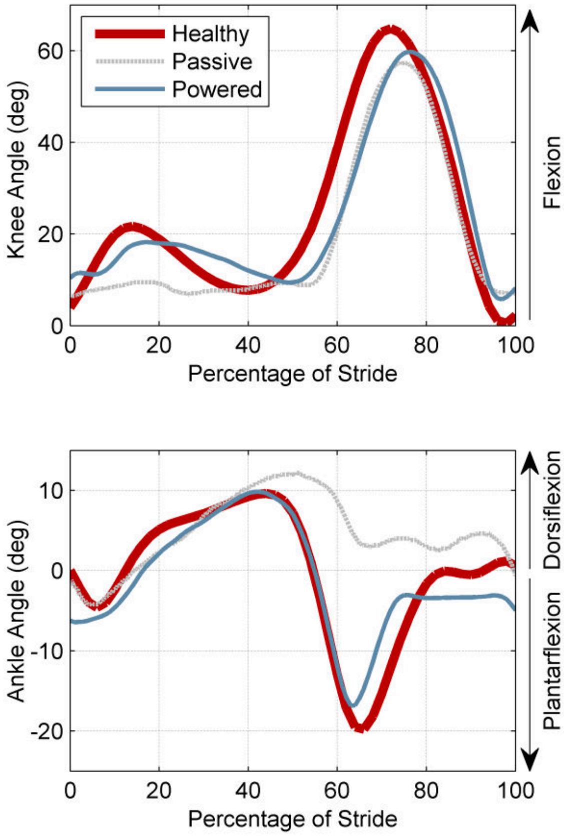

Figure 5. Kinematic comparison between healthy subjects (reprinted from [1]), a subject with bilateral knee disarticulations using microprocessor-controlled knees (reprinted from [9]), and a healthy subject using the bilateral powered prostheses with able-body adapters.

strides plotted in Fig. 3 for comparison purposes. Without offering any statistical claims, we can see that the gait achieved by the healthy subject using bilateral powered prostheses produces temporal characteristics very close to the means for the healthy population.

TABLE II. TEMPORAL GAIT CHARACTERISTICS

Parameter	Healthy Men (ages 19-32) [a]	Subject
Cadence (stride/min)	50 ± 8	40.2
Speed (m/min)	71.9 ± 18.3	59.0
Stride Length (m)	1.48	1.47
Right Stance %	60 ± 2	59.5
Left Stance %	60 ± 3	62.6
Single Stance % (R)	40 ± 2	37.4
Double Stance % (R)	10.2 ± 2.6	12.0

a. Reprinted from [2]. Values shown are mean ± standard deviation (n = 53).

Thus, the data indicate that a pair of powered transfemoral prostheses is able to provide gait biomechanics that are considerably more representative of healthy biomechanical gait relative to a pair of passive prostheses. Future work will entail testing of these prostheses on bilateral amputee subjects, and refining the controller to improve overall kinetics.

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