Effects of a powered ankle-foot prosthesis on kinetic loading of the contralateral limb: A case series

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***Abstract*—Lower-extremity amputees encounter a series of stress-related challenges. Among them is an increased risk of chronic joint disorders. For unilateral, transtibial amputees, we hypothesize that increasing the power output of the trailing, ankle-foot prosthesis during powered plantar flexion could mitigate kinetic loading applied to the leading, contralateral leg during walking. Here, we present a case series that analyzes kinetic factors of unilateral, transtibial amputee gait and forms a comparison between two types of ankle prostheses with varying power outputs. The factors examined here are impact resultant force, peak foot pressure at heel-strike, step-to-step transition work, and knee external adduction moment. The two prostheses are the amputee participant's daily-use passive ankle-foot prosthesis and the BiOM powered ankle-foot prosthesis capable of biologically accurate powered plantar flexion during late stance. In a preliminary study on two transtibial amputees walking over level terrain at a controlled speed (1.25 m/s), we observed average reductions of 8% in peak impact resultant force, 18% in impact resultant force loading rate, 8% in peak heel-strike foot pressure, and 15% in the 1st peak knee external adduction moment when the powered ankle-foot prosthesis was compared to the conventional passive prosthesis. Overall, our preliminary results suggest that more biomimetic prosthetic ankle-foot push-off during late stance may limit leading-leg musculoskeletal stress in walking.**

Keywords—amputee, ankle, biomechanics, external adduction moment, gait, loading rate, pressure, prosthesis, resultant force, transtibial

I. Introduction

Over 100,000 lower-extremity amputations occur every year in the United States alone [1]. Though advanced materials have spurred great strides in lower-extremity prostheses for this population, the biological ankle generates far more power and mechanical energy than even the most commonly used passive ankle prostheses [2, 3]. Further, quasi-passive prosthetic ankle joints that employ computer-controlled damping and swing phase position modulation improve gait stability but are still incapable of emulating normal biomechanical ankle function (élan Ankle Prostheses from www.endolite.com, Proprio Foot Ankle Prosthesis from www.ossur.com). Using passive and quasi-passive prostheses, lower-extremity amputees continue to experience gait pathologies and exhibit higher metabolic demands, gait asymmetry, and reduced walking speed [4, 5]. These abnormalities put them at an increased risk of several medical complications, including degradation of leg joints, skin disorders, and excessive pain in the contralateral leg [6, 7]. Each has been linked to excessive stress or pressure being applied to specific regions of the lower extermities.

More specifically, transtibial amputees using passive or quasi-passive prostheses display abnormal gait patterns due in part to the lack of a calf muscle and a fully functioning ankle joint. The calf musculature plays a key role in human walking, generating propulsion forces during mid to late stance through ankle powered plantar flexion to push the body upwards and forwards with each walking step [4]. Nearly zero net mechanical work is performed on the center of mass (COM) during a single stride; however, both legs must perform positive and negative work on the center of mass during double support to transition between steps [8]. This work is a function of the ground reaction forces and COM velocity vector, and is critical for efficient movement. At constant walking speed, both legs should perform nearly equal and opposite magnitudes of work. For amputees, the affected trailing limb performs an insufficient amount of work, leaving the unaffected leading limb to compensate [9], perhaps contributing to excessive loads applied to the contralateral leg.

Among the joint disorders for which unilateral, transtibial amputees have increased susceptibility is knee osteoarthritis, especially in the leg opposite amputation. The development of this disorder has been linked to abnormal levels of knee external adduction moment (EAM) [7]. Further, the severity of the disorder has been linked to high peak EAM magnitudes. On the other hand, skin ulcerations are prevalent amongst amputees and have been linked to excessive pressure. Specifically, high values of peak foot pressure (PP) are associated with an increased risk of foot ulcerations [10, 11]. Faced with these issues, often amputees decide to reduce or forgo use of their prosthesis, resulting in a more sedentary lifestyle and possibly accelerating vascular disease.

Recent endeavors in lower-extremity prostheses have attempted to combat these deficiencies by becoming more biomimetic, incorporating advanced sensor and actuator technology to enhance functionality [12–16]. Now, lower extremity prosthetic components are beginning to more closely emulate the biological leg. The BiOM Ankle, the first commercially-available powered ankle-foot prosthesis, has restored nearly normal metabolic demand to transtibial

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amputees through its biologically accurate power generation and timing [9]. Biomimetic approaches are also being extended into powered knee prostheses, as well as powered knee-ankle prostheses, to assist swing phase, improve stair ascent, and restore ankle actuation to transfemoral amputees [14, 15].

This case series seeked to uncover how the BiOM ankle-foot prosthesis, which more closely emulates the biological ankle joint compared to passive and quasi-passive prostheses [9], affects kinetic loading applied to the contralateral limb of walking, unilateral transtibial amputees. Increased levels of powered plantar flexion by the trailing leg prosthesis has been shown to mitigate the 1st peak knee EAM on the leading, intact leg of walking unilateral transtibial amputees [16]. Thus, in this case-series investigation, we hypothesized that the BiOM prosthesis would mitigate kinetic loading applied to the intact leg of unilateral, transtibial amputees when campared to a passive prosthesis for a level-ground walking speed of 1.25 m/sec. Here, we measured six factors to describe kinetic loading on the contralateral limb: peak resultant ground reaction force (PRF), resultant-force loading rate (FLR), peak foot pressure (PP), leading and trailing leg step transition work (TW), peak knee external adduction moment (EAM), and EAM rate. As a preliminary evaluation of this hypothesis, we analyzed amputee gait characteristics through a set of level-ground walking experiments on two unilateral, transtibial amputees using conventional passive prostheses and the powered BiOM prosthesis.

II. Methods

A. Study Participants

We recruited two unilateral, transtibial amputees to participate in the study (See Table I). The primary factors influencing participant selection were activity level and socket to floor length. To ensure that each participant was capable of completing the study, all were characterized as K3 walkers with the capacity to vary both cadence and walking speed. A certified prosthetist referred all participants and verified that each had no additional medical disorders and met the activity level requirement. The distance from the bottom of the prosthetic socket to the floor was critical because it dictated whether the amputee could use the powered prosthesis. All participants had a socket to floor length greater than the approximate 9-inch height of the powered prosthesis.

The MIT Committee on the Use of Humans as Experimental Subjects (COUHES) pre-approved this research. Participants signed an informed consent agreement prior to the start of the experimental study.

B. Procedure

This study examined level surface walking at a controlled speed equal to 1.25 m/s. Two separate sessions were held for each study participant. The first focused on collecting the anthropometric information needed to fit the powered prosthesis to the participant and prepare for the experiments. The prosthetic fitting process included ankle control parameter tuning, prosthetic height adjustments, and alignment adjustments. A Certified Prosthetist was present to perform proper fitting and alignment.

TABLE I. **Participant Information.** Characteristics of each study participant.

Participant	Sex	Age (yrs)	Height (m)	Mass (kg)	Ambulatory K Level
1	Male	28	1.75	72.1	K3
2	Male	40	1.73	69.3	K3

Participant	Etiology	Years Since Amputation	Prosthesis Type
1	Congenital Disorder	28	Silhouette Freedom Innovations©
2	Traumatic	9	Renegade Freedom Innovations©

All data were recorded in the second session to ensure consistency in pressure sensor locations and results. Participants were asked to walk along an 8-meter walkway with embedded force plates. Two different walking conditions were tested: one condition used the participant's prescribed prosthesis (Table I) and the second condition used a powered ankle-foot prosthesis. We recorded kinematic and ground reaction force measurements inside a motion capture facility that contained a 12-camera VICON Motion Analysis System sampled at 120 Hz and two AMTI force plates sampled at 960 Hz. A modified Helen Hayes marker set and forty-six 12 mm reflective markers were used to pinpoint key locations along the participant's entire body. Additionally, we used marker tracking to measure and control walking speed, only accepting trials within ± 5% of a desired speed equal to 1.25 m/sec.

Further, participants were fit with pressure sensing shoe insoles to monitor real-time pressure characteristics along the biological foot. We used the Wireless F-Scan Pressure Mapping System manufactured by Tekscan© Inc., which is made specifically for gait studies. It comes equipped with thin, flexible insoles that insert easily into the shoe to map foot pressure distribution. We altered/trimmed the insoles to fit each participant's foot size and shoe contour. Attachment and security are critical to preserving the sensors and obtaining accurate recordings. We used thin, clear, double-sided tape to adhere the sensors to the shoe interior so as to minimize shifting or creasing that may have contaminated recordings. In addition, participants wore socks to reduce friction between their foot and the sensor. All pressure recordings were collected at a 100 Hz sampling frequency.

Prior to completing the powered prosthetic evaluations, participants were given 1 hr to acclimate to walking with the new device. Participants walked around the lab space until they felt comfortable. The same shoe and sock were used for both conditions to ensure consistency in marker placement and shoe sensor placement.

Motion capture data were filtered using a butterworth filter with a 30 Hz cutoff frequency before being uploaded into Musculographics' Software for Interactive Musculoskeletal Modeling (SIMM) to obtain kinematics and kinetics. We computed the resultant ground reaction force as the magnitude of the ground reaction force vector. Average resultant force loading rate (FLR) was defined as the change in resultant force from heel strike to the first peak divided by the elapsed time from heel strike to the first peak, or equivalently:

$$FLR = \frac{RF_{fp} - RF_{hs}}{t_{fp} - t_{hs}} \qquad (1)$$

where RF_{fp}, RF_{hs}, t_{fp}, and t_{hs}, respresent resultant ground reaction force at the first peak, resultant ground reaction force at heel strike, time at the first peak, and time at heel strike, respectively. Step-to-step transition cost was determined using the method introduced by Donelan et al [8]. SIMM's Dynamics Pipeline allowed us to perform inverse dynamics to obtain knee EAM. Similar to average resultant force loading rate, we computed EAM rate using the following:

$$EAM\ Rate = \frac{EAM_{fp} - EAM_{hs}}{t_{fp} - t_{hs}} \qquad (2)$$

where EAM_{fp}, EAM_{hs}, t_{fp}, and t_{hs}, respresent the knee external adduction moment at the first peak, knee external adduction moment at heel strike, time at the first peak, and the time at heel strike, respectively.

C. *BiOM Ankle-Foot Prosthesis*

The BiOM Ankle-Foot prosthesis differs from conventional passive and quasi-passive devices in its ability to inject nonconservative positive work into the user's walking stride throughout the stance period [9]. The device employs both passive and motorized elements to more closely emulate human ankle-foot functions. Its mechanisms are described in Figure 1. Like the biological ankle, the device generates net positive work during the stance phase and permits toe clearance during the swing phase [12]. It uses a series-elastic actuator (SEA), configured with a brushless motor and ball screw transmission in series with a carbon composite leaf spring, to store and release motor energy and improve efficiency and power output. The BiOM features a carbon-composite foot at its base for added compliance, electronics completely encapsulated within a single housing, and a modular Lithium-Polymer battery to power the motor. The prosthesis weighs 2.36 kg.

Sensory feedback data allow the powered prosthesis to achieve biomimetic function, constantly varying joint torque and impedance throughout the gait cycle to match biological norms. Biologically-inspired control schemes govern the behavior of the device, enabling proper timing and magnitude of ankle power for a wide range of velocities [17, 18]. The adaptive ankle controller employs feedback sensory information from both the actuator torque contribution and the net torque on the ankle joint.

Parameter tuning was a critical step in the setup procedure for the powered prosthesis. It was accomplished via a Bluetooth connection between the prosthesis and an Android tablet. This connection allowed for real-time control of three control parameters that govern ankle mechanical behavior, namely, the amount of powered plantar flexion, the timing of powered plantar flexion, and the stiffness of the device during controlled plantar flexion. These three parameters were

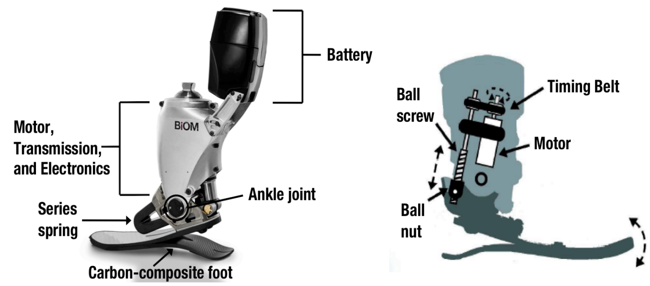

Fig. 1. **Powered Ankle-Foot Prosthesis.** The powered prosthesis uses a series-elastic actuator comprised of a brushless 200 Watt DC motor, ball screw transmission, and a carbon-composite series leaf spring. The actuator is capable of performing dorsiflexion and plantar flexion movements, stiffness and damping impedance control, as well as nonconservative positive work about the ankle joint. The motor, transmission, and all electronics are contained about the prosthetic ankle joint, and a modular Lithium-polymer battery is housed just proximal to the joint. The prosthesis uses a carbon-composite leaf spring at its base for added compliance at the heel and forefoot.

TABLE II. DYNAMIC BEHAVIOR OF THE POWERED PROSTHESIS. TOE-OFF ANGLE, STANCE ANKLE NET WORK, AND STANCE PEAK MECHANICAL POWER FOR THE POWERED ANKLE PROSTHESIS ARE LISTED FOR A WALKING SPEED OF 1.25M/SEC. BIOLOGICAL NORMS FOR EACH QUANTITY ARE PROVIDED FOR DIRECT COMPARISON. BIOLOGICAL DATA OF TOE-OFF ANGLE AND NET WORK WERE TAKEN FROM HERR & GRABOWSKI [9] AND PEAK POWER FROM WINTER [19].

Participant	Toe-off Angle (deg)	Ankle Net Work (J/kg)	Peak Power (W/kg)
1	20 (1)	0.17 (0.03)	5 (1)
2	9 (2)	0.12 (0.01)	3 (1)
Avg (s.d.)	15 (7)	0.15 (0.04)	4 (2)
Biol. Norms	18 (3)	0.10 (0.05)	3 (1)

adjusted using the Bluetooth connection until normative values of net prosthetic ankle work, peak power, and toe-off angle were achieved.

We compared the powered prosthesis ankle data to biological norms, as reported in [9, 19], to determine the effectiveness of control parameter adjustments. Using this comparison, we found control parameters that produced reasonable values compared to the biological norms for ankle toe-off angle, stance phase ankle net work, and stance peak power. These values are reported in Table II along with corresponding normative values.

III. Results

We present kinetic-loading results for the two unilateral, transtibial amputees described in Table I. Our results compare each participant's prescribed, passive prosthesis to the powered prosthesis shown in Figure 1. We sought to quantify the work done by both devices during double support to facilitate step-to-step gait transitions. As shown in Table III, we observed 54% and 44% decreases for participant #1 and participant #2, respectively, in negative leading-leg transition work when each used the powered prosthesis compared to the conventional passive prosthesis. This was more than likely spawned by the 267% and 400% increases in positive trailing-leg transition work of the powered prosthesis compared to the passive prostheses.

Peak resultant ground reaction force, peak foot pressure, and knee external adduction moment for both participants are shown in Figure 2. Each is displayed versus gait cycle

TABLE III. **Step-to-Step Transition Work, Contralateral Limb Force, Loading Rate, Foot Pressure, EAM Peaks, and Average EAM Loading Rate.** Trailing and leading leg step-to-step transition work (TW), peak resultant ground reaction force (PRF), resultant force loading rate (FLR), and peak foot pressure (PP), all normalized by body mass, are listed for each study participant and prosthesis. In addition, we compared the 1st and 2nd peaks of the external adduction moment (EAM) for the passive and powered prostheses. EAM rate, or the average derivative of the knee EAM from the beginning of stance to the EAM 1st peak, is also shown.

Participant	Passive Leading TW (J/kg)	Powered Leading TW (J/kg)	% Diff	Passive Trailing TW (J/kg)	Powered Trailing TW (J/kg)	% Diff
1	0.13	0.06	54	0.06	0.16	267
2	0.16	0.09	44	0.04	0.16	400
Avg (s.d.)	0.15 (0.02)	0.08 (0.03)	49	0.05 (0.01)	0.16 (0.04)	334

Participant	Passive PRF (N/kg)	Powered PRF (N/kg)	% Diff	Passive FLR (N/kg/ms)	Powered FLR (N/kg/ms)	% Diff	Passive PP (KPa/kg)	Powered PP (KPa/kg)	% Diff
1	12.4	12	3	7.1	5.8	18	2.3	2.2	4
2	12.7	11	13	6.4	5.3	17	2.6	2.3	12
Avg (s.d.)	12.6 (0.7)	12 (1)	8	6.8 (0.7)	5.6 (0.5)	18	2.5 (0.4)	2.3 (0.6)	8

Participant	Passive EAM Rate (Nm/kg/ms)	Powered EAM Rate (Nm/kg/ms)	% Diff	Passive 1st Peak EAM (Nm/kg)	Powered 1st Peak EAM (Nm/kg)	% Diff	Passive 2nd Peak EAM (Nm/kg)	Powered 2nd Peak EAM (Nm/kg)	% Diff
1	0.45	0.4	11	0.55	0.5	9	0.6	0.55	8
2	0.37	0.3	19	0.63	0.5	21	0.5	0.48	4
Avg (s.d.)	0.41 (0.08)	0.3 (0.1)	15	0.59 (0.07)	0.5 (0.1)	15	0.6 (0.1)	0.52 (0.08)	6

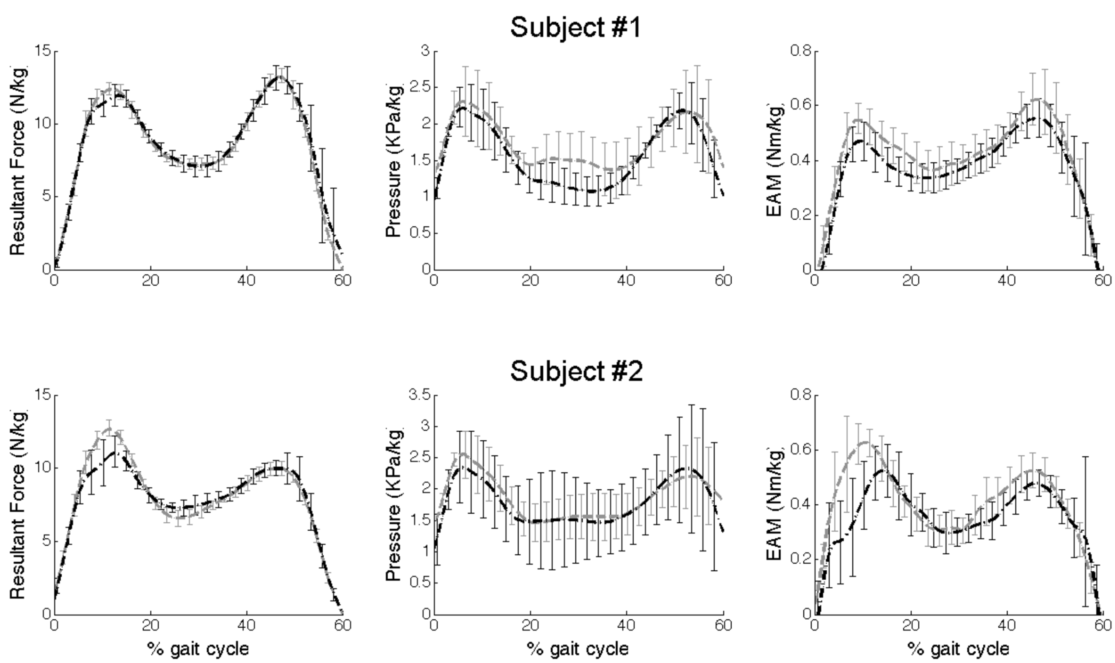

Fig. 2. **Contralateral Limb Force, Foot Pressure and Knee External Adduction Moment (EAM).** The grey lines indicate resultant force, foot pressure, and EAM on the leading, contralateral limb while walking with the prescribed passive prosthesis. The black lines represent the powered prosthesis. For all values, mean ± one standard deviation are shown. Data are plotted versus percent gait cycle where 0% occurs at heel strike of the leading leg.

percentage and normalized to body mass. From these plots, we extracted leading or unaffected leg peak resultant force, resultant force loading rate, and peak foot pressure (See Table III). Decreases were observed in all three quantities. The percentage differences in peak resultant force after heel strike were 3% and 13%, and the percent differences in resultant force loading rate were 17% and 18%. Additionally, the participants exhibited 4% and 12% reductions in peak pressure when walking with the powered prosthesis compared to their passive prosthesis.

Contralateral knee EAM results are also shown in Table III. We measured the 1st and 2nd peaks of EAM, as well as the EAM rate. Our results show that the 1st peak EAM decreased by 9% and 21%, 2nd peak EAM decreased by 8% and 4%, and EAM rate decreased by 11% and 19%.

IV. Discussion

This study investigated the effect of two distinct prosthetic designs on kinetic loading applied to the leading, intact leg. One prosthesis was passive and the other was powered. The desired outcome was to gain insight into the effect of powered plantar flexion, the major difference between the two devices, on factors of kinetic loading on the contralateral leg: step-to-step transition work, resultant force, pressure, and knee EAM. By measuring these factors, which have been linked to common medical complications [7, 8, 10, 11, 20], we hoped to uncover a relation between prosthesis selection and potential vulnerability to future disorders.

Previous studies and models have shown the importance of powered plantar flexion on center of mass displacement during the walking gait cycle [8], [20]. These studies can be used to explain the compensatory mechanisms displayed by lower-extremity amputees and excessive resultant ground reaction force of the leading leg during double support. Both cases shown here exhibit large increases in trailing leg transition work, which seem to produce decreases in leading leg transition work, as well as decreases in peak resultant force, loading rate, and peak foot pressure. This result suggests that increased powered plantar flexion mitigates some of the compensation seen in the contralateral leg of unilateral, transtibial amputees.

A large 1st peak knee EAM has been linked to knee osteoarthritis in the general population [7]. Similar to a previous study by Morgenroth et al. [16] that showed an inverse relationship between ankle push-off and peak knee EAM using passive and quasi-passive ankle-foot prostheses, we found that increasing ankle push-off between a passive prosthesis and a powered prosthesis decreased EAM in the cases presented here. Additionally, we observed a 6% reduction in the 2nd EAM peak. Given the case series nature of the present study, we cannot draw any definitive conclusions about these findings. However, it is interesting to note that given the nearly equal values of 2nd peak resultant force for both participants (Figure 2) the difference in 2nd EAM peak must be due to a difference in the knee/ground reaction force moment arm. This could mean that the participants traveled along a more stable trajectory using the powered prosthesis compared to the passive device. Future investigations are necessary to determine whether these observed behaviors are significant across a larger study population.

The major limitation of this study is the small number of participants (n=2). The results of our two cases suggest interesting differences between the prostheses, but further research is required to show significance across a larger amputee population. In addition, although our acclimation period was consistent with previous research [16], the relatively low accommodation period, as compared to the prescribed prosthesis, could be another limitation of the present investigation. Clearly, the impact of a relatively small acclimation period and the lack of specific gait training are important considerations for future research.

V. Conclusions

Lower-extremity amputees continue to experience gait pathologies while using prostheses that fail to emulate normal biological dynamics during the walking gait cycle. These gait abnormalities put them at an increased risk of musculoskeletal and skin comorbidities, many attributed to excessive kinetic load [6, 7]. In this investigation, we hypothesized that a prosthesis that exhibits normative levels of powered plantar flexion during the late stance period would mitigate kinetic loading applied to the leading intact leg of unilateral transtibial amputees when campared to a passive prosthesis. The data collected in the present case-series investigation are in support of this hypothesis, highlighting the potential importance of biomimetic prosthetic ankle interventions.

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