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Altering the tuning parameter settings of a commercial powered prosthetic foot to increase power during push-off may not reduce collisional work in the intact limb during gait

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Abstract

Background: Increased knee osteoarthritis risk in patients with unilateral lower extremity limb loss is attributed to increased intact limb loading. Modulating powered ankle prosthesis push-off power may be an effective way to modulate intact limb loading. We examined how changes in the parameter settings of a commercial prosthetic ankle affect power delivery during push-off and the resulting collisional work experienced by the intact limb in persons with unilateral lower extremity limb loss.

Methods: Five subjects with unilateral transtibial amputation were fitted with a commercially available powered ankle prosthesis (Ottobock Empower). Subjects walked on a treadmill in seven conditions, where ankle power delivery settings were adjusted using methods accessible to clinicians. Kinetics and kinematics data were collected.

Results: Standard adjustment of parameter settings within the prosthetic foot did not alter timing of peak prosthesis power or intact limb collisional work but did have a significant effect on the magnitude of positive prosthesis ankle work. Increased prosthesis work did not decrease intact limb collisional work as predicted.

Conclusions: Altering the parameter settings on a commercial powered ankle prosthesis affected the magnitude, but not the timing, of power delivered. Increased prosthesis push-off power did not decrease intact limb loading.

Keywords

transtibial amputation, robotic prosthesis, gait, biomechanics

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Background

Pain in the intact limb is a principal complaint of individuals with unilateral lower extremity amputation, particularly for long-time prosthesis users.[1] Intact limb pain may result from an increased loading of the intact limb during locomotion.[2] These factors likely contribute to the finding that individuals with unilateral lower extremity amputation are roughly 17 times more likely to develop knee osteoarthritis in their intact limb.[3] Because knee osteoarthritis has a profound impact on quality of life,[4] it is imperative to reduce this risk for the growing population of individuals with lower extremity amputation.[5]

Increased loading of the intact limb during locomotion has been linked with elevated risk of knee osteoarthritis in individuals with unilateral lower extremity amputation.[2] Dynamic walking models predict that this asymmetric loading is a result of the decreased push-off power provided by a prosthetic ankle.[6] With each step-to-step transition, the center of mass (COM) velocity needs to be redirected from moving forward and downward before the leading limb making heel contact to moving forward and upward after propulsion work performed by the trailing limb.[7,8] This involves the transfer of positive work from the trailing limb during push-off to offset the negative or collisional work absorbed by the leading limb during loading response.[7-9] In persons without amputation, the ankle is responsible for providing the push-off power and most of the positive propulsion work performed by the whole trailing limb needed for this COM redirection and forward propulsion.[10] According to dynamic walking models, decreased push-off power from the ankle in the trailing limb results in decreased propulsion work performed by the trailing limb on the COM. Walking speed is often maintained in these cases through increased contralateral hip joint work in both able-bodied and amputee populations.[6,11] This hip compensation, however, results in increased collisional work in the leading limb because of the limited redirection of COM velocity.[6,7] This then leads to greater loading and collisional work on the leading or intact limb, which may increase the knee joint loads linked with greater knee osteoarthritis risk.[12,13]

One method to combat asymmetric limb loading during gait in individuals with unilateral lower extremity amputation is to increase the push-off power delivered by the prosthetic foot. Dynamic walking principles predict that increasing the power delivered by the prosthetic foot during push-off would increase the propulsion work performed by the trailing limb on the body's CoM, thereby reducing the collisional work experienced by the intact limb. Several studies have also demonstrated that increased prosthesis power leads to a reduction in vertical ground reaction

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forces experienced by the intact limb.[14-17] Yet, loading symmetry has not been successfully restored with commercially available powered prostheses at normal walking speeds.[18,19] Increased prosthesis power has not been observed to have an effect on intact limb collisional work in healthy subjects or in persons with amputation when using a tethered, prosthesis emulator.[15,16] It remains to be seen whether persons with an unilateral amputation using a commercially available, untethered powered prosthesis would have reduced intact limb loading.

The Empower (Ottobock, Bedford, MA) is a commercially available powered prosthetic foot designed as a motor in series with a spring that is deformed through the combination of walking mechanics and motor output. Given the interplay between the user and the device, the relationship between the prosthesis motor output and the timing and magnitude of power delivered by the prosthetic foot is not straightforward. Therefore, the purpose of this study was to characterize the effects of adjusting the tunable parameter settings of a commercially available powered prosthesis on actual ankle power delivery and, ultimately, on the collisional work experienced by the leading (intact) limb in patients with unilateral transtibial limb loss. We tested two primary hypotheses. We hypothesized that altering the tunable parameters of a commercial prosthesis would alter both the time of peak power and the magnitude of positive work performed by the amputated leg during push-off. In addition, we hypothesized that the collisional work of the intact leading limb would be negatively correlated with positive work done during push-off with the prosthetic foot.

Methods

Five male subjects with unilateral transtibial amputation (mean age, 41 ± 17 years; mass, 79.0 ± 8.0 kg; height, 1.84 ± 0.08 m; time since amputation, 15 ± 9 years) gave written informed consent before participating in this institutionally approved study. Subjects were included in the study if they met the following inclusion criteria: cause of amputation was not associated with a dysvascular disease; there was adequate clearance to fit the robotic ankle prosthesis; the individual regularly used a prosthesis for ambulation and could demonstrate variable cadence; and were at least 8 months postamputation. Exclusion criteria included having a cardiovascular or balance problem that would prohibit individuals from walking on a treadmill.

All subjects were fitted with the same commercial robotic ankle prosthesis (Empower, Ottobock, Bedford, MA) by a certified prosthetist working in conjunction with a representative from the manufacturer. The subject's own prosthetic socket and suspension system were retained. Three participants had a vacuum suction suspension system, one used a patellar tendon bearing suspension system, and one had a pin/lock system. Prosthetic alignment was initially set up to duplicate the sagittal and coronal plane alignment of the subject's normal prosthesis. The alignment was then fine-tuned through clinical dynamic alignment.

Prosthesis tuning parameters were set by a certified prosthetist. The delivery of push-off power in this device is set using three tuning parameters that are accessible to clinicians through a graphical-user interface (GUI) on a mobile tablet. Two of the tuning parameters, termed "fast power time" and "slow power time," are indicated to correspond to the power delivery in fast and slow walking modes, respectively, as detected by the Empower device. In our study, these two input parameters were simultaneously adjusted on the GUI according to a predefined relationship provided by the manufacturer, which effectively constrained their use to a single-control dimension. A third parameter input called "fast power" could be engaged to either add or subtract from the overall effect of the prosthesis controller. Because of the proprietary nature of the control scheme, we were not made aware of how these seemingly arbitrary tuning parameters would affect the magnitude or time of peak push-off power delivered by the prosthesis. Therefore, a primary goal of our study was to first characterize the functional power delivered by the prosthesis to a user during gait. The baseline setting (B0) was determined using vendor-recommended "standard" parameter settings (Table 1). Upper and lower limits of the most extreme conditions were determined by adjusting the GUI parameter inputs to just within those where a very experienced Empower user demonstrated stumbling reactions or expressed feelings of unease. We then systematically decreased (−B1, −B2) and increased (+B1, +B2) these GUI parameter inputs in 9% increments from the baseline setting (Table 1). We then created two extreme conditions (−B3, +B3) by engaging the "fast power" GUI setting at +4% and −4% to more fully explore the parameter space. All together, we tested seven different prosthesis control parameter settings.

Subjects performed 3–5 minutes of level treadmill walking to acclimate to treadmill walking at 1.1 m/s with the prosthesis in the baseline setting. A comfortable setup was verbally acknowledged by each user and visually confirmed by the investigator. Each subject then walked on a level treadmill at 1.1 m/s for 3 minutes for each of seven randomized trials for the following conditions: baseline tuning parameters and six altered parameter conditions (−B3, −B2, −B1, B0, +B1, +B2, +B3). Kinetic data were obtained through force plates (AMTI, Watertown, MA) instrumented into a custom, dual-belt treadmill with a sampling rate of 1080 Hz.[19] We collected lower body kinematics at 120 Hz using an 8-camera motion capture system (Vicon, Centennial, CO; Visual 3D, C-Motion, Germantown, MD). Reflective markers were placed on boney landmarks using a modified Helen Hayes marker set.[21]

The last 15 strides were analyzed for each subject per condition. Visual 3D (C-Motion, Germantown, MD) was used to filter data (fourth-order Butterworth low-pass with cutoff frequencies at 25 Hz and 10 Hz for force and marker data, respectively) and to calculate

Table 1. Tuning parameter settings used for each testing condition.

	−B3	−B2	−B1	Baseline B0	+B1	+B2	+B3
Fast power time	30%	30%	39%	48%	57%	66%	66%
Slow power time	5%	5%	14%	23%	32%	41%	41%
Fast power	+4%	Base	Base	Base	Base	Base	−4%

Parameter names were supplied by manufacturer and do not necessarily indicate the actual effect on the end user.

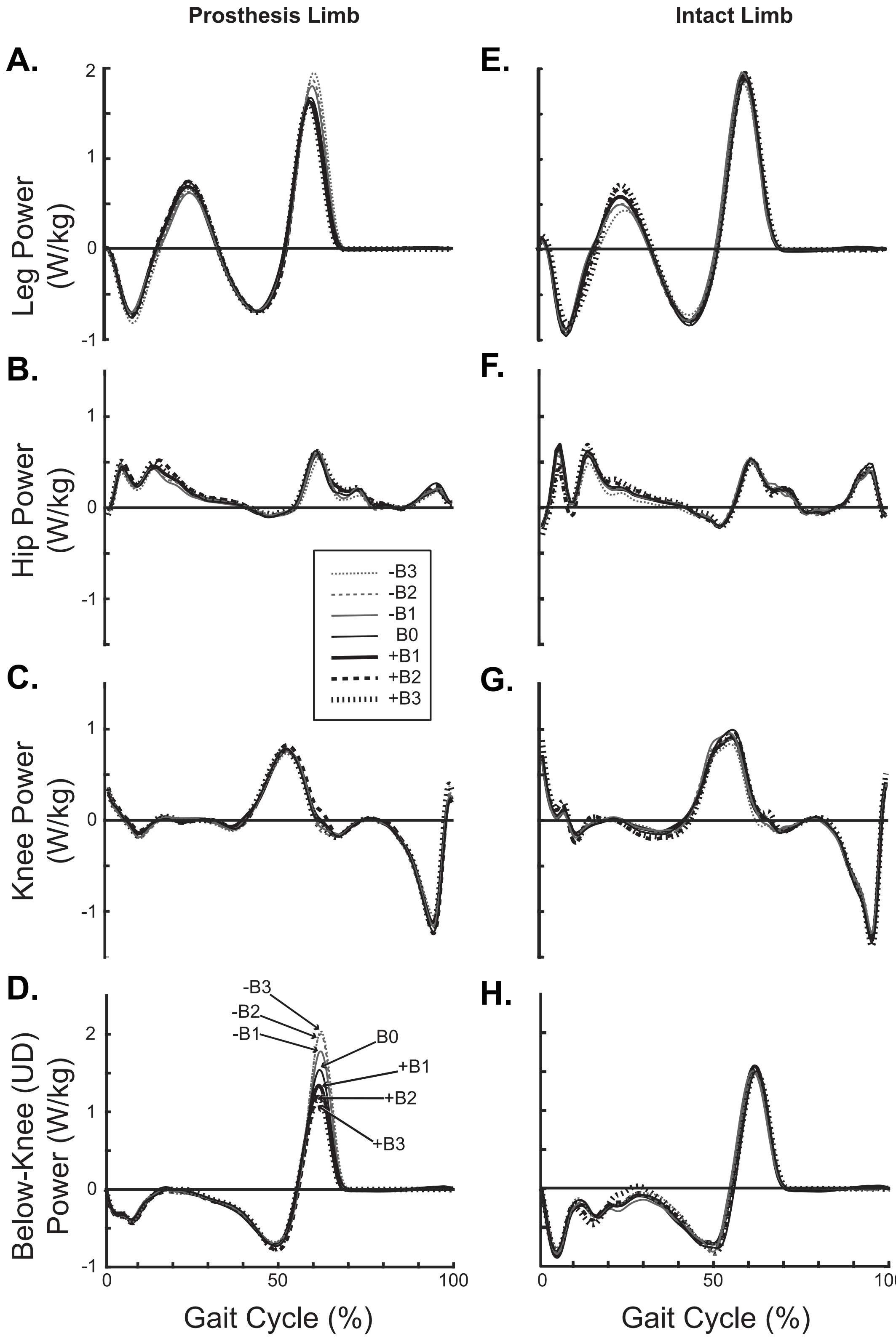

Figure 1. Prosthesis and intact joint and leg powers. Joint powers for the prosthesis/amputated limb (a–d) and intact limb (e–h) from a representative subject. Mean joint power trajectories over the gait cycle are shown for the baseline tuning parameter condition (B0, thin black line) along with the six tuning parameter conditions: +B1 (thick black line), +B2 (thick dashed black line), +B3 (thick dotted black line), −B1 (thin gray line), −B2 (thin dashed gray line), and −B3 (thin dotted gray line). Power produced by the prosthesis (d) increased during push-off across all tuning parameter conditions, but this did not result in corresponding changes in other joint and leg powers.

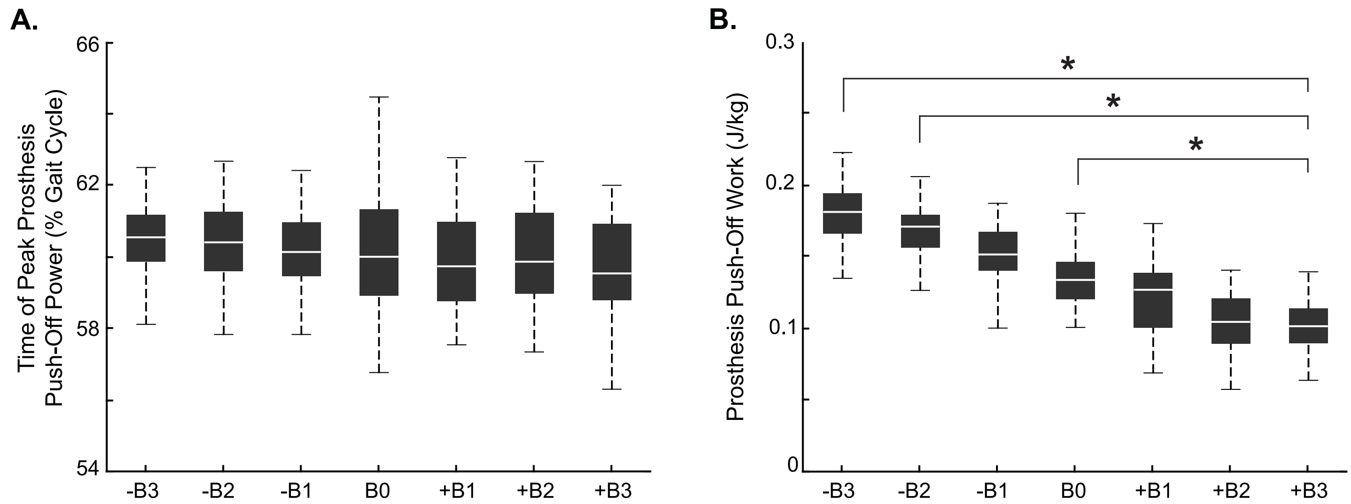

Figure 2. Time of peak prosthesis push-off power and prosthesis push-off work. Box and whiskers plots of time of peak prosthesis push-off power (a) and magnitude of positive prosthesis push-off work (b) averaged across subjects for each timing condition (n = 4). * indicates significant difference from +B3 tuning condition ($P \leq 0.05$).

joint angles, moments, and powers. Kinematic and kinetic data were separated into individual strides and normalized to the percentage of the gait cycle by identifying foot contact and lift-off events using a ground reaction force threshold of 32 N as in previous studies.[22,23] The work ratio was calculated as the ratio of positive energy returned to negative energy absorbed and is presented with other kinetics variables in Table S1, Supplemental Digital Content, http://links.lww.com/POI/A47. The step length was calculated using Visual 3D software as the sagittal plane distance between ankle markers at each heel strike and is presented with other stride parameter data in Table S2, Supplemental Digital Content, http://links.lww.com/POI/A47. After a spectral analysis of the force data, Matlab was used to design a second-order notch filter to remove some additional 46 Hz noise that was not fully attenuated by the low-pass filter. Power below the knee on the prosthetic side was calculated using a unified deformable (UD) model.[24] The power exerted on the COM by each leg was calculated using ground reaction force data, treadmill speed, and subject mass according to the individual limbs method.[8] Powers were then normalized to body mass.

Data were exported to MATLAB (R2017b, Mathworks, Inc.) for additional processing. For each stride, time of peak prosthesis push-off power was determined by identifying the percentage of the gait cycle at which peak UD power occurred. The last positive region of UD power was integrated to determine the positive work done on the COM by the prosthesis during push-off. The negative work performed by the intact limb on the COM, or intact limb collisional work, was determined by integrating the first negative region of COM power.[25]

Repeated measure ANOVAs were conducted in SPSS (IBM Corp., v22, Armonk, NY) to determine the effects of tuning parameter condition on time of peak prosthesis push-off power, positive prosthesis push-off work, and intact limb collisional work. We also performed ANOVAs on several other kinetic and stride parameter data. Effect size was estimated using partial eta-squared (η_p^2) calculated in SPSS with any value greater than 0.14 to be

Table 2. Main results for the effects of prosthesis tuning parameter settings on gait outcomes during level treadmill walking at 1.1 m/s.

	−B3	−B2	−B1	B0	+B1	+B2	+B3	Effect of Condition	
Time of peak prosthesis push-off power (% gait cycle)	60.5 ± 1.0	60.4 ± 0.6	60.1 ± 0.8	60.1 ± 1.5	59.9 ± 1.0	60.6 ± 2.3	59.6 ± 1.2	F = 0.876	P = 0.456
								η_p^2 = 0.226	1−β = 0.133
Positive prosthesis push-off work (J/kg)	0.182 ± 0.005[a]	0.169 ± 0.012[a]	0.153 ± 0.010	0.134 ± 0.012[a]	0.121 ± 0.025	0.105 ± 0.018	0.101 ± 0.015	F = 25.093	P = 0.002[b]
								η_p^2 = 0.893	1−β = 0.997
Intact limb collisional work (J/kg)	−0.094 ± 0.045	−0.090 ± 0.035	−0.090 ± 0.038	−0.086 ± 0.039	−0.097 ± 0.038	−0.094 ± 0.031	−0.087 ± 0.030	F = 0.471	P = 0.660
								η_p^2 = 0.136	1−β = 0.100

[a]*Conditions that were different from +B3 condition (P ≤ 0.05).*
[b]*Main effect difference (P ≤ 0.05).*

considered large. The observed power was calculated as 1−β. When a difference was found, we performed a post hoc analysis with alpha set to 0.05 with Bonferroni adjustment for multiple comparisons. A Shapiro-Wilk test showed that time of peak power ($W(28) = 0.949$, $P = 0.192$), positive push-off work ($W(28) = 0.946$, $P = 0.160$), and intact limb collisional work ($W(28) = 0.964$, $P = 0.428$) did not significantly depart from normality. A Greenhouse-Geisser correction was used if the assumption of sphericity was not met. The Pearson correlations were used to determine the relationship between the two dependent variables, prosthesis push-off work and intact leg collisional work, across all conditions. Because of our small sample size, within-subject correlations were also performed to assess the individual subject responses. To confirm whether power generated by the prosthetic device was being delivered to the COM during push-off, we ran a correlation analysis between the power produced by the prosthesis during push-off and the propulsion work of the amputated limb on the COM. All correlations were considered significant at an alpha level of .05.

Results

One subject relied heavily on the handrails during testing, resulting in unusable ground reaction force data and they were removed from analysis. The remaining four subjects using the robotic ankle prosthesis were retained for analysis and are presented here (mean age, 35 ± 13.6 years; mass, 80.9 ± 7.9 kg; height, 1.86 ± 0.08 m; and time since amputation, 13 ± 9.5 years).

Prosthetic foot tuning parameter settings had the greatest effect on the positive impulse during late stance, yet little effects on other joints (Figure 1). Step lengths, energy absorbed by the prosthetic foot, propulsion work performed by the amputated limb, propulsion work performed by the intact limb, and intact limb collision work were not affected by changing in the tuning parameters (Tables S1-S2, Supplemental Digital Content, http://links.lww.com/POI/A47). Tuning parameter settings had no significant effect on the time of peak prosthesis push-off power (Figures 1(a) and 2(a), Table 2, $F = 0.88$, $P = 0.446$). Although parameter settings did not influence the time of push-off power, there was a significant effect on the magnitude of prosthesis push-off work (Figures 1(a), 1(d) and 2(b), Table 2, $F = 25.09$, $P = 0.002$). The largest push-off power was generated in the −B3 setting and the smallest in the +B3 condition, although no differences were observed between any of the +B1-+B3 conditions.

We also observed that intact leg collisional work did not significantly vary across the different parameter settings (Figure 1(e), Table 2, $F = 0.471$, $P = 0.660$). In addition, there was no correlation between prosthesis push-off work and intact leg collisional work (Figure 3, $(R)^2 = 0.043$, $P = 0.108$). We also did not observe any obvious differences in the individual subject correlations (Figure S1, Supplemental Digital Content, http://links.lww.com/POI/A47). Power produced by the prosthesis during push-off was, however, positively correlated with propulsion work performed by the prosthetic limb on the COM ($R^2 = 0.174$, $P = 0.027$).

Discussion

Overloading of the intact limb is often believed to have a relatively simple solution: restore power to the area of the amputated limb that has lost power-producing musculature in an effort to better emulate a biological ankle.[26] Biomechanical models support this solution,[6] yet this study highlights that this problem may be more complicated. In this study, we characterized the effects of adjusting the parameter settings of a commercially available robotic prosthesis on ankle power delivery and loading of the intact limb. We observed that the tuning parameters of this powered prosthesis had an effect on the magnitude of prosthesis push-off work, but not on the time of peak push-off power (Figure 2). In addition, we hypothesized that the collisional work of the intact leading limb would be correlated with propulsive work performed by the amputated trailing limb. We observed no such correlation between positive prosthesis push-off work and intact leg collisional work.

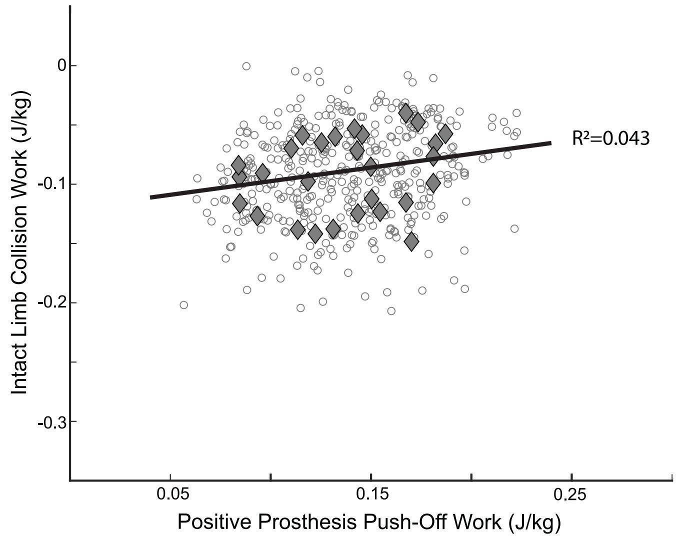

Figure 3. Correlation between intact limb collisional work and prosthesis limb push-off work. Positive work produced by the prosthesis during push-off did not correlate with the subsequent negative collision work in the intact limb ($r = 0.206$, $P = 0.108$). A regression was performed across tuning parameter conditions, with each diamond representing the average of 15 strides from each subject at each condition. Individual stride data (circles) are also shown to indicate the large within-subject variability with no discernible trend at the stride level.

Dynamic walking models predict that enhanced positive push-off work of the trailing prosthetic limb should result in a decrease in the collisional, or negative, work experienced by the leading intact limb at heel strike.[6] In contrast to what dynamic walking principles might predict, we did not observe a reduction of intact limb collisional work with increased prosthesis push-off work. Although we observed an 80% increase in the magnitude of increased push-off work generated by the prosthesis across conditions, we cannot completely rule out the possibility that this might not have been enough of a change to affect a change in collisional work. However, our findings do agree with recent studies using a tethered prosthesis emulator on individuals with intact limbs[15] and with lower extremity amputation,[16] which demonstrated no significant relationship between prosthesis leg push-off work and intact leg collisional work.

Although additional power was being provided by the prosthetic limb, we observed no resulting changes in intact leg collisional work suggesting that this additional power was not all being used to perform work on the COM. It is not clear why increases in prosthesis push-off work did not result in decreases in intact leg collisional work. It was not due to altered step kinematics because our data exhibit similar step lengths, as well as joint and leg power profiles to previous studies.[6,15,17] Positive work produced by the prosthesis during push-off was only weakly correlated with work done on the COM by the prosthetic limb ($R^2 = 0.174$). This indicates that much of the energy provided by the prosthesis did not perform work on the COM, which suggests that increased prosthesis power was dissipated somewhere between the prosthetic ankle and the COM. One possibility is that the knee or hip joints could be dissipating this energy from the prosthetic ankle. Caputo and Collins[15] observed a concomitant decrease in prosthesis-side hip power during terminal stance when prosthesis push-off work was increased, which may have accounted for the lack of collisional work decrease in the intact limb. Toney and Chang[27] observed that the knee joint also acts as a brake on the ankle joint during terminal stance, which can attenuate the effects of ankle joint power production during push-off. However, we observed no significant change in prosthesis limb hip or knee powers across conditions (Figure 1). We cannot rule out the possibility that prosthesis power may have been damped because of the dynamics of the residual limb/prosthetic socket interface.[28] Although we were not able to measure energy lost to relative socket motion, this is certainly an important topic for future studies.

Despite the small sample size of this study, our main significant finding of increased positive push-off work yielded a large effect size and statistical power (Table 2). The time of peak power at push-off and intact limb collisional work indicated a lower statistical power as did many other variables measured (e.g. Tables S1-S2, Supplemental Digital Content, http://links.lww.com/POI/A47), which increases the possibility of a type II error, and these results should be taken with some caution. Yet, all subjects showed little variability in their mean responses across conditions, which might suggest that the failure to reject the null may be correct in many of these cases. For example, the maximum changes in mean responses across all conditions for the time of peak push-off power were $<1\%$ of the gait cycle and only 0.011 J/kg for intact limb collisional work. Thus, it is likely that the same general conclusions about prosthesis function would have been drawn even with a greater sample size.

This is the first study to test the relationship between push-off prosthesis work and intact limb collisional work in a commercially available, untethered powered prosthesis. It is also the first to provide a functional characterization of the tuning parameters associated with this particular clinical device. The precise effect of the tuning adjustments on prosthesis push-off power remains unclear and could be nonlinear in nature. Because of our limited access with the device, we were not able to provide an extended acclimation time for our subjects. The added power may be more effectively used if subjects were given more time to acclimate to the device or were trained to optimize this metric specifically. To minimize this risk, we analyzed data from the last 15 steps in each condition for all subjects. Notably, one of our subjects had extensive experience with this prosthetic foot, and we did not observe qualitatively different results in those data. Having all subjects use the same powered foot may have resulted in a deviation from clinical protocol and may not represent a best-case clinical scenario. Although this could affect specific magnitudes of individual responses, we do not believe this should have affected the general trends observed.

Conclusion

Adjusting the tuning parameter settings on the Empower prosthesis influences the magnitude of push-off work, but not the timing of power delivery. This suggests that more explicit and accurate descriptions of tuning parameters could help researchers and clinicians make more informed choices. In contrast to dynamic walking model predictions, increasing positive prosthesis push-off power did not reduce intact leg collisional work for individuals with unilateral lower extremity amputations. It is unclear why subjects did not effectively use the added power provided by the robotic ankle prosthesis, but this points to a need for increased research on how to improve power utilization and individualized training to optimize prosthesis utility.

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Supplemental material

Supplemental material is available via direct URL citations in the HTML and PDF versions of this article on the website (www.POIjournal.org).

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