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Human-prosthesis coordination: A preliminary study exploring coordination with a powered ankle-foot prosthesis

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ABSTRACT

Background: Powered ankle-foot prostheses were developed to replicate the mechanics of the biological ankle by providing positive work during the push-off phase of gait. However, the benefits of powered prostheses on improving overall human gait efficiency (usually quantified by metabolic cost) have not been consistently shown. Here, we have focused on the mechanical work produced at the prosthetic ankle and its interaction with the amputee's movement.

Methods: Five unilateral transtibial amputees walked on a treadmill using 1) a powered ankle-foot prosthesis and 2) their daily passive device. We determined the net ankle work and ankle work loops on the prosthesis-side to quantify the efficiency of the human-prosthesis physical interaction. We further studied peak propulsion timing and the posture of the amputee's lower limb and prosthesis as indicators of the human-prosthesis coordination. Comparisons were made between the passive and powered prosthesis conditions for each participant.

Findings: The powered prosthesis did not consistently increase net ankle work compared to each participant's passive device. For participants that lacked efficiency in interacting with the powered prosthesis, we observed 1) early prosthesis-side peak propulsion timing ($\geq$ 4% earlier) and 2) a more vertical residual shank at the time of peak propulsion ($> 2^{\circ}$ more vertical) indicating that the human's limb movement and the prosthesis control during push-off were not well coordinated.

Interpretation: Results from this preliminary study highlight the need for future work to systematically quantify the coordination between the human and powered prosthesis and understand how such coordination at the joint level influences overall gait efficiency.

1. Introduction

Advanced powered ankle-foot prostheses have been designed to replicate biological ankle range of motion and generate positive ankle work during the propulsion phase of gait (Au et al., 2007). However, amputees' overall gait efficiency, usually quantified by metabolic cost, when using a powered ankle-foot prosthesis, compared to that when using their daily passive prosthesis, has been inconsistent in the literature (Müller et al., 2019). Some studies show powered prostheses reduce the metabolic cost of walking in amputees (Au et al., 2007; Herr and Grabowski, 2012), while other studies have reported no significant difference (Gardinier et al., 2018; Montgomery and Grabowski, 2018). Further still, one study found that the greatest metabolic cost reductions occurred when prosthesis ankle work at propulsion was double that of biological ankle work (Ingraham et al., 2018). Additionally, some studies have shown powered prostheses' ability to increase the affected limb's step-to-step transition work walking on level ground (Herr and Grabowski, 2012; Russell Esposito et al., 2015), but another study reported no significant differences between the two devices (Montgomery and Grabowski, 2018).

These inconsistencies pose a question to the field as to why the energy provided by the powered prosthesis has not consistently been transferred to the prosthesis users to improve gait efficiency. Among many speculations offered by various groups (Ferris et al., 2012; Grabowski and D'Andrea, 2013; Hill and Herr, 2013; Montgomery and Grabowski, 2018; Quesada et al., 2016), the loss of coordination between the motor control of the human's biomechanics and machine control of the prosthesis's motor is a particularly intriguing reason for the observed gait inefficiencies. The joints in the lower limbs are inherently coordinated through bi-articular musculature (e.g. medial and

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lateral gastrocnemius) and a single controller: the human central nervous system (Dietz, 2002); however, for transtibial amputees who wear a powered ankle-foot prosthesis, the prosthetic ankle joint is operated by a computer, while the residual shank and intact joints are controlled by the human. Even though the prosthesis is programmed to yield normative ankle mechanics during walking, if the amputee users do not coordinate their intact joints and body segments with the prosthesis's action, gait performance may not improve. However, this potential explanation remains mainly as a postulation. Little evidence based on actual tests involving individuals with transtibial amputations exist to support our contention that the human-prosthesis incoordination at the *local* joint level of the prosthesis can be one of the potential reasons for the lack of improvement in *global* measures (e.g. metabolic cost) of gait performance when walking with a powered prosthesis.

Therefore, as the first step to address the aforementioned question, the purpose of this preliminary study was to explore the interaction and coordination of a transtibial amputee's lower limb with a commercial powered ankle-foot prosthesis during level ground walking compared to the coordination with their prescribed passive prosthesis. New empirical evidence obtained in this study might lead to future systematic investigations to elucidate the sources that contribute to the individual differences of *global* gait efficiency among transtibial amputees while walking with a powered ankle-foot prosthesis.

2. Methods

2.1. Participants

Five individuals with unilateral transtibial amputation (age: 38 years (IQR 25-59); mass: 101 kg (IQR 89-112.5); and height: 1.85 m (IQR 1.84-1.875)) participated in this study (Table 1). Participants provided written, informed consent to participate in this study approved by the University of North Carolina at Chapel Hill Institutional Review Board. Participants were recruited from the local community who were conveniently available to participant in this study and the following inclusion/exclusion criteria were used: able to walk on a treadmill for at least 20 min without assistance, did not have any known comorbidities such as cardiovascular or neurological problems that may affect their performance in this study, and had sufficient limb clearance (> 11 in.) to wear the BiOM ankle-foot prosthesis (BiOM T2, Ankle, BionX Medical Technologies Inc., Cambridge, MA, USA).

2.2. Experimental protocol

We measured gait kinematics and kinetics as participants walked on a treadmill. Gait kinematics were captured by 43 light-retroreflective markers placed on the participant's trunk, pelvis, and bilaterally on the thighs, shanks, and feet. Markers were placed on the acromia, iliac crests, greater trochanters medial and lateral femoral epicondyles, medial and lateral malleoli, first and fifth metatarsals, and calcanea to define the torso, pelvis, thigh, shank, and foot segments, respectively, in addition to tracking markers. Markers on the feet were placed on the outside of the shoes over the bony landmarks on the intact limb and position-matched on the prosthetic limb. The marker positions were captured with an 8-camera motion capture system (VICON, Oxford, UK), sampled at 100 Hz. Bilateral ground reaction forces (GRFs) were recorded by an instrumented, split-belt treadmill (Bertec Corp. Columbus, OH, USA), sampled at 1000 Hz. Both measurements were synchronized.

Participants made a single visit to the lab, in which they walked with both the BiOM powered ankle-foot prosthesis (BiOM) and their prescribed passive prosthesis (Passive) (Fig. 1). Both devices were aligned and fit by a certified prosthetist. The participants walked on the treadmill with their daily passive prosthesis first in order to acclimate to treadmill walking and determine self-selected walking speed. Self-selected walking speed was determined from each participant's preference similar to a previously described method (Plotnik et al., 2015). Walking speed was measured with each participant's passive device and this speed was used for both conditions. Then, a two -minute walking trial was collected while the participants wore their passive prosthesis. Next, participants were fitted with the BiOM and the control parameters were adjusted while the participant walked on level ground following the tuning procedure specified by the manufacturer. The participant then acclimated to walking with the device (described below). After acclimation, a second two-minute walking trial was collected while the participant walked with the BiOM on the treadmill.

To acclimate to wearing the BiOM, the prosthesis was first heuristically tuned over-ground using the software provided by the BiOM manufacturer and following the tuning procedure specified in the BiOM manual (BiOM ® T2 Ankle Instructions for Use. (accessed), n.d.). After tuning, participants walked at their self-selected treadmill speed for 10-min bouts with at least two minutes of rest between bouts until completion criteria were met. Note that during tuning and the acclimation period, we monitored the net ankle work displayed on the commercial software to ensure the value fell within the desired normative work range. Completion criteria for acclimation was defined as 1) completing at least two bouts, 2) no use of handrails during walking in each bout, and 3) reaching steady-state step length symmetry, where the mean step length symmetry of the last minute of the 10 min session was within the 95% confidence interval of the mean of the first 30 s of the 10 min session. Additionally, participants verbally confirmed they were comfortable walking with the device at the end of each bout. All participants met these criteria after twenty minutes of acclimation which was similar to previous studies (Gardinier et al., 2018; Ingraham et al., 2018). In addition, clinical tuning and fitting of a powered ankle-foot prosthesis usually takes one to two clinical visits with a short-term acclimation, comparable with the duration used in this study. It is noted that additional acclimation time may be needed, but it is currently unknown how long it takes to acclimate to a new prosthesis (Wanamaker et al., 2017).

2.3. Data processing

To capture steady-state walking behavior, twenty consecutive steps were selected for analysis during the last minute of the walking trial for both the BiOM and Passive conditions. Commercially-available data

Table 1
Summary of participants. All participants wore a passive prosthesis as their everyday device.

Participant ID	Sex	Height (m)	Mass (kg)	Self-Selected Walking Speed (m/s)	Tuned Self-Selected Power	Age (years)	Time Since Amputation (years)	Socket Suspension	Reason for Amputation
TB01	Male	1.85	122	1.10	40	29	5	Vacuum	Trauma
TB02	Male	1.87	101	1.10	36	64	16	Pin/Lock	Trauma
TB03	Male	1.85	97	1.00	38	29	11	Suction	Trauma
TB04	Male	1.88	103	1.20	46	54	5	Vacuum	Cancer
TB05	Female	1.83	81	1.15	32	21	6	Suction	Trauma
Median (IQR)		1.85 (1.84–1.875)	101 (89–112.5)	1.10 (1.05–1.175)	38 (34–43)	29 (25–59)	6 (5–13.5)		

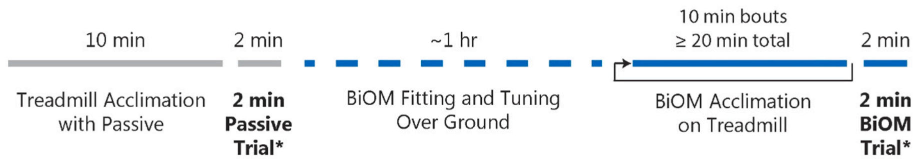

Fig. 1. Explanation of methods. Participants were first acclimated to treadmill walking with their passive prosthesis and a two-minute trial was collected. Next, participants were fit with the BiOM powered ankle-foot prosthesis and tuning was performed over ground. After tuning, participants were acclimated to the device on the treadmill in 10 min bouts until completion criteria were achieved. After this, a second two-minute trial was collected for comparison against the passive prosthesis.

analysis software (Visual 3D, C-Motion, Inc., Germantown, MD, USA) was used to process the data. GRFs were smoothed by a 4th order Butterworth filter with a cutoff frequency of 25 Hz; the marker positions were low-pass filtered by a 4th order Butterworth filter with a cutoff frequency of 6 Hz. GRFs of each limb and a threshold of 20 N were used to determine heel-strike and toe-off gait events.

Prosthesis ankle work during gait reflects the ability of the prosthesis to assist an amputee's walking. Therefore, we defined the participant's efficiency with interacting with the prosthesis as the net work of the prosthesis ankle and work loops (ankle angle vs. ankle moment). We additionally quantified coordination with spatiotemporal parameters of gait, such as timing of propulsion as well as the position of the limb during propulsion. Peak propulsion timing was defined as the time of the peak anteriorly-directed GRF during the late stance phase, normalized to stance time (i.e. duration from ipsilateral heel strike to toe-off). Shank angle was defined as the angle of the shank (i.e. segment connecting the ankle and knee joint centers) relative to the vertical axis of the lab. These measures were chosen since they are local to the prosthesis and could therefore be more easily measured in a clinic for future tuning and training purposes.

Comparisons were made within each participant between the two devices (Passive and BiOM) in order to understand how the human-prosthesis coordination is different between the two prosthetic devices on an individual basis. We also made comparisons across participants to explore differences between users wearing the powered prosthesis. Nonparametric *t*-tests (Mann-Whitney *U* tests) and a nonparametric one-way ANOVA with Tukey HSD comparisons (Kruskal-Wallis with Steel-Dwass method) were used (alpha = 0.05), which are commonly used for small sample sizes and non-independent samples (i.e. consecutive walking strides).

3. Results

Fig. 2 shows the net prosthetic ankle work loops averaged over 10 steps. Comparing the BiOM and Passive conditions, all five participants generated negative net ankle work with their passive prosthesis. With the BiOM, however, three participants produced counter-clockwise ankle work loops (i.e. positive net ankle work) with the powered prosthesis, whereas two participants had clockwise work loops (i.e. negative net ankle work). Most participants in this study, with the exception of TB04, did not experience a notable increase in ankle work or in composite lower limb work on the prosthesis side (see details in the Appendix). To make comparisons easy, we ranked participants in order of net ankle work for all results.

The timing of peak propulsion varied across participants for both prosthesis conditions (Fig. 3). TB05 and TB01 had a significantly earlier peak propulsion time on the prosthetic limb when wearing the BiOM than that when wearing the Passive prosthesis. TB05's peak propulsion time was at 87% of stance phase with the BiOM, but 91% with the Passive ($X_{(1)}^2 = 13.523, p = 0.0002$). TB01's peak propulsion time was 86% with the BiOM, but 91% with the Passive ($X_{(1)}^2 = 15.759$, $p < 0.0001$). Conversely, the remaining participants (TB02, TB03, TB04) had peak propulsion timing that was not different between the devices. Also of note, TB04 (the participant with the greatest amount of net work) had peak propulsion timing that was later in the stance phase than that of all other participants. For all participants, the timing of peak propulsion of the intact limb did not change between the BiOM and Passive conditions and was similar across participants.

Our examination of limb position revealed that the prosthesis-side shank was oriented more vertically at peak propulsion with the powered prosthesis than that with the passive prosthesis for four participants (Fig. 4). When comparing across participants, TB02, TB03, TB05, and TB01 were not significantly different from one another; however, the shank angle of TB04 (participant with greatest amount of net work), was significantly greater (directed more forward) compared to the other participants, which was demonstrated in Table 2. Additionally, there was no difference in shank angle at peak propulsion for the intact limb between the BiOM and Passive conditions and was similar across participants.

4. Discussion

The main results of this study showed that even at the *local* joint level, the powered prosthesis did not consistently improve joint mechanics during walking, compared to each individual's passive prosthesis. In addition, we found that participants who did not increase net ankle work with the powered prosthesis tended to have earlier peak propulsion timing with the powered prosthesis compared to their daily passive prosthesis and have a more vertical shank position at the time of peak propulsion. These concurrent observations suggest that the net ankle work was influenced not only by the active torque provided by the prosthesis during push-off, but also by other confounding factors, such as the coordination between the human and their prosthesis when dynamically interacting with the ground (environment) during walking. When the human's motor control of their lower limb and the control of the prosthesis action were not coordinated, the ability of the active prosthesis to empower the human's walking was diminished. More specifically, the action of the BiOM for producing propulsive torque was probably too early for some of the users, and the residual shank, controlled by these amputee users, had not progressed to the appropriate position yet. This would cause the active propulsive torque produced by the electromechanical motor in the prosthesis to direct the limb more vertically (upward), instead of anteriorly (forward).

One contribution of this study was that, although preliminary, we provided empirical evidence of human-prosthesis incoordination that might be associated with the decreased performance of a powered ankle-foot prosthesis in individuals with transtibial amputations. Many previous studies have only speculated the potential contribution of misaligned timing of the action of the powered prosthesis on the global gait efficiency without showing evidence (Ferris et al., 2012; Grabowski and D'Andrea, 2013; Hill and Herr, 2013; Montgomery and Grabowski, 2018; Quesada et al., 2016). One study did specifically investigate the optimal timing of powered propulsion in a powered ankle prosthesis in order to minimize the metabolic cost of walking (Malcolm et al., 2015). However, the study was conducted with a prosthesis emulator on able-bodied individuals, and research has shown that results from studies with able-bodied individuals wearing an emulator do not necessarily translate to an amputee population due to significantly different body dynamics and motor control capability (Quesada et al., 2016). Our preliminary evidence highlights the importance for a further

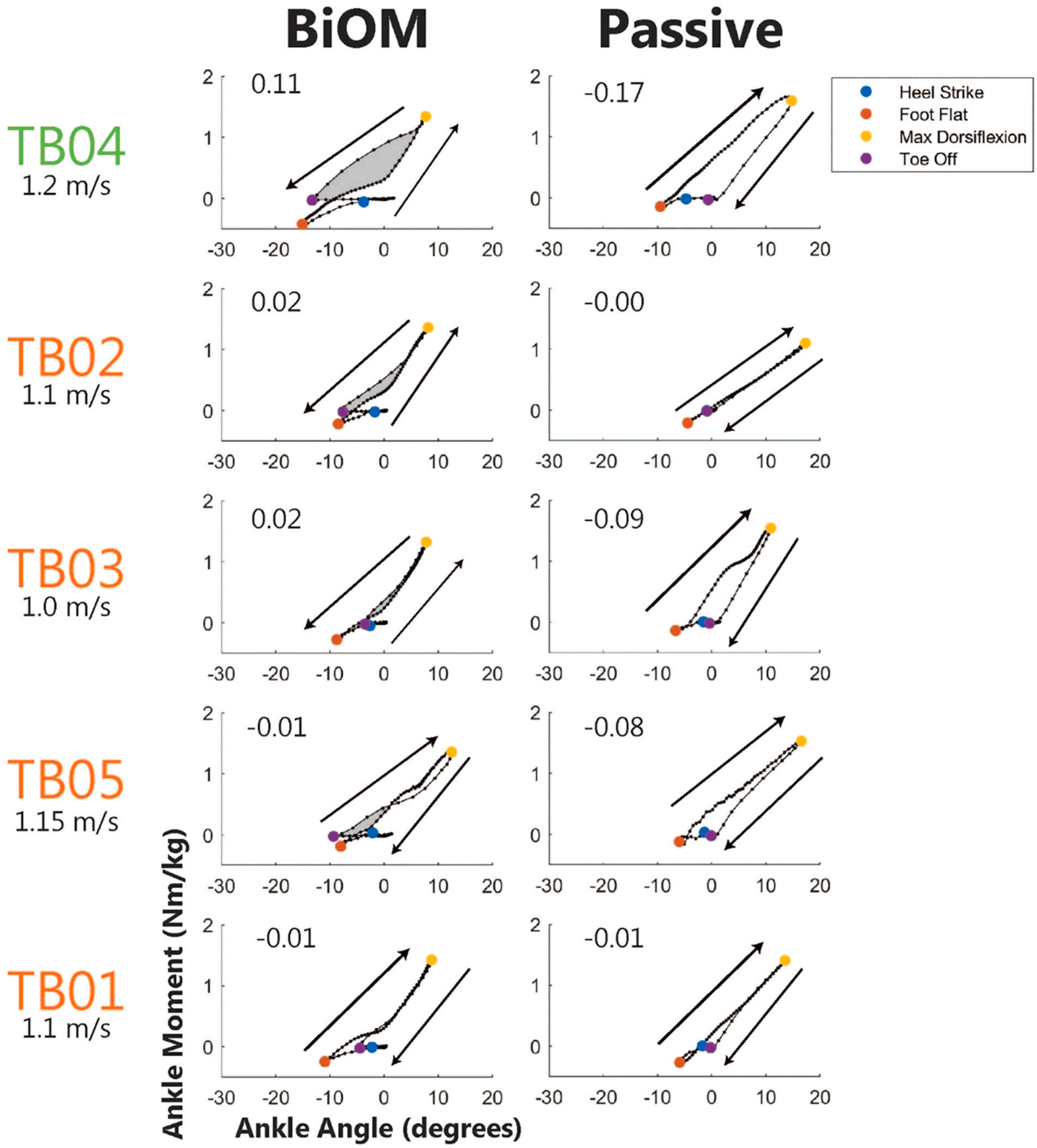

Fig. 2. Ankle work loops for the BiOM and Passive prostheses. Ankle moment (normalized to body weight) is plotted versus ankle angle for each participant. Ankle moment and ankle angle were averaged over 10 steps. Positive work is indicated by counterclockwise arrows and is shaded in gray. Negative work is indicated by clockwise arrows and is not shaded. Net ankle work is displayed in the upper left corner of each plot. The colored point characters correspond to gait events: heel strike (blue), foot flat (red), maximum dorsiflexion (yellow), and toe off (purple). Each participant's self-selected walking speed is provided with the participant labels for reference. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

systematical investigation of the human-prosthesis coordination and its association with the joint work as well as overall *global* walking efficiency.

Based on our study, we advocate for the evaluation of the efficacy of powered prostheses at various levels, from the *local* joint biomechanics to *global* energy expenditure. In the existing literature, global measures such as metabolic cost have been used extensively to evaluate and optimize powered prostheses (Felt et al., 2015; Kim et al., 2017). Metabolic cost is a great indicator of overall gait performance – it captures how efficiently someone is moving based on the assumption that optimal coordination could minimize energy cost. However, if there are inefficiencies in the human-robot coordination, metabolic cost might be insensitive to the action of the powered prosthesis. This may explain the inconsistencies in the current literature regarding powered ankle-foot prostheses.

Another potential implication of our study results is to expand the engineering framework in powered prostheses from “human-in-the-loop optimization” to “human-robot coordination/co-adaptation” for optimal human-prosthesis system performance. Our study showed that optimizing or tuning the prosthesis alone was probably not enough to optimize the gait of amputees. For example, the clinical procedure for tuning and fitting a powered prosthesis usually only takes a couple of

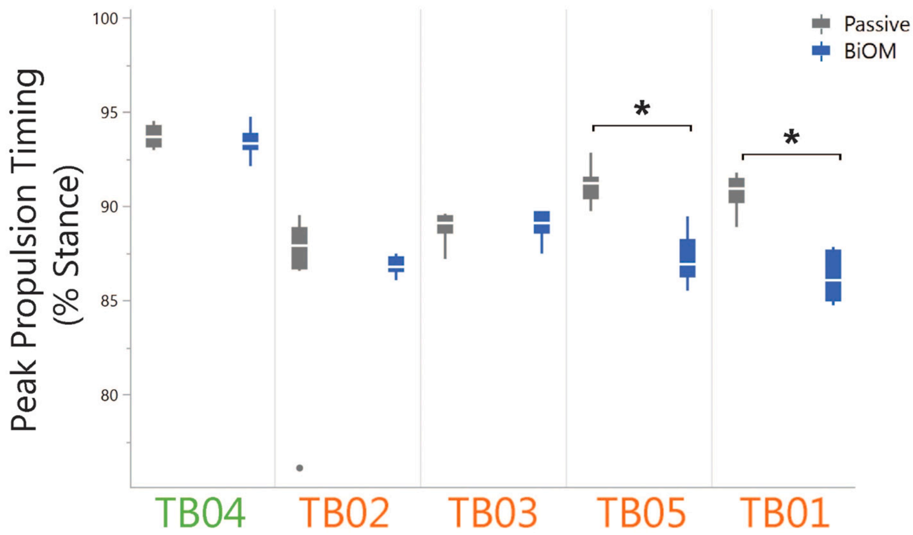

Fig. 3. Boxplots of timing of peak propulsion with their prosthesis as a percentage of stance time for each participant over 10 steps. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Shaded bands indicate the timing of peak propulsion on the intact side. Outliers are defined as 1.5 times the interquartile range from Q1 and Q3. Asterisks (*) indicate significant differences between the two devices ($p < 0.05$). TB01 and TB05 had earlier push-off when wearing the BiOM compared to their passive prosthesis. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

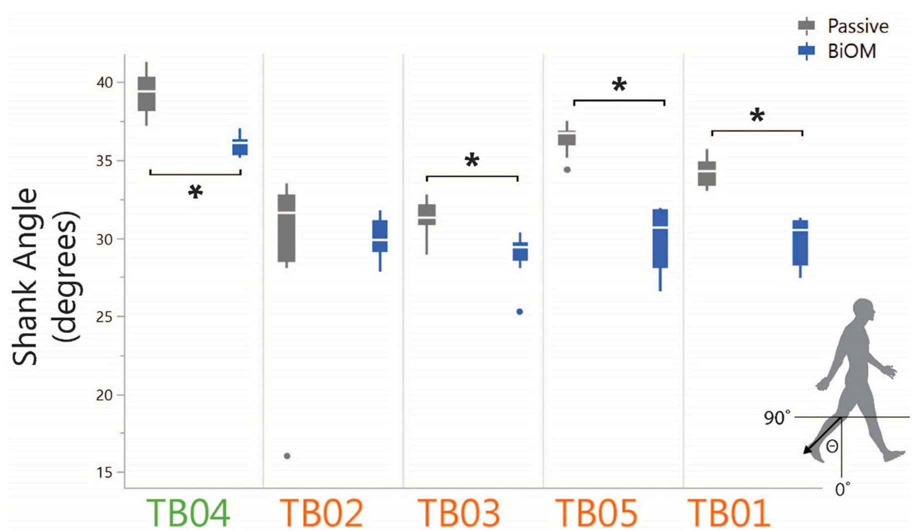

Fig. 4. Boxplots of prosthesis side shank angle as a percentage of stance time for each participant over 10 steps. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Shaded bands indicate the timing of peak propulsion on the intact side. Outliers are defined as 1.5 times the interquartile range from Q1 and Q3. Asterisks (*) indicate significant differences between the two devices ($p < 0.05$). Shank angle was defined as the angle of the shank relative to the vertical axis as detailed in the image in the bottom right corner. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 2
Comparison of shank angle at peak propulsion while wearing the BiOM between TB04 and other participants. Comparisons between all other participants were not different and are not shown. Asterisks (*) indicate significant differences.

Participant ID	Mean (Standard Error)	*P*-value compared to TB04
TB04	35.91 (0.18)	–
TB02	29.88 (0.37)	0.0008*
TB03	29.10 (0.42)	0.0008*
TB05	30.04 (0.63)	0.0012*
TB01	29.93 (0.42)	0.0008*

hours, assuming that the human automatically coordinates with the powered prosthesis' action and makes use of the active power from the device. However, our results suggest that this assumption is invalid; perhaps the amputee user's gait behavior may also need to be modified in order to produce seamless human-prosthesis coordination. Therefore, a novel framework that can enhance human-prosthesis coordination is necessary. This framework first requires measurements of coordination. This study suggested that peak propulsion timing and residual shank movement can be good indicators for quantifying the human-prosthesis coordination in walking. Both measurements can be obtained from the intrinsic sensors mounted on the powered ankle-foot prosthesis, making the framework easy to be adopted in clinics. Beside prosthesis tuning/optimization, the framework should also consider directed training, such as biofeedback (Brandt et al., 2019), that can train/modify the human's gait behavior in order to make the best use of modern powered prostheses. This could be a future direction for the field of prosthetics gait training.

Finally, it is noted that these results are not without limitations. First, the sample size was small, and the study results are therefore preliminary. Further research efforts are needed to confirm our observations on more persons with transtibial amputations. Next, in this study, we controlled walking speed for both the Passive and BiOM conditions using amputees' self-selected walking speed when using their daily passive prosthesis. The same speed was chosen to reduce possible confounding effects. However, we do think that it is interesting to investigate the effect of speed on human-machine coordination. Future work should evaluate these measures of coordination at different walking speeds to investigate how coordination changes across speeds. In addition, all participants met our criteria after two bouts of acclimation to get comfortable with the powered prosthesis after approximately an hour of device alignment and tuning. Currently, the literature varies widely on acclimation time and training - studies have reported as little as twenty minutes to three weeks and there are no set criteria to end acclimation. There is currently no standard for assessing training time or how to properly instruct amputees to best use a new device and it is currently unknown how long it takes to acclimate to a new prosthesis (Wanamaker et al., 2017). Future work is needed to determine how long acclimation and training should be when using a powered prosthesis.

5. Conclusion

This study aimed to conduct a preliminary study on understanding how individuals with transtibial amputations coordinate with a powered ankle-foot prosthesis during treadmill walking. Results from this study showed that even at the *local* joint level of the prosthesis ankle, the ability to produce positive net ankle work was not consistently shown across the amputee participants. For participants that did not improve the net ankle work with the powered prosthesis, we concurrently observed earlier peak propulsion time and a more vertical shank at the time of peak propulsion relative to their passive prosthesis indicating an incoordination between the human and their prosthesis. These preliminary study results highlight the need to systematically investigate the human-prosthesis coordination at the *local* joint and its association with the *global* measures of gait performance in the future. In addition, studying human-prosthesis coordination/co-adaptation could also improve how powered prostheses are tuned/optimized as well as how amputees are trained.

Declaration of Competing Interest

The authors declare that they have no competing interests.

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Appendix. Table 3 summarizes the net work of ankle, knee, hip, and composite limb work on the prosthesis side as supplementary data

Table 3
Net Work of the ankle, knee, hip, and composite limb work (A + K + H) for the prosthesis side for both the BiOM and Passive conditions. Values are averaged over 10 steps. A: ankle; K: knee; H: hip.

Participant	Joint	Prosthesis Side	
		BiOM	Passive
		Net Work (J/kg)	Net Work (J/kg)
TB04	**Ankle**	**0.11**	**−0.17**
	Knee	**−0.23**	**−0.23**
	Hip	**0.34**	**0.43**
	A + K + H	**0.22**	**0.02**
TB02	**Ankle**	**0.02**	**0.00**
	Knee	**−0.13**	**−0.12**
	Hip	**0.43**	**0.42**
	A + K + H	**0.32**	**0.30**
TB03	**Ankle**	**0.02**	**−0.09**
	Knee	**−0.14**	**−0.17**
	Hip	**0.18**	**0.20**
	A + K + H	**0.05**	**−0.06**
TB05	**Ankle**	**−0.01**	**−0.08**
	Knee	**−0.18**	**−0.21**
	Hip	**0.22**	**0.33**
	A + K + H	**0.03**	**0.04**
TB01	**Ankle**	**−0.01**	**−0.01**
	Knee	**−0.24**	**−0.22**
	Hip	**0.04**	**0.19**
	A + K + H	**−0.21**	**−0.04**

References

Au, S.K., Herr, H., Weber, J., Martinez-Villalpando, E.C., 2007. Powered ankle-foot prosthesis for the improvement of amputee ambulation. Conf. Proc. IEEE Eng. Med. Biol. Soc. 3020–3026. 2007. https://www.ncbi.nlm.nih.gov/pubmed/18002631.
BiOM ® T2 Ankle Instructions for Use. (accessed).
Brandt, A., Riddick, W., Stallrich, J., Lewek, M., Huang, H.H., 2019. Effects of extended powered knee prosthesis stance time via visual feedback on gait symmetry of individuals with unilateral amputation: a preliminary study. J. Neuroeng. Rehabil. 16 (1), 112. https://www.ncbi.nlm.nih.gov/pubmed/31511010.
Dietz, V., 2002. Do human bipeds use quadrupedal coordination? Trends Neurosci. 25 (9), 462–467. https://www.ncbi.nlm.nih.gov/pubmed/12183207.
Felt, W., Selinger, J.C., Donelan, J.M., Remy, C.D., 2015. "Body-in-the-loop": optimizing device parameters using measures of instantaneous energetic cost. PLoS One 10 (8), e0135342. https://www.ncbi.nlm.nih.gov/pubmed/26288361.
Ferris, A.E., Aldridge, J.M., Rabago, C.A., Wilken, J.M., 2012. Evaluation of a powered ankle-foot prosthetic system during walking. Arch. Phys. Med. Rehabil. 93 (11), 1911–1918. https://www.ncbi.nlm.nih.gov/pubmed/22732369.
Gardinier, E.S., Kelly, B.M., Wensman, J., Gates, D.H., 2018. A controlled clinical trial of a clinically-tuned powered ankle prosthesis in people with transtibial amputation. Clin. Rehabil. 32 (3), 319–329. https://www.ncbi.nlm.nih.gov/pubmed/28750586.
Grabowski, A.M., D'Andrea, S., 2013. Effects of a powered ankle-foot prosthesis on kinetic loading of the unaffected leg during level-ground walking. J. Neuroeng Rehabil. 10, 49. https://www.ncbi.nlm.nih.gov/pubmed/23758860.
Herr, H.M., Grabowski, A.M., 2012. Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation. Proc. Biol. Sci. 279 (1728), 457–464. https://www.ncbi.nlm.nih.gov/pubmed/21752817.
Hill, D., Herr, H., 2013. Effects of a powered ankle-foot prosthesis on kinetic loading of the contralateral limb: a case series. IEEE Int. Conf. Rehabil. Robot (2013) 6650375. https://www.ncbi.nlm.nih.gov/pubmed/24187194.
Ingraham, K.A., Choi, H., Gardinier, E.S., Remy, C.D., Gates, D.H., 2018. Choosing appropriate prosthetic ankle work to reduce the metabolic cost of individuals with transtibial amputation. Sci. Rep. 8 (1), 15303. https://www.ncbi.nlm.nih.gov/pubmed/30333504.
Kim, M., Ding, Y., Malcolm, P., Speeckaert, J., Siviy, C.J., Walsh, C.J., et al., 2017. Human-in-the-loop Bayesian optimization of wearable device parameters. PLoS One 12 (9), e0184054. https://www.ncbi.nlm.nih.gov/pubmed/28926613.
Malcolm, P., Quesada, R.E., Caputo, J.M., Collins, S.H., 2015. The influence of push-off timing in a robotic ankle-foot prosthesis on the energetics and mechanics of walking. J. Neuroeng. Rehabil. 12, 21. https://www.ncbi.nlm.nih.gov/pubmed/25889201.
Montgomery, J.R., Grabowski, A.M., 2018. Use of a powered ankle-foot prosthesis

reduces the metabolic cost of uphill walking and improves leg work symmetry in people with transtibial amputations. J. R. Soc. Interface 15 (145). https://www.ncbi.nlm.nih.gov/pubmed/30158189.

Müller, R., Tronicke, L., Abel, R., Lechler, K., 2019. Prosthetic push-off power in transtibial amputee level ground walking: a systematic review. PLoS One 14 (11), e0225032. https://www.ncbi.nlm.nih.gov/pubmed/31743353.

Plotnik, M., Azrad, T., Bondi, M., Bahat, Y., Gimmon, Y., Zeilig, G., et al., 2015. Self-selected gait speed–over ground versus self-paced treadmill walking, a solution for a paradox. J. neuroeng. rehabil. 12, 20. https://www.ncbi.nlm.nih.gov/pubmed/25881130.

Quesada, R.E., Caputo, J.M., Collins, S.H., 2016. Increasing ankle push-off work with a powered prosthesis does not necessarily reduce metabolic rate for transtibial amputees. J. Biomech. 49 (14), 3452–3459. https://www.ncbi.nlm.nih.gov/pubmed/27702444.

Russell Esposito, E., Aldridge Whitehead, J.M., Wilken, J.M., 2015. Step-to-Step Transition Work During Level and Inclined Walking Using Passive and Powered Ankle-Foot Prostheses. https://doi.org/10.1177/0309364614564021.

Wanamaker, A.B., Andridge, R.R., Chaudhari, A.M., 2017. When to biomechanically examine a lower-limb amputee: a systematic review of accommodation times. Prosthetics Orthot. Int. 41 (5), 431–445. https://www.ncbi.nlm.nih.gov/pubmed/28946826.