ABSTRACT

FYLSTRA, BRETTA LYNN. Understanding Human-Prosthesis Coordination to Improve Performance for Individuals with Lower-Limb Amputation (Under the direction of Dr. He (Helen) Huang).

Traditionally, individuals with lower-limb loss are prescribed mechanically passive prostheses that result in increased energy expenditure, reduced balance, and increased incidence of injury to the intact joints. To combat these limitations, powered prostheses have been developed to replicate the behavior of the biological joints and restore “normative” joint motion. However, the benefit of these powered prosthesis has not been consistently shown in the literature – there are now two distinct control systems: the human control system and prosthesis control system that must work together. Therefore, research has investigated the prosthesis control system by customizing (aka tuning) the control parameters of the powered prosthesis to adapt to the user’s behavior. This research, however, has largely neglected the influence of the human’s control system on its ability to also influence the prosthesis control. Hence, studying the **coordination** of the human and prosthesis is vital to maximize performance. This work aims to quantify human-prosthesis coordination and explore the relationship between adaptive control of the prosthesis and changing behavior of the prosthesis user.

In Chapter 2, I present a preliminary study of possible reasons why individuals walking with a powered ankle-foot prosthesis do not consistently observe benefits in overall performance. Results from this study indicate that timing of powered push-off assistance may be too early for some users resulting in the power being directed vertically instead of anteriorly to assist with propelling the body forward. This work quantified coordination with measurements *local* to the human-prosthesis interface opposed to using *global* measurements that measure the total lumped effect of the device as well as the human’s response. Therefore, it is important for future work to

further understand exactly how the prosthesis is coordinating with the user by investigating measurements closer to the prosthesis and then working to optimize larger *global* variables.

In Chapter 3, I examined the relationship between autonomous prosthesis tuning and visual feedback to understand the effect of the device on the user and the user on the device. In this case, the prosthesis tuning goal was to achieve “normative” knee kinematics and the user goal was to increase prosthesis-side stance time. Results from this study found that adding visual feedback with autonomous tuning had inconsistent benefits to reducing the time to tune the prosthesis, and the tuning did not significantly impact the user’s ability to complete the feedback task. After tuning the prosthesis parameters while the user was in a more favorable gait pattern (i.e. extended prosthesis-side stance time), the user maintained extended stance time and better stance time symmetry in the short term; however, this benefit was not retained over time. This study offered a new framework of combining tuning with feedback to better understand the coordination and interaction between the two control systems and future work is needed to explore additional goals for the human-prosthesis system.

Finally, in Chapter 4, I further investigated the effect of changing prosthesis control parameters on both *local* and *global* variables. Results from this preliminary study revealed that differences can be observed in both local and global variables during tuning; however, the *local* variable had a clear relationship with the control parameters themselves while changes in the *global* variable cannot be explained by the control parameters alone. The intact joints more proximal to the user may also change their behavior affecting the propagation of the changes of the knee control on the *global* measurements. Again, more analysis is needed to fully understand the effects of the changing control on the user to best select an objective function to maximize user performance.

Overall, this dissertation aims to better characterize the relationship between the human and prosthesis systems.

Understanding Human-Prosthesis Coordination to Improve Performance for Individuals with Lower-Limb Amputation

by
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DEDICATION

To my parents, for instilling in me my drive, determination, and love of learning (including fractions). And to the memory of my Pop-Pop, who would always ask how school was going – I finally did it!

BIOGRAPHY

Bretta Fylstra received her Bachelor's degree in Biomedical Engineering from the University of Delaware in 2016. As an undergraduate student, Bretta conducted research under the direction of Dr. Elisa Arch studying the biomechanics of substituting ankle-foot orthosis footplate curvature for limited ankle range of motion. At the University of Delaware, she also had the opportunity to complete a clinical internship with Independence Prosthetics & Orthotics as well as co-found the Prosthetics & Orthotics Club and serve as the public relations chair in the Assistive Medical Technologies club. In 2016, Bretta was awarded a National Science Foundation Graduate Research Fellowship and began pursuing her PhD in the Joint Department of Biomedical Engineering between NC State University and UNC Chapel Hill. She began researching the coordination between amputee users and their prosthetic device under the direction of Dr. He (Helen) Huang in the Neuromuscular Rehabilitation Engineering Lab (NREL) and served as both an undergraduate mentor and peer leader. Bretta plans to continue her passion for rehabilitation engineering through leadership, clinical experience, and engineering expertise.

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CHAPTER 1: INTRODUCTION

1.1 The Human: Amputee Gait

There are over 1.6 million persons affected by limb loss in the United States and this number is estimated to more than double to 3.6 million by the year 2050, assuming no change in current healthcare practices [1]. Of these individuals, 65% are affected by lower limb amputations and the most common cause of amputation is dysvascular disease from diabetes mellitus [1]. The average lifetime healthcare cost for persons with limb loss is over $500,000 with roughly $90,000 spent in the first two years after amputation [2]. Therefore, there is a great need to understand and develop prosthesis technology to better treat this growing population and reduce healthcare costs.

A primary goal of walking is the forward advancement of the body, which is largely achieved during the propulsion phase of gait which generates a large burst of power to propel the lower limb into swing phase and transfer one's body weight to the contralateral limb [3, 4]. However, this function is impaired in lower limb amputees. Traditionally, lower-limb amputees are prescribed a mechanically-passive, prosthesis with fixed mechanics, which are unable to generate positive work during the propulsion phase of gait [5]. This leads to diminished mechanical contributions from the amputee's amputated limb toward forward movement during walking [6], increased compensations from the intact limb [7], increased metabolic cost [8], increased risk of falling [9, 10], and slower preferred walking speeds [11] compared to able-bodied individuals.

In addition, individuals with lower limb amputation vary significantly in their physical capabilities, compared to able-bodied individuals, due to factors such as prosthesis alignment

[12], ambulation ability (K-Level) [13], quality or type of surgical procedure [14], and other comorbidities such as trauma to the intact limb [15] and neuropathy [16]. These characteristics can influence the amputee's response to their prosthesis [17, 18]. Considering the large variations in physical capabilities in the amputee population, averaging data across all participants could mask the important individual responses [19]. Therefore, it is important to note that analysis of amputee gait on an individual level is necessary in addition to group averages to better understand performance of various prosthetic devices.

1.2 The Prosthesis: Powered Ankle-Foot and Powered Knee Prostheses

To overcome the limitations of passive prostheses, powered prostheses have been designed to replicate biological lower-limb joints. For the ankle, prostheses have been developed to provide net positive work (integration of power provided per stride) during the propulsion phase of gait [20-24]. Unlike passive prostheses, these powered ankle-foot prostheses are able to replicate biological ankle range of motion and generate positive ankle work during the propulsion phase of gait [23]. Similarly, for the knee, powered prostheses have been developed [25-34] and enable individuals with an amputation to navigate uneven terrain such as ramps and stairs [33-37] and improve symmetry in muscle activation [38] and temporal-spatial gait symmetry [39, 40].

However, amputees' gait performance with powered prostheses has been inconsistent in the literature. Some studies show powered ankle-foot prostheses, compared to passive devices, reduce the metabolic cost of walking in individuals with an amputation [23, 41], while other studies have reported no significant difference [17, 42]. Similarly, other studies with powered knee prostheses have not consistently been able to improve performance in terms of reduced loads on the intact limb and patient preference for the device [43, 44]. Therefore, the question arises: how do we improve performance of the human-prosthesis system?

1.3 Human-Prosthesis Coordination: Tuning, Training, and Rehabilitation

In individuals without an amputation, joints in the lower limbs are inherently coordinated through by a single controller: the human nervous system [45]. Additionally, there is bi-articular musculature to link the limb segments and provide sensory feedback. In contrast, for individuals who wear a powered prosthesis, there are now two distinct controllers: the human's control system and the control system of the powered prosthesis. The only link between these two systems is at the socket-residual limb interface. To restore the coordination between the prosthesis user and their robotic prosthesis, the device itself must be controlled properly and the user must learn to adapt to the device [46].

For the prosthesis control, it is most common to use "normative" kinematics/kinetics as the goal to replicate the action of the biological limb in individuals without an amputation [20, 47-49]. To do so, the device must be able to accurately sense the gait phase (e.g. stance vs. swing phase) [20, 47, 50] and set the control for each phase typically with the use of a finite-state machine. However, due to the large variations in the prosthesis users wearing these devices, it is necessary to personalize the control through tuning the control parameters. Clinically, this is completed manually – a prosthetist will heuristically update the parameters as the patient walks in a straight line across the clinic. This process can be long and relies on communication between the prosthesis user and the clinician to arrive at a set of parameters that optimize performance. Recent advances have begun to automate this process to reduce the burden on clinicians and can tune high-dimensional control spaces. Methods such as human-in-the-loop optimization (HILO)

[51-53] and reinforcement learning (RL) [54-56] have demonstrated the ability to tune the parameters, but the resulting performance has not been consistently maximized [51]. Therefore, the tuning of the prosthesis parameters alone may not optimize performance – the user has the ability to affect the prosthesis control.

When investigating research into the human-side of the human-prosthesis system, little evidence exists quantifying the human's effect on the prosthesis control. However, one study has pointed out that the tuning algorithm could not precisely tune the timing features of the knee angle profile goal due to the user's ability to control the timing of heel strike and push-off [54]. Further, as mentioned above, the bulk of the research involving powered prosthetic devices centers around engineers designing the control and not around instructing the prosthesis user how best to coordinate or acclimate to the control. There is no standard for how to train users how to use their device and know when device acclimation is complete [57]. Without instructions, individuals may carry-over their strategies learned with their passive prosthesis [58] that have inherent limitations such as limited range of motion and powered push-off. Thus, incorporating new training techniques with the use of biofeedback [58-61] may help to train the user to align and coordinate with their new powered prosthesis.

Overall, the work laid out in this dissertation aims to explore human-prosthesis coordination and investigate how the user and device influence one another. In Chapter 2, we explore the timing and limb position of individuals walking with a powered ankle-foot prosthesis for the first time to better understand, at the *local* level of the intersection of the human and prosthesis, potential reasons why overall gait performance has not been significantly improved. In Chapters 3, we explore a novel tuning and training paradigm combining autonomous prosthesis tuning of a powered knee prosthesis with visual feedback to the user to further understand the effect of the prosthesis control on the user and the effect of the user on the prosthesis control. Lastly, in Chapter 4, we further explore the effect of prosthesis tuning on *local* and *global* variables to better understand the necessary objective function when tuning powered prostheses. This work strives to offer evidence to better inform future prosthesis tuning and human training protocols to enable the best coordination possible between the user and their device. When perfect coordination/harmony/symbiosis is achieved, we hope this will allow more possibilities for prosthesis users to achieve any goal they set their mind to.

CHAPTER 2: HUMAN-PROSTHESIS COORDINATION: A PRELIMINARY STUDY EXPLORING COORDINATION WITH A POWERED ANKLE-FOOT PROSTHESIS

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2.1 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˚ 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.

2.2 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 [23]. 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 [62]. Some studies show powered prostheses reduce the metabolic cost of walking in amputees [23, 41], while other studies have reported no significant difference [17, 42]. 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 [18]. Additionally, some studies have shown powered prostheses' ability to increase the affected limb's step-to-step transition work walking on level ground [41, 63], but another study reported no significant differences between the two devices [42].

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 [42, 51, 64-66], 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 lateral gastrocnemius) and a single controller: the human central nervous system [45]; 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.

2.3 Methods

Participants

Five individuals with unilateral transtibial amputation (age: 38 years (IQR 4); mass: 101 kg (IQR 6); and height: 1.85 m (IQR 0.02)) 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 minutes 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 inches) to wear the BiOM ankle-foot prosthesis (BiOM T2, Ankle, BionX Medical Technologies Inc., Cambridge, MA, USA).

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) (Figure 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 [67]. 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 [68]. After tuning, participants walked at their self-selected treadmill speed for 10-minute 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 minute session was within the 95% confidence interval of the mean of the first 30 seconds of the 10 minute 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 [17, 18]. 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 [57].

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

2.4 Results

Figure 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 (Figure 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 ($X^2_{(1)} = 13.523$, *p*=0.0002). TB01's peak propulsion time was 86% with the BiOM, but 91% with the Passive ($X^2_{(1)} = 15.759$, $p<0.0001$). 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 (Figure 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.

2.5 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 [42, 51, 64-66]. 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 [52]. 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 [51]. Our preliminary evidence highlights the importance for a further 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 [69, 70]. 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 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 [58], 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 [57]. Future work is needed to determine how long acclimation and training should be when using a powered prosthesis.

2.6 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.

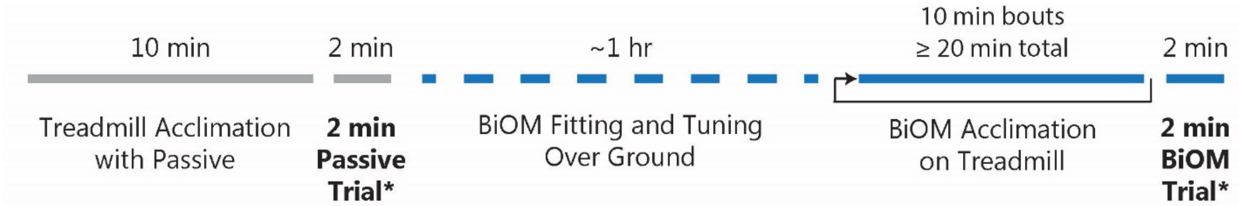

Figure 2.1: Explanation of methods. Participants were first acclimated to treadmill walking with their passive prosthesis and a two-minute trial was collected. Next, participants were fit with the BiOM powered ankle-foot prosthesis and tuning was performed over ground. After tuning, participants were 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.

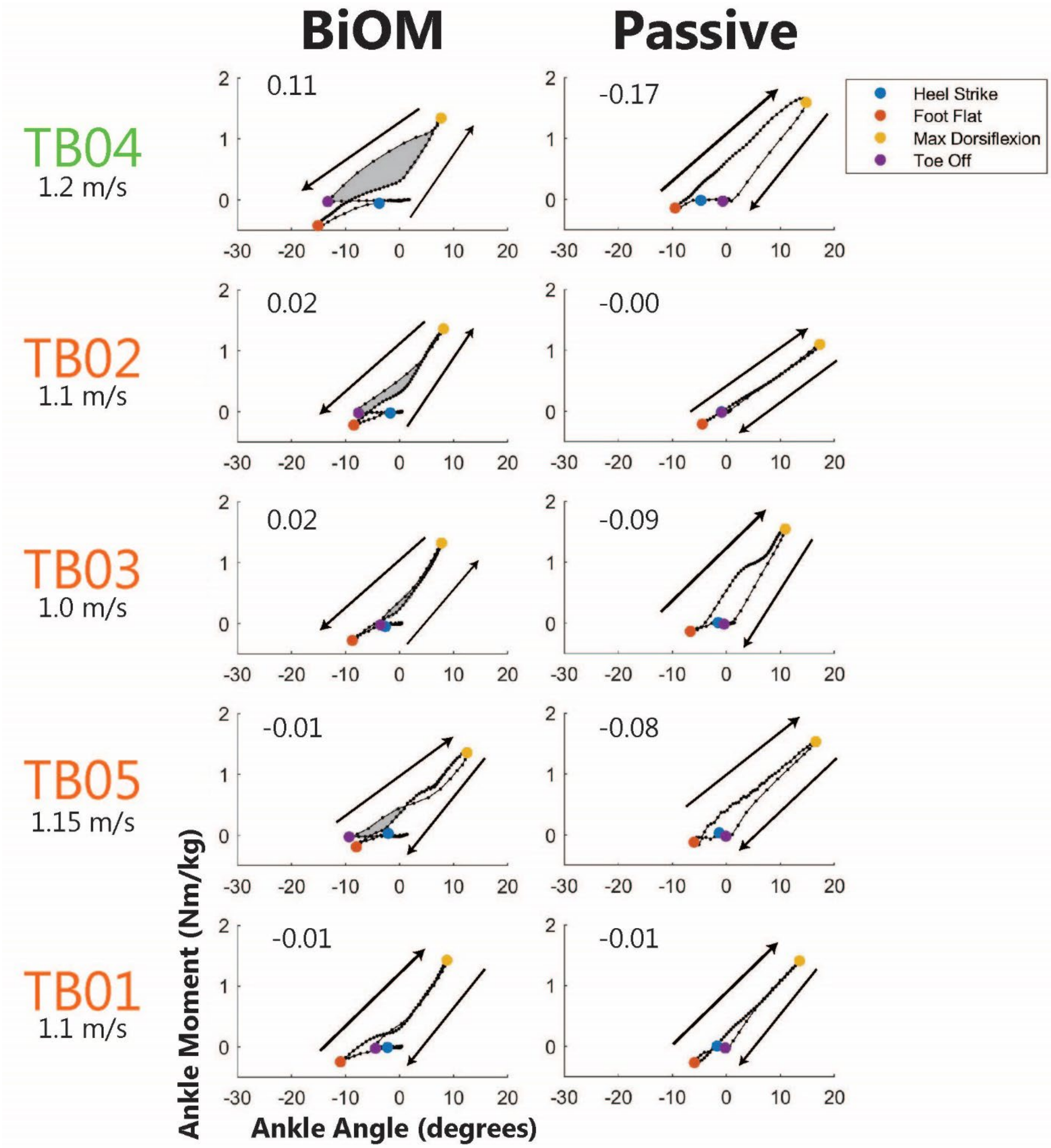

Figure 2.2: Ankle work loops for the BiOM and Passive prostheses. Ankle moment (normalized to body weight) is plotted versus ankle angle for each participant. Ankle moment and ankle angle were averaged over 10 steps. Positive work is indicated by counterclockwise arrows and is shaded in gray. Negative 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.

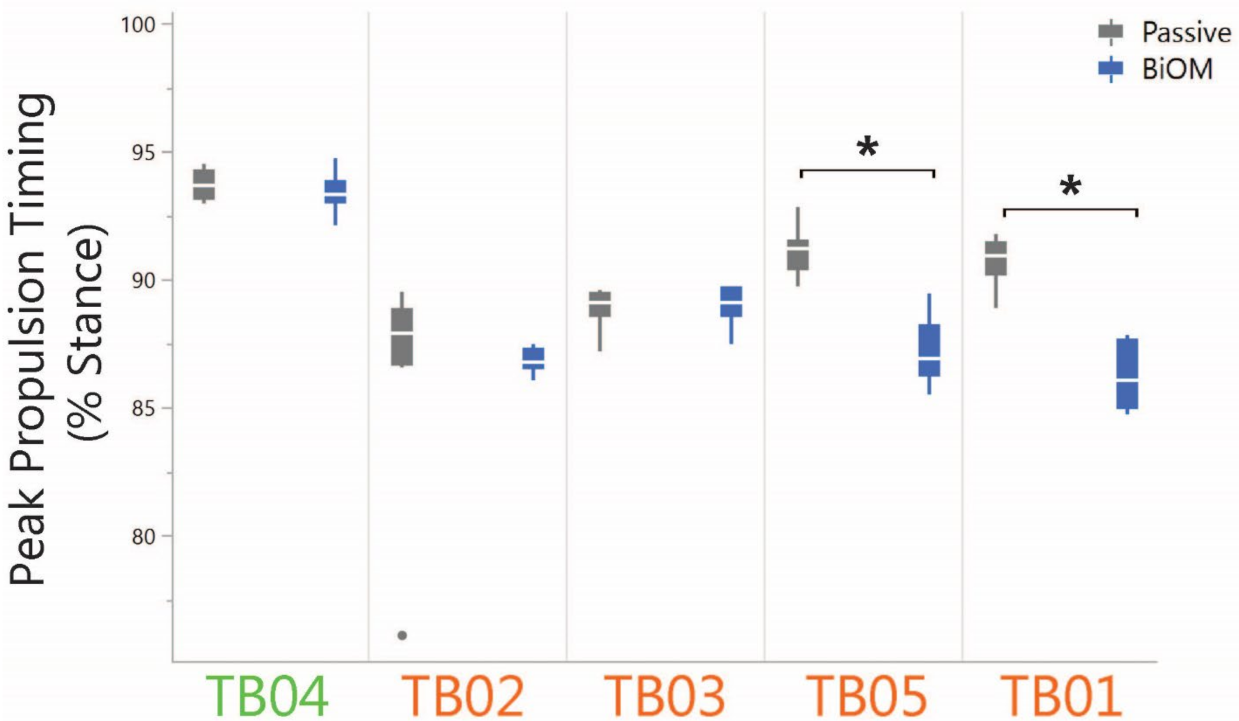

Figure 2.3: Boxplots of timing of peak propulsion with their prosthesis as a percentage of stance time for each participant over 10 steps. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Outliers are defined as 1.5 times the interquartile range from Q1 and 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.

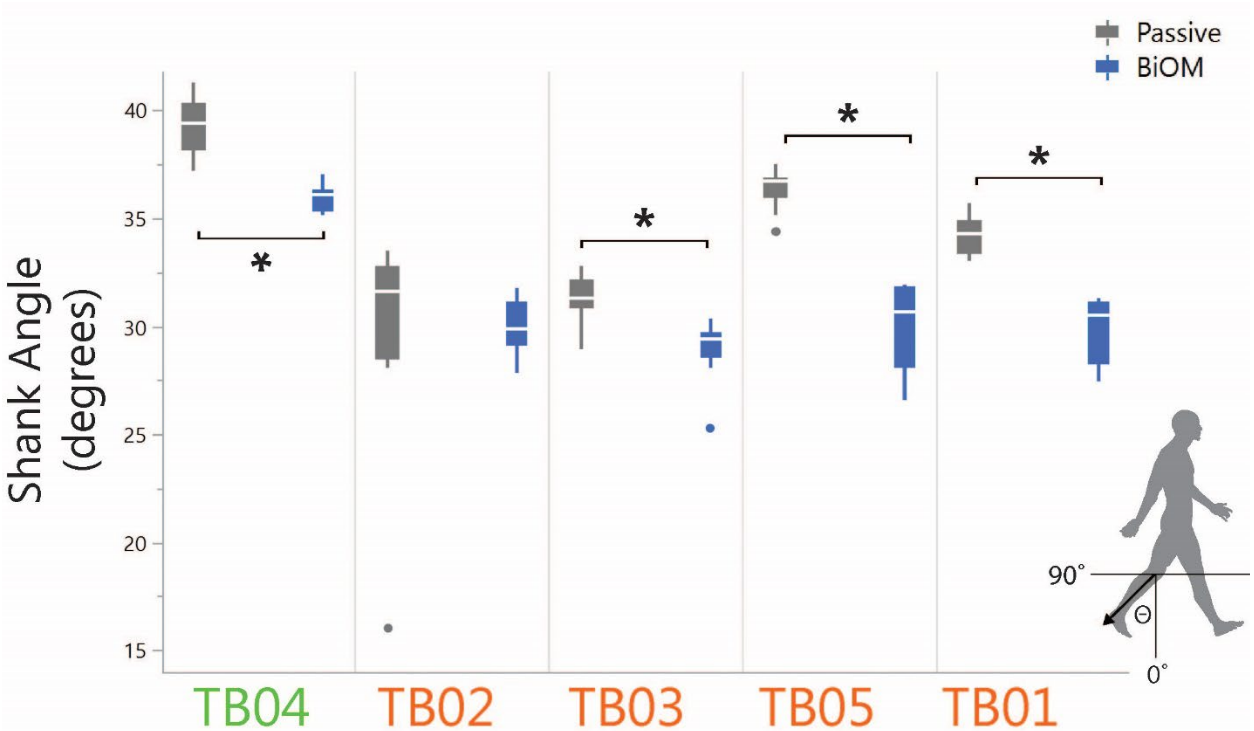

Figure 2.4: Boxplots of prosthesis side shank angle for each participant over 10 steps. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Outliers are defined as 1.5 times the interquartile range from Q1 and Q3. Asterisks (*) indicate significant differences 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.

Table 2.1: 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)		

Table 2.2: 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*

CHAPTER 3: EFFECTS OF COMBINING EXTENDED STANCE TIME VISUAL FEEDBACK WITH AUTONOMOUS PROSTHESIS TUNING ON HUMAN-PROSTHESIS PERFORMANCE

In preparation for *Journal of NeuroEngineering and Rehabilitation*

3.1 Abstract

Background: When walking with a powered prosthesis, compared to traditional passive prostheses, the user must learn to coordinate their own control system with the control system of the robotic device. Previous research has investigated methods of tuning the control parameters of the device to improve human-prosthesis performance while other research has tested the capabilities of the user to improve performance with visual feedback. However, in the case of the tuning study, the tuning algorithm experienced difficulty in tuning the timing and in the case of the feedback study, the user could control their stance time, but the device control was fixed and a trade-off with balance was observed. Therefore, the goal of this study was to combine autonomous prosthesis tuning with visual feedback to the prosthesis user to explore human-prosthesis interaction – can feedback to the user assist the tuning algorithm and can tuning with feedback result in parameters that support better performance for the user?

Methods: Seven individuals without amputation (AB) and four individuals with an above-knee amputation (TFA) participated in this study. Participants completed three rounds of prosthesis tuning with three feedback (FB) conditions: no FB, self-selected stance time FB (SS FB), and increased stance time FB (Inc FB). The prosthesis tuning utilized a reinforcement learning algorithm in bouts of two minutes until convergence criteria were met. After tuning converged, participants walked with the steady-state parameters for an additional two-minute trial and in the

case of SS FB and Inc FB, the visual feedback was removed for the second minute to test for short-term retention. At the end of the experiment, three one-minute trials were completed with the three parameter sets tuned from the feedback conditions to test for any "long-term" retention. Time for tuning to converge, variability in reaching the feedback target, as well as stance time and stance time symmetry were measured and compared across population (AB and TFA) and feedback level (no FB, SS FB, and Inc FB).

Results: When analyzing the time to tune the device with and without feedback, a significant effect was observed both between the feedback levels and population – individuals without an amputation significantly decreased their tuning time with feedback while individuals with an amputation did not. Additionally, when comparing feedback target accuracy during tuning and during steady-state, no significant difference was noted showing encouraging results that tuning does not disrupt the user's ability to complete the feedback task. Finally, users were able to increase their stance time and achieve better stance time symmetry with Inc FB and maintain this behavior in the minute after feedback was removed; but, this behavior did not translate to the post-test condition. However, upon individual analysis, roughly half the participants (6/11) had significant differences in stance time and/or stance time symmetry indicating that within each participant, the parameters may influence their behavior, but not necessarily in the same way across all participants.

Conclusions: This study offers a new framework of combining user-fed feedback with prosthesis tuning to understand human and prosthesis interaction and coordination. The results from this study showed modest benefits; however, additional research investigating additional tuning and feedback goals may enable better performance.

3.2 Introduction

Traditionally, most individuals with a lower-limb amputation are prescribed a passive prosthesis that acts as a simple spring to store and return energy during the push-off phase or lock and unlock to support stance and swing phases, respectively. Due to their simple nature, the user is easily able to predict and adapt to the device. However, the lack of push-off power for the passive prosthesis results in gait compensations and increased energy cost [6, 41, 71]. Therefore, in recent years, powered prosthetic limbs have been developed to replicate biological lower-limb joints to return active energy to the user via motors [20-24, 72]. When walking with a powered prosthesis, there are two control systems working together – the human control system and the prosthesis control system. The user can influence the prosthesis behavior and the prosthesis can influence the user's behavior. For example, the user can modulate how long they stand on the prosthesis in the stance phase and the device can modulate the amount of knee flexion in the swing phase. Researchers have begun to explore the influence of the device on the user and of the user on the device; however, this intersection of human-prosthesis coordination is not well understood.

Multiple research groups have created and investigated their own autonomous prosthesis tuning algorithms that will adjust the control parameters of the prosthesis as the user is walking to achieve a pre-determined goal. One example from Wen et. al. [55] set the goal of the prosthesis tuning to achieve normative knee kinematics. Tuning the prosthesis to replicate biological joint kinematics is a common tuning goal [26, 29] and the knee profile is also associated with stance time symmetry between the intact and prosthetic limbs [73]. The results from this study successfully tuned the prosthesis to achieve a normative knee angle profile; however, it was noted that the tuning algorithm was able to converge to the peak values of the knee profile more easily than the timing of the peak values. Because the user has input into the timing of the gait cycle (i.e. when to heel strike and push-off), it may be advantageous to include the user into the tuning protocol to assist the tuning algorithm.

On the human side of human-prosthesis coordination, many studies have utilized forms of biofeedback to return information back to the prosthesis user in order to modify their behavior [74, 75]. However, research investigating feedback with individuals with an amputation wearing a powered prosthesis is less prolific. One study of note used visual feedback to control the user's stance time of their prosthetic limb while walking with a commercial powered knee prosthesis with fixed parameters [58]. This study successfully demonstrated that the user is able to modify their stance time with visual feedback and when asked to increase the time spent on the prosthetic limb, every participant was able to increase their stance time and improve their stance time symmetry. However, a follow-up analysis found a decrease in medial-lateral balance [59] demonstrating the possible limitation of using fixed control parameters and not re-tuning the device to support the increased stance time. If the prosthetic knee were tuned while the user is in a more favorable gait pattern (i.e. better stance time symmetry), the performance between the human and the prosthesis may be improved.

Therefore, this study aims to combine visual feedback to the prosthesis user and autonomous prosthesis tuning to understand how the human and prosthesis control systems influence one another. To our knowledge, this is the first study to combine feedback and tuning and therefore there are a few questions that arise: 1.) will controlling the user's timing (via visual feedback) help the tuning algorithm to converge faster? 2.) will the tuning process (adjusting the control parameters) influence the user's ability to reach and maintain the feedback goal? 3.) will the parameters tuned while the user is in a more favorable gait pattern (i.e. extended stance time resulting in better symmetry) allow the user to maintain this behavior even after feedback is removed? Therefore, we hypothesize that 1.) the addition of visual feedback to auto-tuning will allow faster tuning because the user can control the timing component of the prosthesis tuning 2.) feedback performance will not be significantly affected by the tuning procedure because the changes made in the device iteration-to-iteration are small to avoid dangerous and unstable parameters and 3.) the prosthesis parameters tuned with increased stance time feedback will improve the user's stance time and stance time symmetry after feedback is removed because the parameters will be tuned to support this improved behavior.

3.3 Methods

Participants

Seven able-bodied (AB) individuals (age: 28 ± 8; mass: 72.2 ± 8.0; and height: 1.76 ± 0.04) and four individuals with a unilateral transfemoral amputation (TFA) (age: 38 ± 17; mass: 82.5 ± 7.5; and height: 1.76 ± 0.07) participated in this study (Table 1). Participants were recruited from the local community who were conveniently available to participate and the following inclusion/exclusion criteria were used: able to walk on a treadmill at 0.6 m/s for at least two minutes unassisted, no known comorbidities such as cardiovascular or neurological conditions that may affect their performance in this study, and were at least 170 cm tall to be able to wear the powered prosthesis with reasonably tall contralateral shoe lift. Participants provided written, informed consent to participate in this study approved by the University of North Carolina at Chapel Hill Institutional Review Board.

Prosthesis Training & Acclimation

Participants without an amputation were first trained to walk with the powered knee prosthesis. Participants trained with the device for at least five visits. The purpose of training was to ensure that all participants were able to adapt to the powered knee prosthesis, be confident to walk with the device on the treadmill, and produce a consistent gait cycle. All participants without an amputation wore an L-shaped, bent-knee adapter to connect to the powered prosthesis to generate a human-prosthesis system. During the first training day, the device was aligned following the L.A.S.A.R. protocol [76]. If needed, a shoe lift on the contralateral limb was added to make sure the hips were level. After alignment, participants were instructed to shift their bodyweight side-to-side while observing their real-time ground reaction forces (GRFs) with the goal to weight-bear symmetrically. Participants were then instructed to march in place while holding the handrails to learn how to trigger the prosthesis to walk. After approximately 30 successful triggers, participants walked over-ground with parallel bars for support if needed. On training days 2-5, participants walked on the treadmill for at least 18, 45 second trials starting at 0.3 m/s and gradually increasing speed to 0.6 m/s. Participants had their speed increased by 0.05 m/s every three trials. After participants reached 0.6 m/s they could choose to increase their speed if they had less than two handrail touches in the previous three trials. The maximum speed participants could choose was 0.8 m/s. Participants were instructed to try to walk as fast as possible without touching the handrails. For the treadmill walking conditions, participants walked with generic knee control parameters that were hand-tuned to allow comfortable and safe walking. After the five days of training, all participants wearing the prosthesis adapter fit the criteria for a high functioning K3 level amputee that posses the ability to ambulate with variable walking speed independently. They were able to adapt to the powered prosthesis and generate a stable and consistent gait pattern.

New participants with an amputation underwent similar training with a few modifications. A certified prosthetist assisted with the initial alignment of the powered prosthesis with their current socket and add a shoe lift if needed on the contralateral limb. Participants then underwent 1-2 days of training with a physical therapist starting with static weight-bearing symmetry, marching in place, over ground walking, and then treadmill walking. The desired knee impedance control parameters were calibrated for each individual by an experienced experimenter to allow comfortable walking. Participants were able to achieve steady-state walking at 0.6 m/s after one or two sessions. Two of the four participants had previous walking experience with the prosthesis and were given approximately 20 minutes at the beginning of the first day to re-acclimate to the device and fine-tune the parameters if needed.

Experimental Protocol

After training, participants returned to the lab for two visits to complete the study protocol. Participants completed three conditions as detailed in Figure 1 – auto-tuning without visual feedback (No FB), auto-tuning with self-selected stance time visual feedback (SS FB), and auto-tuning with increased stance time visual feedback (Inc FB). On the first day, participants first completed the baseline condition (no FB) to establish how long tuning typically takes to converge for each participant and find a set of baseline tuned parameters. Participants finished the first testing day with exploration of the visual feedback. After exploration, participants walked with their self-selected stance time (measured from their baseline walking trial) as the feedback goal to confirm the goal felt similar to their natural walking pattern. Afterwards, the feedback goal was slowly increased to find their increased stance time FB goal (see Visual Feedback section for details). During the second day, the remaining two conditions (SS FB and Inc FB) were randomized with roughly half of the participants receiving each condition first. For the auto-tuning FB trials, participants walked in bouts of approximately two minutes with the feedback until the tuning converged (see Prosthesis Tuning section for details of convergence criteria). Then a separate two minute trial was collected where the final tuning control parameters were held constant and the participant walked with the feedback for the first minute and then feedback was removed for the second minute. Participants were instructed to "walk most comfortably" when feedback was removed to observe if there was any short-term retention of the feedback on their chosen stance time. The instructions were explicitly vague so as not to influence their chosen behavior when feedback was removed.

After the two conditions on day two, participants were given a longer break to rest and an additional three, one minute trials were completed in which participants were randomly given the three different control parameters, determined by the auto-tuning no FB, auto-tuning self-selected FB, and auto-tuning increased FB. These post-test trials were collected to measure the effect of the control parameters themselves on the user's performance.

Gait kinematics and kinetics were measuring during these three conditions and three post-test trials. Kinematics were captured with 43 light-reflective markers place on the boney landmarks of the acromia, iliac crests, greater trochanters, anterior and posterior superior iliac spine, medial and lateral femoral epicondyles, medial and lateral malleoli, first and fifth metatarsals, and calcanea to measure define the torso, pelvis, thighs, shanks, and feet segments, respectively with the addition of tracking markers on each segment. Markers were placed over clothes and shoes and placed either on the rotation centers of the prosthesis or position-matched to the intact limb. Markers were recorded with a 12-camera motion capture system (VICON, Oxford, UK) sampled at 100 Hz and bilateral ground reaction forces (GRFs) were synchronously recorded via a Bertec split belt treadmill (Bertec Corp, Columbus, OH, USA) sampled at 1000 Hz.

Prosthesis Tuning

For the auto-tuning trials, the prosthesis control parameters were adjusted every four strides (counted as one iteration) to achieve a normative knee angle profile [77] (Figure 1). We characterized the profile with four discrete feature points, and each point is a local extrema in the corresponding gait phase along the profile. A total of 12 impedance control parameters (stiffness, equilibrium position, and damping for each of the four gait phases – stance flexion, stance extension, swing flexion, and swing extension) were adjusted to match the normative knee angle profile. The adjustment was realized by an existing reinforcement learning algorithm as detailed in Li et. al [56], and it was considered a success when the tuning convergence criteria were met, where the peak and duration errors between the measured and normative knee angle profiles (as defined in Fig. 1) were within 2° and 3% (i.e. target range), respectively, for 8 out of 10 consecutive iterations.

During auto-tuning trials, participants walked in bouts of approximately two minutes consisting of 10 iterations (40 strides) to prevent fatigue. For the no FB and SS FB conditions, our goal was the normative knee profile as shown in Figure 1. In the case of the Inc FB condition, after each bout of 10 iterations, the duration error was calculated and if 8/10 of the iterations were not in the target range the stance phase of the normative knee angle profile was increased by 2% to accommodate changes in the knee profile curve due to the increased stance time FB goal. This allows the prosthesis tuning to adapt to the user's changing behavior.

Visual Feedback

Similar to Brandt et. al [58], we utilized a unilateral, temporal metric (i.e. prosthesis-side stance time) to display as visual feedback due to the user's high spatial symmetry variability [78] and difficulty with gait symmetry [79]. The visual feedback was created via custom code using the

Vicon DataStream SDK (VICON, Oxford, UK) and MATLAB (The MathWorks, Inc, Natick, Massachusetts, USA) and displayed on a computer monitor, approximately one meter in front of the participant on the treadmill. Prosthesis-side stance time was calculated from the real-time GRF signals using a 20N threshold. The feedback displayed to the user was averaged over four strides to smooth the signal and reduce any large stride-to-stride corrections.

To determine the feedback goal for the two feedback conditions, the participant's self-selected stance time was first calculated with their hand-tuned parameters on the first day of the experiment. Participants could then fine-tune the goal until it felt comfortable. The visual feedback level was then increased by increments of 0.05 seconds until participants noted at least an increase of at least 1 in their self-reported rate of perceived exertion, the prosthesis-side stance time did not overlap with their self-selected stance time, and they could reliably maintain the target stance time. Most participant's increased stance time goal was 0.10 to 0.15 seconds longer than their self-selected stance time. Both targets were centered on the screen and the display range remained at +/- 0.3 seconds from the target line to maintain participant's perceived accuracy. Both goals were re-introduced at the beginning of day 2 and adjusted if needed until they felt comfortable with the two feedback levels.

Data Processing

Commercial data analysis software (Visual 3D, C-Motion, Inc., Germantown, MD, USA) was used to process the data. Marker data was low-pass filtered with a 4th order Butterworth filter with a cutoff frequency of 6Hz and GRF signals were smoothed by a 4th order Butterworth filter with a cutoff frequency of 25Hz. Gait events (heel contact and toe off) were determined with a threshold of 20N. To determine gait symmetry, we used a standard symmetry index where x_i and x_p are the stance time of the intact and prosthesis sides, respectively.

$$SI = \frac{x_i - x_p}{(x_i + x_p)/2} * 100\%$$

Target-hitting error was defined as the absolute value of the distance between the goal stance time and the user's prosthesis-side stance time and the target-hitting variability was the standard deviation of their error. Extrapolated center of mass (xCOM) was calculated using the equation from Hof et al [80]. that was developed for transfemoral amputees and margin of stability (MOS) was estimated using the heel marker of each foot as the base of support (BOS).

$$xCOM(t) = z(t) + \frac{1}{\sqrt{g/h}} \cdot \frac{dz}{dt}$$

$$MOS = xCOM - BOS$$

Statistical tests

To answer the first hypothesis, if tuning with visual feedback facilitates faster tuning, we performed a two-way cross tabulation Chi-square test for independence to examine if there is a relationship between populations (TFA and AB) and feedback level (no FB and SS ST FB). If the Chi-square test reaches significance, the adjusted residual will be used for the post-hoc comparison (the z critical values of ± 1.96 were set as the significance level). To answer the second hypothesis, if tuning affects feedback performance, pooled t-tests were used comparing error and error variability during tuning and during steady-state for both the SS FB and Inc FB conditions. Finally, to answer our third hypothesis, if participants can retain their increased stance time and better stance time symmetry after feedback was removed, two-way mixed model ANOVAs were used including participants as random effects and feedback, population, and their crossed effects as fixed effects. We compared across population (TFA vs. AB) and across the Inc

FB conditions (with FB, immediately after FB, and long after FB (post test)) as well as across the post test conditions (no FB – post, SS FB – post, and Inc FB – post). For the fixed effects that reached the significant level, Tukey's honestly significant difference test was used for the post hoc comparisons. A follow-up analysis within participant was conducted using the Friedman non-parametric test. A non-parametric test was chosen because the samples within participant were non-independent samples (i.e. consecutive walking steps). The Shapiro-Wilk test was used to test the normality before running the ANOVAs. The significance level was set at $\alpha = 0.05$, and all analyses were performed using JMP (JMP, SAS Institute, Cary, NC, USA).

3.4 Results

Time to tune the prosthesis

We defined the tuning performance with the number of tuning iterations during tuning without FB and with self-selected stance time FB (Figure 2). There was a significant effect both for population and feedback ($X^2_{(1)} = 14.134$, $p < 0.0001$). The comparison between the feedback level in the population without an amputation showed that the no FB condition required significantly more tuning iterations than the SS FB condition (adj. residual: no FB 22.9 and SS FB -22.9). In addition, in the population with an amputation, a significantly longer tuning time was required with feedback compared to without feedback (adj. residual: no FB -22.9 and SS FB 22.9). Comparing the two conditions, seven participants did not experience a noticeable difference in tuning time (< 5 iterations difference (~20 steps)). Three participants had a noticeable decrease in the number of tuning iterations (tuning was faster with the addition of feedback) and one amputee participant (TF02) had a noticeable increase in the number of iterations (tuning was slower with the addition of feedback).

When investigating an explanation for why tuning was longer with FB for TF02, balance was investigated. Figure 3 shows the medial-lateral position of the extrapolated center of mass (xCOM) relative to the base of support (BOS) as defined by the prosthesis and intact limb heel markers over the prosthesis stride. On average, there was no noticeable change in the xCOM movement in relation to the base of support. However, TF02 experienced a 23% decrease in the average margin of stability (average difference between the xCOM and prosthesis-side BOS per stride) when the parameters were tuned with FB.

Visual Feedback Target Accuracy

We defined feedback performance as the absolute value of the error between the user's stance time and the goal as well as the variability around the goal (standard deviation of the absolute error) as seen in Figure 4. For the two trials with feedback (self-selected and increased stance time FB), there was no significant difference in the error ($p = 0.1149$ for SS FB and $p = 0.4471$ for Inc FB) or variability during tuning and during steady-state ($p = 0.9660$ for SS FB and $p = 0.1527$ for Inc FB).

Stance Time and Stance Time Symmetry After Feedback was Removed

Figure 5 shows stance time and stance time symmetry for both populations (AB and TFA) across the baseline condition, Inc FB parameters with feedback (Inc FB with FB), Inc FB parameters immediately after feedback was removed (Inc FB no FB), and Inc FB parameters long after feedback was removed (Inc FB post). All participants increased their stance time (1.05 sec → 1.15 sec for AB participants and 0.96 → 1.06 sec for TFA participants) and decreased their stance time asymmetry (30% → 23% for AB participants and 24% → 19% for TFA participants). Comparing the three trials with Inc FB parameters, a significant difference was observed in the prosthesis-side stance time ($p = 0.0021$). In a post hoc analysis, both the Inc FB with FB and Inc FB no FB conditions were significantly different from the stance time in the post test with Inc FB parameters ($p = 0.0032$ for Inc FB with FB and $p = 0.0089$ for Inc FB no FB). Further, when comparing our two populations (participants with and without an amputation), we observed that TFA participants had slightly shorter stance time and slightly better symmetry than the AB participants (~0.1 second shorter and ~5% more symmetric), but both groups followed similar trends and were not significantly different.

Stance Time and Stance Time Symmetry in the Post Test

Across all participants, there were no significant differences in stance time or stance time symmetry between the three post-test conditions (no FB, SS FB, and Inc FB); however, there was a significant effect due to the individual participant ($p = 0.0373$ for stance time and $p = 0.0370$ for stance time symmetry). To further investigate this significant effect of the individual participants, we ran a follow-up analysis comparing the three post test conditions within participant. Participants AB04, AB05, AB06, TF01, TF02, and TF04 all had a significant difference in stance time across the post test conditions ($ps < 0.0443$) and TF03 and TF04 had a significant difference in stance time symmetry ($ps < 0.0019$). If we compare the knee angle profiles for these three post test conditions (Figure 6), we can see that for the previously named participants, there are subtle differences in the stance flexion peak between the three tuned parameter sets which may explain the individual differences in stance time and stance time symmetry.

3.5 Discussion

Overall, this study partially followed our hypotheses. The tuning time was faster for participants without an amputation, but resulted in longer tuning time for one amputee participant. This participant; however, had a noticeable change in margin of stability when walking with the addition of feedback. In addition, parameter tuning did not impact the user's ability to reach the feedback target and maintain the stance time target during prosthesis tuning. Furthermore, users were able to maintain their increased stance time and better stance time symmetry after feedback was removed; however, this better performance was short-lived. These results show the promise of adding user control to the prosthesis tuning protocol to possibly speed up tuning and help the user to maintain a more favorable gait pattern overall.

Although the tuning performance was not significantly faster for all participants, some participants (3/11) did experience a large reduction in the time to tune the prosthesis. For these participants, the decrease in the number of tuning iterations would equal approximately two minutes less of continuous walking. In the clinic, this reduction in time could be valuable to prevent patient fatigue and time spent resting as well as allow more time for other training with the prosthesis. From pilot testing within our lab, we found that the time to tune the device varied by 5-6 iterations over 3 trials, so even though the tuning algorithm has some random components, this alone does not explain the large differences we observed.

For the one participant that needed additional time for the tuning to converge, the participant's balance may explain this result. Almost all participants had similar margin of stability with and without feedback, but TF02 had a large decrease in their margin of stability. Previous work has indicated that individuals with an amputation adopt a larger step width and have a larger MOS [80, 81], so a decrease in MOS may indicate increased instability since the user is closer to the bounds of their BOS [82]. Regardless, this participant's balance was noticeably different from all other participants which may explain their longer tuning time.

Feedback performance was unchanged during tuning compared to during steady-state. This is an encouraging sign – the constant change in tuning parameters do not seem to disturb the user's performance to achieve the FB goal. Participants were able to maintain the FB goal with precision and accuracy regardless of prosthesis parameters. The user is able to dominate any changes in the device to control their overall performance.

Similar to Brandt et al. [58], we observed that users are able to increase their prosthesis-side stance time and improve their stance time symmetry index with the addition of visual feedback. In addition, we also found that users were able to maintain this symmetry even when the feedback was removed and they were not explicitly instructed to maintain their increased stance time. However, when these same parameters were re-introduced in the post test trials, on average, the users reverted to approximately their baseline stance time and stance time symmetry. However, roughly half our participants did observe individual differences in stance time and/or stance time symmetry and a difference in the stance phase gait kinematics was observed. Therefore, further investigation into the knee kinematics themselves and their influence on gait performance is needed. If a specific set of knee kinematics assists the user to maintain longer stance time, then the knee tuning goal itself may need to be updated.

This study, to our knowledge, is the first to combine feedback to the user in combination with auto-tuning of the prosthesis device to simultaneously "tune" both the user and device. Even though the results do not show a strong benefit in terms of tuning time and maintained stance time after feedback was removed, this study offers a framework for exploring further combinations of tuning and feedback to improve overall user performance. As seen in previous human-in-the-loop optimization (HILO) studies, performance after tuning is not always significantly improved [51, 83]. By adding feedback to the user, we can control the user's performance and possibly achieve tuned parameters that support this better performance. Further, results from this study demonstrate feedback with tuning does not negatively affect the user's behavior. By exploring further feedback and tuning combinations (aside from prosthesis-side stance time and a standard knee angle profile), we may be able to define a goal for the user and for the device to enable lasting performance gains. One recent study proposed a reinforcement learning algorithm to tune prosthesis parameters to gait symmetry [84] – this tuning goal coupled with our visual feedback may see increased benefits. We can also explore how the feedback and tuning are used, such as alternating between giving the user feedback and tuning the device instead of simultaneously combining both to explore if there is a relationship of how these two "tunings" affect one another.

Finally, this study is not without limitations. Although we were able to recruit 11 participants, seven of our participants were individuals without an amputation wearing a prosthesis adapter. We observed relatively similar observations between the two groups; however, previous work has highlighted the necessity of involving participants with an amputation because previous work has shown significant differences between populations [51]. Future work should recruit additional participants with an amputation. In addition, here we only focused on feedback of prosthesis- side stance time and tuning to a normalized knee angle profile. Future research is needed to explore further combinations of feedback and tuning goals – possibly changing the knee profile tuning goal to the user's preference will enable better retention of stance time symmetry.

3.6 Conclusion

Overall, the results of this study offer a promising framework for future tuning paradigms. When adjusting the control of wearable robotics, it may be beneficial to control the user's behavior via feedback to potentially speed up the tuning algorithm performance and impact the user's end behavior. Although the results of this study found modest reductions in tuning time and only short-term effects of the feedback on overall performance, future studies including different tuning and feedback goals may result in more impactful performance outcomes.

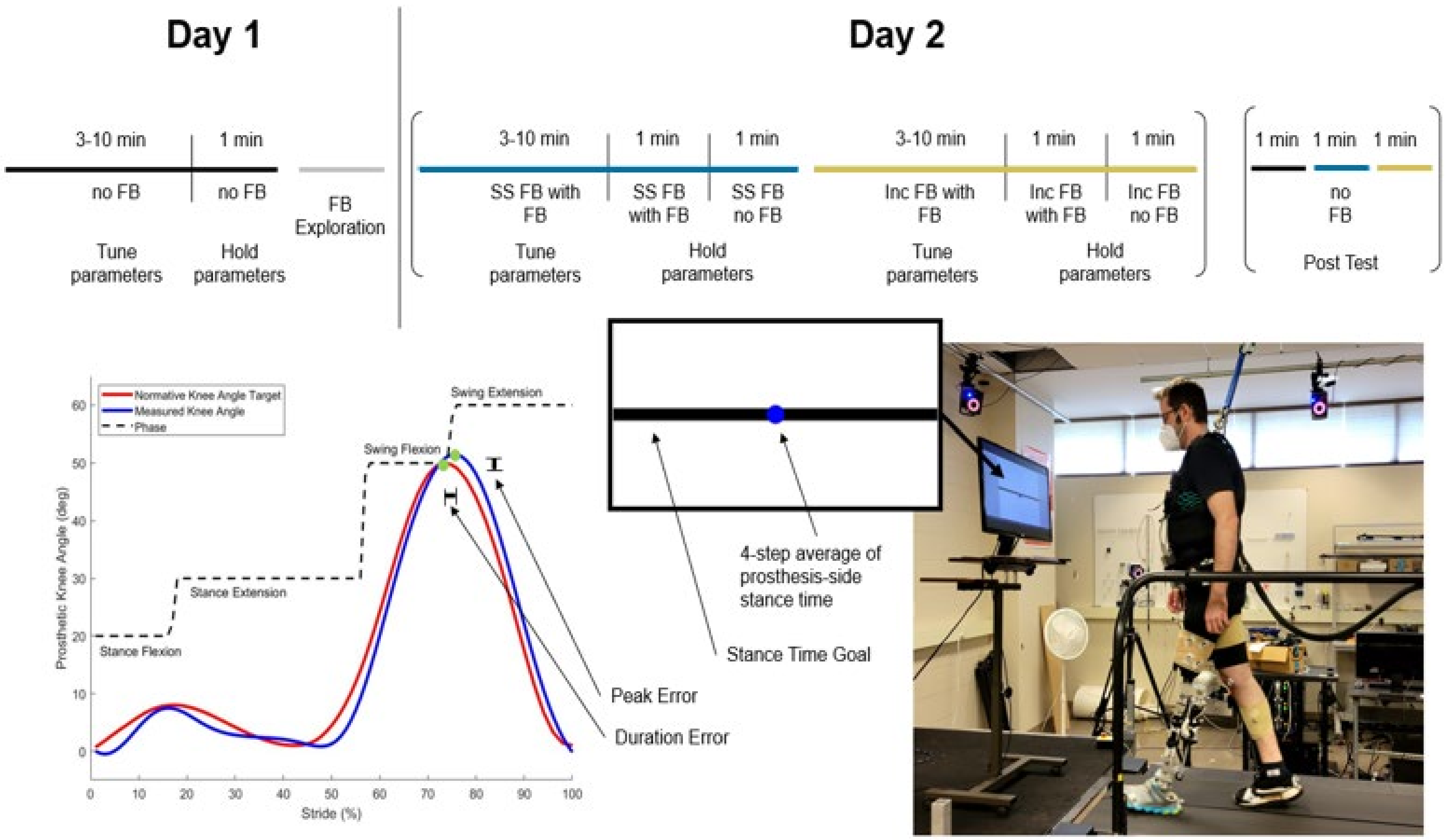

Figure 3.1: (Top) Explanation of methods. On day 1, participants underwent auto-tuning without visual feedback (no FB). After tuning converged, participants walked with the final tuned parameters. At the end of day 1, participants were exposed to the visual to explore. On day 2, participants randomly completed either the self-selected feedback (SS FB) or increased feedback (Inc FB). After tuning converged, participants walked for 2 minutes with the final tuned parameters, but feedback was removed for the second minute. The post test trials were without feedback and were randomly assigned the parameters tuned under the no FB, SS FB, and Inc FB conditions. (Bottom left) Explanation of prosthesis tuning. The red line shows the target knee profile and the blue line shows an example knee profile during tuning. Peak and duration errors were calculated for each phase as shown. (Bottom right) Demonstration of the visual feedback and motion capture during data collection.

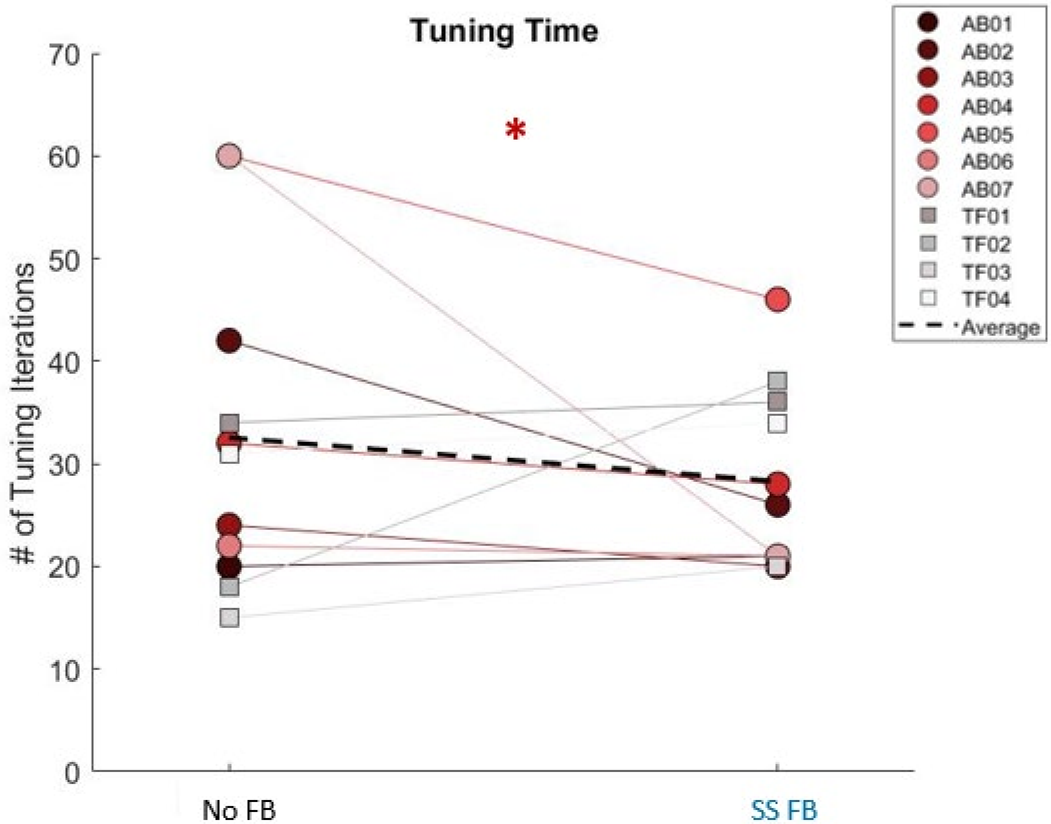

Figure 3.2: Number of tuning iterations with and without feedback. Many participants had similar tuning times with and without feedback; however, some participants had noticeable decreases in tuning time when tuning with preferred stance time visual feedback. One amputee participant (TF02) had a noticeably longer tuning time with feedback.

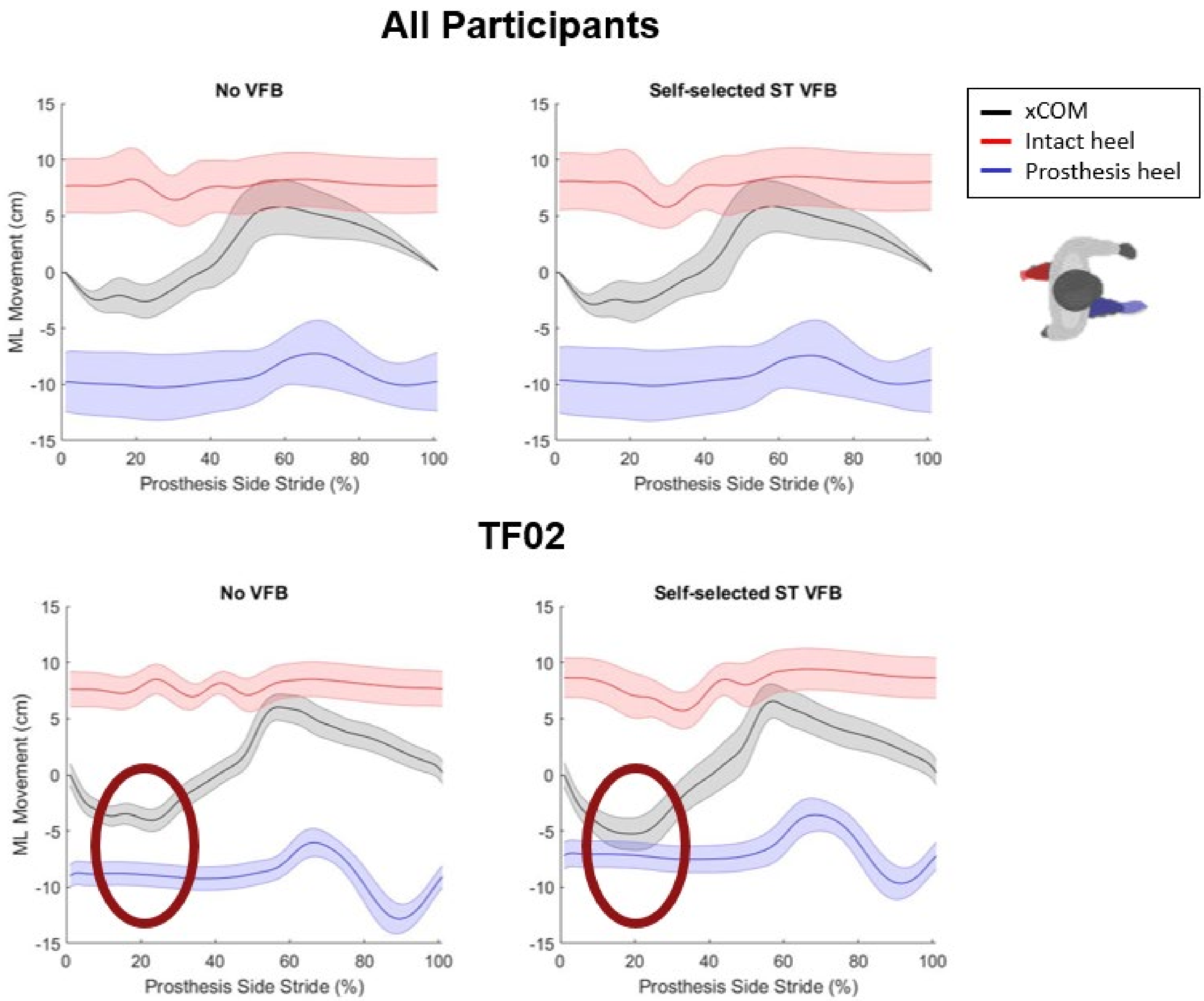

Figure 3.3: (Top) Extrapolated center of mass (xCOM) relative to the prosthesis and intact heel marker positions averaged across all 11 participants. (Bottom) xCOM relative to the prosthesis and intact heel marker positions for TF02. On average, there is no discernable difference in balance between auto-tuned parameters without feedback and auto-tuned parameters with self-selected stance time feedback. However, for participant TF02 that had a noticeable increase in tuning time with feedback, the xCOM had greater medial-lateral movement and was closer to the bounds of the base of support.

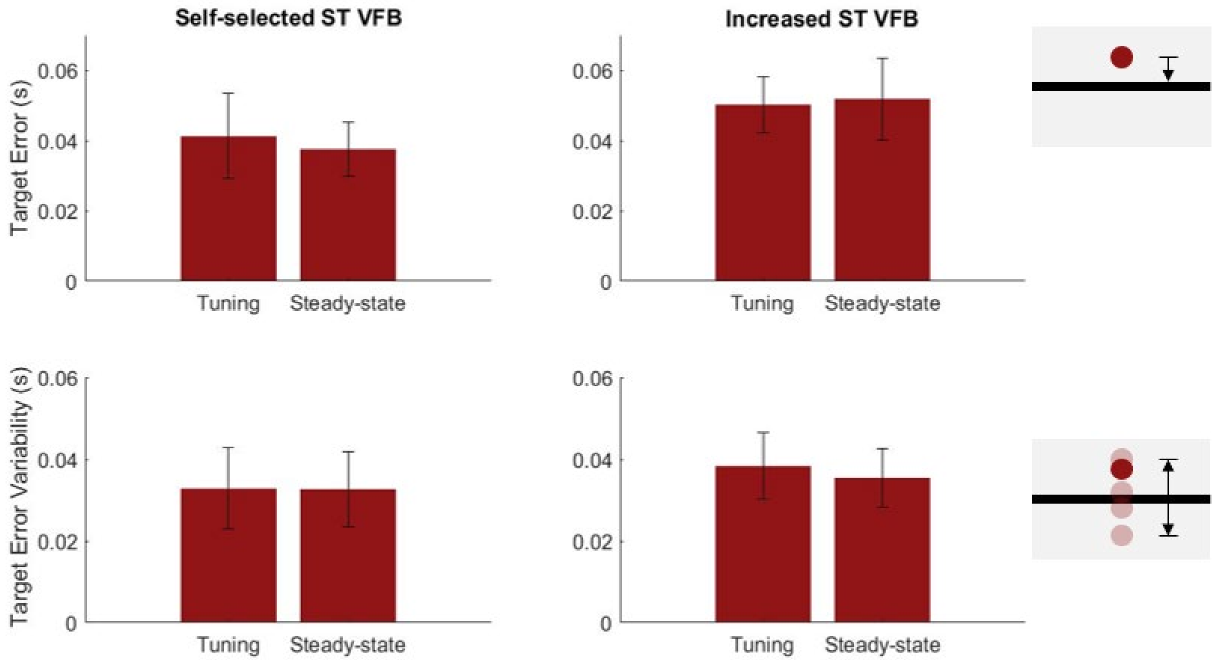

Figure 3.4: (Top) Absolute value of the target error during tuning with self-selected stance time visual feedback and increased stance time visual feedback and during steady-state. (Bottom) standard deviation of the target error. While tuning the prosthesis parameters, the users' performance to hit the target and variability in hitting the target was not different from their steady-state performance. Pictures to the right depict the error and error variability.

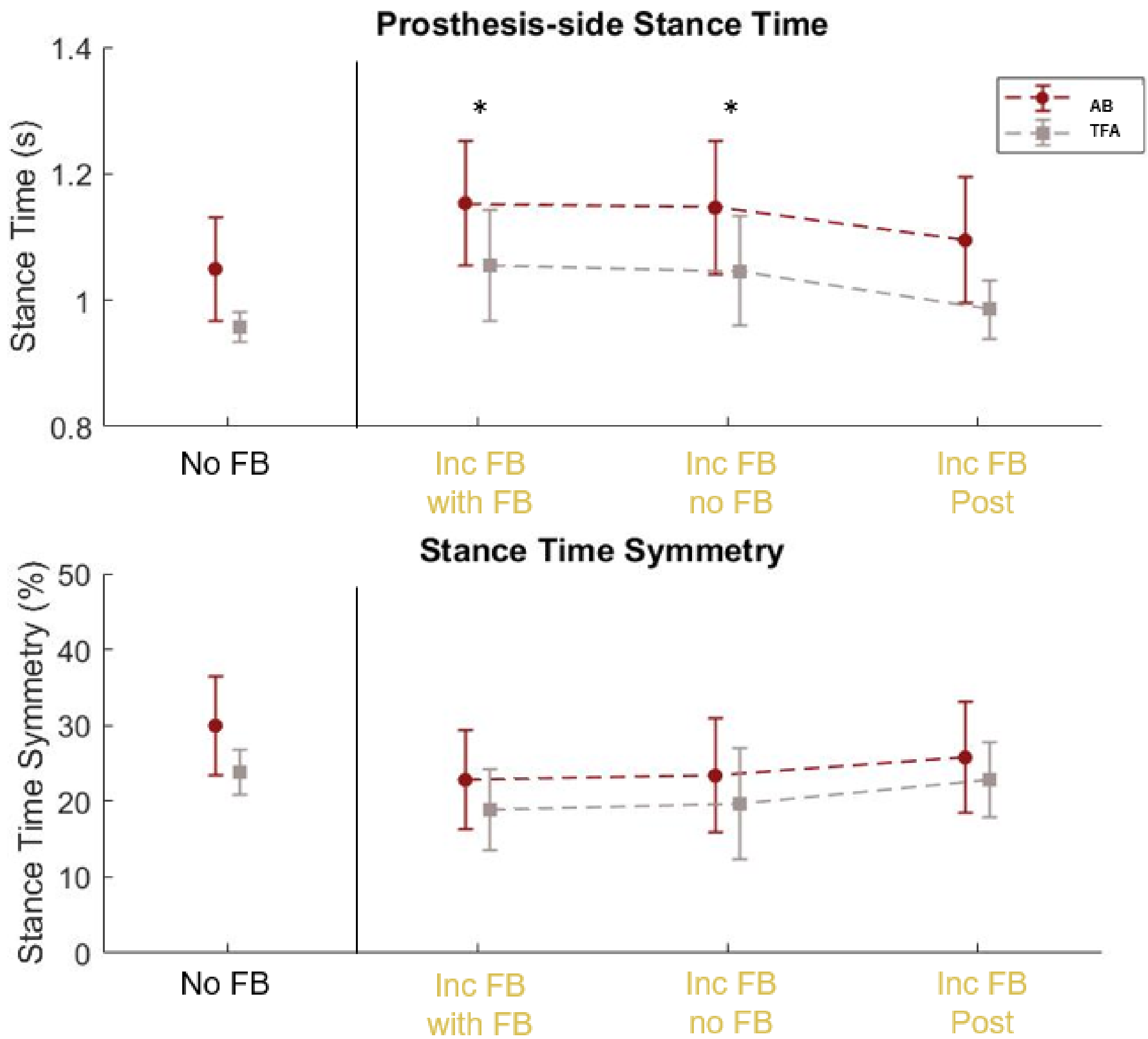

Figure 3.5: Prosthesis-side stance time and stance time symmetry for all participants during no visual feedback (No FB), increased stance time visual feedback during feedback (Inc FB with FB), increased stance time visual feedback after feedback was removed (Inc FB no FB), and increased stance time visual feedback in the post test (Inc FB Post). All participants were able to increase their prosthesis-side stance time with the feedback and maintain it after feedback was removed. However, this longer stance time was not retained in the post test. * indicates significantly different from Inc FB Post

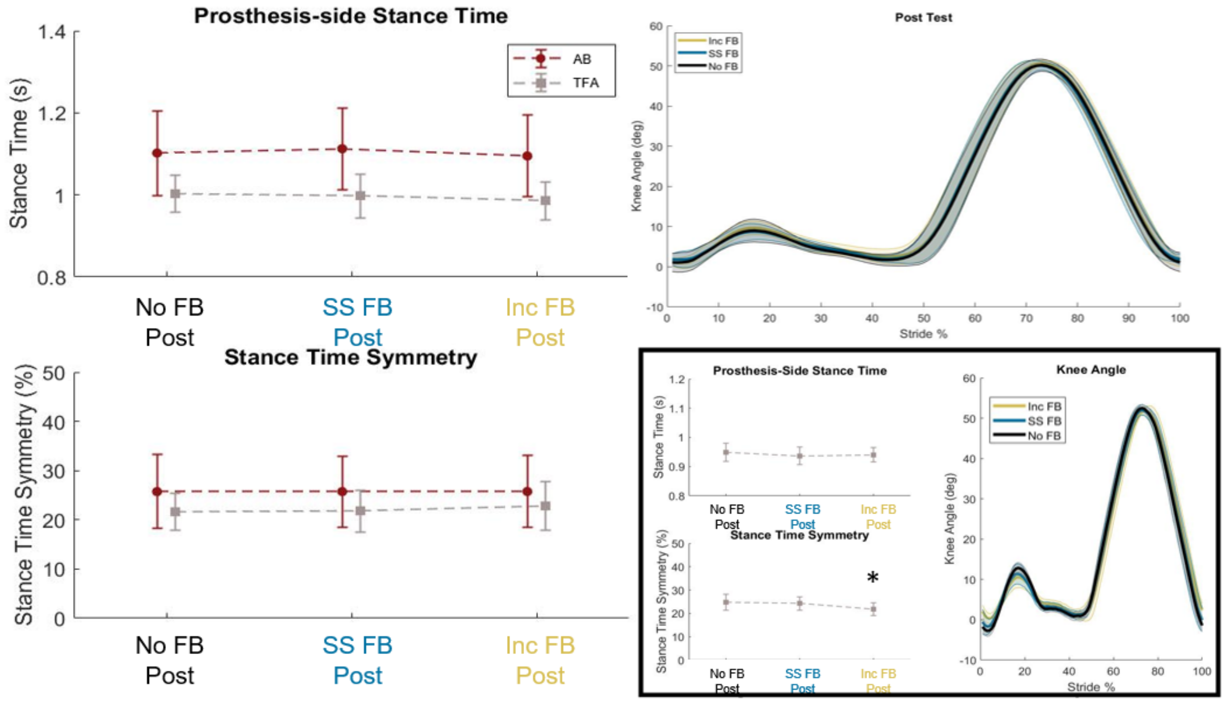

Figure 3.6: (Left) Prosthesis-side stance time and stance time symmetry averaged across all participants for the three post test conditions. The post test compared the three different control parameter sets that were determined after tuning without feedback (No FB Post), with self-selected stance time feedback (SS FB Post), and with increased stance time feedback (Inc FB Post). There was no significant difference across conditions. (Right Top) Knee angle averaged across all participants for the three post test conditions. (Right Bottom) One representative amputee participant stance time, stance time symmetry, and knee angle profile in the three post test conditions. Although the group mean was not significantly different across conditions, roughly half the participants experienced a significant difference in the stance time or stance time symmetry and had visibly different knee angle profiles.

Table 3.1: Summary of participants. Seven individuals without amputation and four individuals with amputation participated in this study. Seven participants had no prior experience with our powered prostheses and underwent training with the device for 1-5 days.

	AB01	AB02	AB03	AB04	AB05	AB06	AB07
Gender	Female	Male	Male	Male	Male	Male	Male
Height	1.78 m	1.75 m	1.79 m	1.75 m	1.72 m	1.70 m	1.83 m
Weight	83.9 kg	70.3 kg	65 kg	75.7 kg	79 kg	61 kg	70.3 kg
Age	23	21	45	29	24	31	24
Prosthesis Side	Right	Left	Right	Left	Left	Left	Left
Previous experience with powered prothesis?	No	No	Yes	Yes	No	No	No

	TF01	TF02	TF03	TF04
Gender	Male	Male	Male	Male
Height	1.68 m	1.73 m	1.83 m	1.80 m
Weight	83.9 kg	72.6 kg	82.9 kg	90.7 kg
Age	23	42	26	61
Prosthesis Side	Left	Left	Right	Left
Type of Amputation	Knee Disarticulation	Transfemoral	Transfemoral	Transfemoral
Previous experience with prothesis?	No	No	Yes	Yes
Time Since Amputation	18 years	3 years	9 years	49 years
Reason for Amputation	Congenital	Trauma	Cancer	Cancer
Prescribed Prosthesis	C-Leg 3 (Ottobock)	X3 (Ottobock)	X3 (Ottobock)	Genium (Ottobock)
Prosthesis Suspension	Suction	Suction	Double-wall with pin to outer socket	Ischial containment suction
Self-reported Functional K-level	4	4	3	3

CHAPTER 4: EFFECTS OF PROSTHETIC KNEE PARAMETERS DURING TUNING ON LOCAL AND GLOBAL JOINT KINEMATICS

In preparation for *IEEE Transactions on Neural Systems and Rehabilitation Engineering*

4.1 Abstract

When tuning a powered prosthesis, defining an objective function to maximize performance is vital. In many recent examples, a global metric (e.g. metabolic cost, lateral balance, etc) is used; however, the resulting performance is not necessarily improved after tuning. Therefore, it is necessary to further investigate the relationship between the changing control parameters during tuning with variables on the local joint level as well as the whole-body global level. To explore this relationship, four individuals with an above-knee amputation were recruited and trained to walk with a powered knee prosthesis. Participants walked with the device while a tuning algorithm updated the impedance control parameters every four steps. Knee angle velocity, hip angle velocity, and center of mass – center of pressure (COM-COP) velocity during the stance phase were compared across tuning. Results show that both the local (hip velocity) and global (COM-COP) variables were sensitive to changes in tuning iteration; however, upon further analysis, the global variable did not have a strong relationship with the changing knee control parameters themselves. This may indicate that either local or global variables are suitable as the objective function; however, the direct relationship between the control parameters and global variable is not well defined.

4.2 Introduction

For individuals that wear a powered lower-limb prosthesis, one major area of research involves tuning the control parameters to enable the user to maximize the performance of the human-prosthesis system. The tuning process searches a large range of impedance control parameters to optimize the end performance using an objective function. Thus, determining an appropriate objective function is needed to minimize error and maximize performance.

In the literature, it is common for a *global* variable, such as metabolic cost, to be used to minimize the user's effort when walking with a prosthesis [51-53, 85]. These *global* variables are defined as all-encompassing variables spanning multiple joints and limb segments. For example, one study reported no significant improvements in walking economy performance when using metabolic cost to tune the prosthesis via human-in-the-loop-optimization (HILO) [51]. As another example, a different study reported no benefits in walking economy performance when using medial-lateral center of mass (COM) movement to tune the prosthesis via HILO [83]. This lack of success may be due to the preservation of these *global* variables due to the infinitely redundant solutions on the level of the individual *local* joint space – the changes in the prosthesis control parameters may not be able to propagate to these *global* level variables. Thus, it is not well studied how the change in the control parameters affects different levels of outcome variables used in the objective function to tune these powered prostheses.

Therefore, our goal was to explore the effect of changing knee prosthesis parameters on *local* and *global* measures of joint kinematics during prosthesis tuning. Specifically, we aim to explore the range of values observed for each tuning iteration of the prosthesis on local and global variables in addition to the direct relationship between control parameters and local/global variables. We hypothesize that local variables will be more sensitive to changes in knee control and show significant differences across tuning iterations compared to global variables.

4.3 Methods

Participants

Four individuals with a unilateral transfemoral amputation (TFA) (age: 38 ± 17; mass: 82.5 ± 7.5; and height: 1.76 ± 0.07) participated in this IRB approved study from the IRB at UNC-Chapel Hill (Table 1). We recruited participants locally in the community who were able to participate based on the following inclusion/exclusion criteria: able to comfortably walk at least 0.6 m/s on a treadmill and have no known comorbidities that may affect their ability to perform the tasks in this study.

Prosthesis Training & Acclimation

Two of the four participants had previous experience walking with the powered knee prosthesis and were given approximately 20 minutes at the beginning of the experiment to re-acclimate to the device. The remaining two participants first learned to walk with the powered prosthesis on an additional 1-2 training sessions prior to the experiment. A certified prosthetist first aligned the powered knee to their current prosthetic socket and added additional components or a shoe lift to ensure the hips were level. Participants then began with walking in place and over ground to learn how to trigger the gait phases of the device under the direction of a physical therapist. Finally participants began treadmill walking and completed training when they could walk comfortably at 0.6 m/s on the treadmill without holding onto the handrails. The parameters of the prosthesis were hand-tuned to allow for safe and comfortable walking without assistance.

Experimental Protocol

For the experiment, participants walked with the powered prosthesis while the prosthesis autonomously adjusted the control parameters to tune the device. The goal of the tuning algorithm was to meet a normative knee angle profile [77] (Figure X). The target profile was defined by four discrete points, the minimum and maximum values of the stance flexion, stance extension, swing flexion, and swing extension phases of gait and a total of three parameters (stiffness, equilibrium position, and damping) were adjusted for each phase resulting in a total of 12 tunable impedance parameters. Tuning converged using a reinforcement learning algorithm developed by Li et. al. [56] and was defined by the peak and duration errors between the measured and normative knee angle profile were within 2° and 3% (i.e. target range), respectively, for 8 out of 10 consecutive iterations.

The control parameters were adjusted every four steps (counted as one iteration) and participants walked in bouts of approximately two minutes to prevent fatigue. After tuning converged, participants walked for a separate two minute trial with the final parameters held at steady-state to measure final performance with fixed parameters.

Gait kinematics and kinetics were measuring across tuning. Kinematics were captured with 43 light-reflective markers place on the boney landmarks of the acromia, iliac crests, greater trochanters, anterior and posterior superior iliac spine, medial and lateral femoral epicondyles, medial and lateral malleoli, first and fifth metatarsals, and calcanea to measure define the torso, pelvis, thighs, shanks, and feet segments, respectively with the addition of tracking markers on each segment. Markers were placed over clothes and shoes and placed either on the rotation centers of the prosthesis or position-matched to the intact limb. Markers were recorded with a 12-camera motion capture system (VICON, Oxford, UK) sampled at 100 Hz and bilateral ground reaction forces (GRFs) were synchronously recorded via a Bertec split belt treadmill (Bertec Corp, Columbus, OH, USA) sampled at 1000 Hz.

Knee angle and hip angle were both calculated in the sagittal plane and segmented into individual strides (Figure 1). Center of mass – center of pressure angle (COM-COP) was also calculated in the sagittal plane relative to a vertical line through the COM. Net COP was calculated using the equation below where *i* refers to the intact side, *p* refers to the prosthetic side and F_z refers to the vertical ground reaction force of the prosthetic or intact sides.

$$COP_{net} = COP_p \cdot \frac{F_{zp}}{F_{zi} + F_{zp}} + COP_i \cdot \frac{F_{zi}}{F_{zi} + F_{zp}}$$

Knee angle was chosen because it is directly related to the control parameters of the prosthesis knee and the tuning goal of the tuning algorithm. Hip angle was chosen since it is the next closest intact joint to the prosthesis and would be expected to coordinate its behavior with the knee (i.e. a *local* measurement of coordination). COM-COP angle was chosen because it is a balance metric [86] that captures performance of the whole body and both limbs (i.e. a *global* measurement of coordination).

To capture the change of these angles over time during the stance phase, the slope of each angle curve was estimated from the time of minimum to maximum knee angle stance flexion. This metric was chosen to best capture the largest change in the knee angle profile over time and include both spatial and temporal information and is referred to as hip velocity, COM-COP velocity, and knee velocity.

To explore the impact of the tuning iterations on the resulting *local* (hip) and *global* (COM-COP) variables, we averaged the hip and COM-COP velocity every four steps (one iteration) and then sampled the highest five or lowest five iterations to define the maximum and minimum values, respectively, during tuning. This range was investigated to measure the effect of the tuning on these local and global measurements.

Statistical Tests

To test the effect of the different knee control parameters, two, one-way ANOVAs comparing the maximum and minimum values for the hip and COM-COP velocities with participant as a random effect was conducted. Additionally, a within-participant analysis was run to explore individual effects. To explore the relationship between the knee control (estimated by knee kinematics) and the local and global measurements, a linear regression model was used. The significance level was set at $\alpha = 0.05$, and analyses were performed using JMP (JMP, SAS Institute, Cary, NC, USA) and MATLAB (MathWorks, Natick, MA, USA).

4.4 Results

Effect of Tuning Iterations on Local and Global Measurements

Figure 2 shows the top five maximum and minimum values of the hip and COM-COP velocities during tuning averaged over four steps (with the same control parameters) for a total sample size of 20 steps. Both hip and COM-COP velocity had a significant difference between the maximum and minimum values indicating that the tuning iterations have a significant effect on both the local (hip) and global (COM-COP) measurements ($p < 0.001$ for hip and $p < 0.001$ for COM-COP). The average range across participants for the hip was 15.7 and the average range across participants for the COM-COP was 11.7. When exploring the individual differences within participant, all participants were significantly different for both hip and COM-COP ($ps < 0.0416$) except for TF03 – there was not a significant difference between the COM-COP velocity minimum and maximum ($p = 0.7117$).

Relationship Between Knee Kinematics and Local or Global Measurements

To further explore if these differences in the local and global measurements are directed related to the differences in knee control itself, the knee angle velocity was plotted versus the hip or COM-COP velocity (Figure 3). The kinematics were used opposed to the knee control parameters themselves because the 12 impedance parameters have multiple solutions resulting in the same knee angle profile, so the resulting kinematics were used for analysis. As seen in the figure, the knee vs. hip velocity plot shows a significant moderate relationship between the kinematics of the knee and hip joints (R^2s > 0.455). However, when looking at the knee vs. COM-COP velocity plot, there is not a significant relationship between the two variables (R^2s < 0.030).

4.5 Discussion

The results of this preliminary analysis present a first glimpse into the effect of knee control on local and global variables during powered prothesis tuning. We revealed that both the *local* and *global* measurements were sensitive to changes in tuning iterations; however, this change cannot be directly related to the knee kinematics themselves in the case of the *global* variable (COM-COP velocity in this case). This indicates that we may be able to optimize using either local or global variables; however, the exact relationship between the control and *global* measures is not clear.

When tuning powered prostheses, choosing an objective function is important to ensure that the tuning procedure is adapting to a variable that is not simply noise (i.e. the variable remains constant regardless of control). In the case of a global variable such as COM-COP velocity, it is not clear how adjusting the knee angle profile via the control parameters will affect the resulting performance. The intact limb and upper body and trunk/pelvis may be compensating and minimizing the resulting performance. In this case, the effect may not propagate from the *local* to the *global* variables and be “washed out”.

For example, when fine-tuning the prosthesis parameters, the adjustments in the control parameters are relatively small and may not be perceived by a global variable (as seen in Figure 3). However, these fine adjustments will likely still impact the local joint. This may be beneficial for protecting the adjacent intact hip joint in the case of individuals with an above knee amputation [87]. Even though small change in the knee kinematics may not be perceivable in the *global* sense when fine-tuning the parameters, the fine-tuning may still benefit the user.

It is noted that this study is not without limitations. Firstly, this was a preliminary analysis with four individuals with amputation. Previous studies have noted the importance of testing with the target population and not necessarily a simulated group especially when quantifying overall performance with a powered prosthesis [51]. More individuals with amputation should be recruited to further explore the relationship between the changing prosthesis and the user’s performance. In addition, here we only investigated two variables – hip velocity and COM-COP velocity. These variables were chosen because they are in the sagittal plane and both measurements of angle. Expanding the analysis to the medial-lateral plane or other kinematic/kinetic measurements may be advantageous to further explore the performance space.

4.6 Conclusions

This preliminary study aimed to begin the investigation into the impact of the changing control parameters on variables in the human-prosthesis system. Results indicate that both local and global measurements are sensitive to the updated prosthesis parameters across iterations of the tuning algorithm; however, one participant did not show a significant difference in the global metric when analyzing each participant separately. Additionally, when exploring the relationship between the knee profile itself and the hip and COM-COP responses over tuning, the COM-COP velocity did not have a strong relationship. There may be other factors in the human system (such as the intact limb or trunk) as captured in a *global* metric such as COM-COP angle that may adapt and change over time to show the change iteration-to-iteration, but the direct relationship between the control and the *global* metric is not well defined.

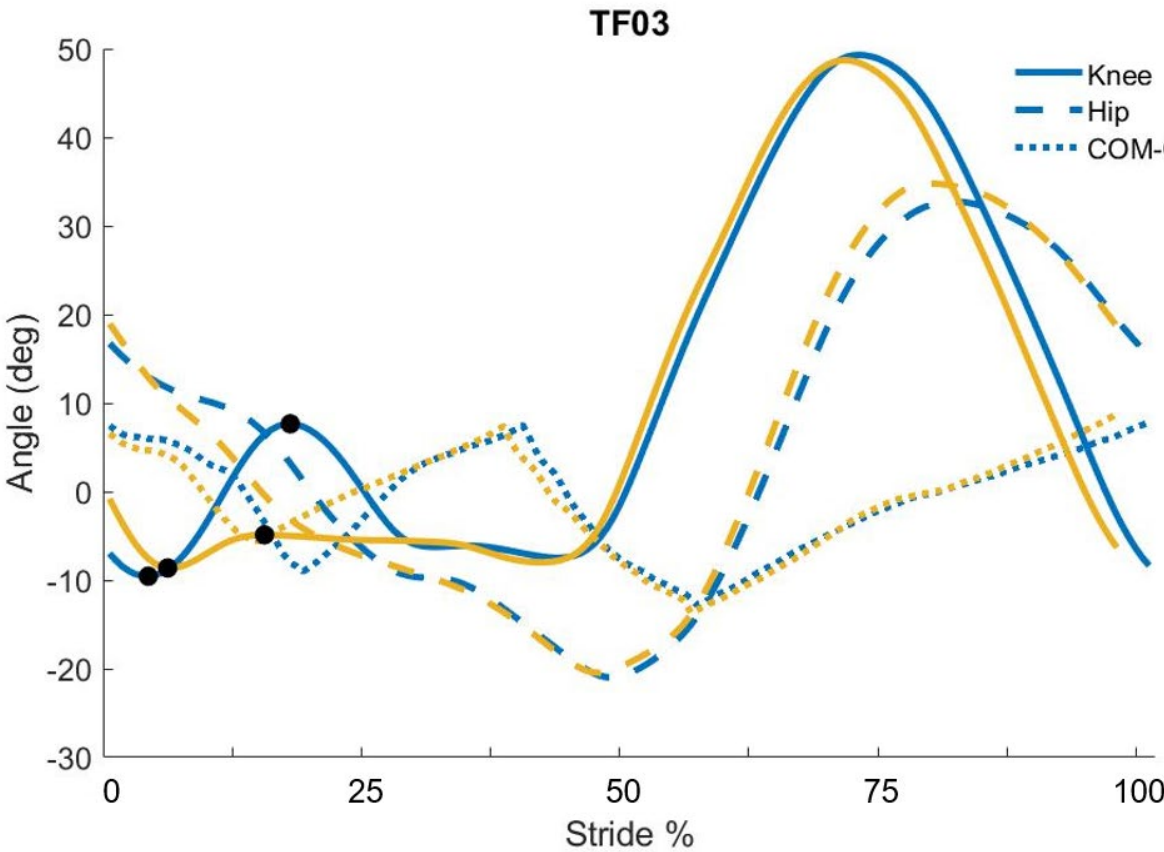

Figure 4.1: Knee, Hip, and COM-COP angle during two example strides (prosthesis-side heel strike to next prosthesis-side heel strike). The blue and yellow strides demonstrate two extremes observed in the knee flexion angle during the stance phase. The corresponding hip and COM-COP curves in early stance also show differences in response to the change in knee kinematics. To quantify this change, the timing of the min and max flexion during stance from the knee angle (noted as the black point dot characters) was used to segment the curves and the velocity was estimated by calculating the change over time between these two points.

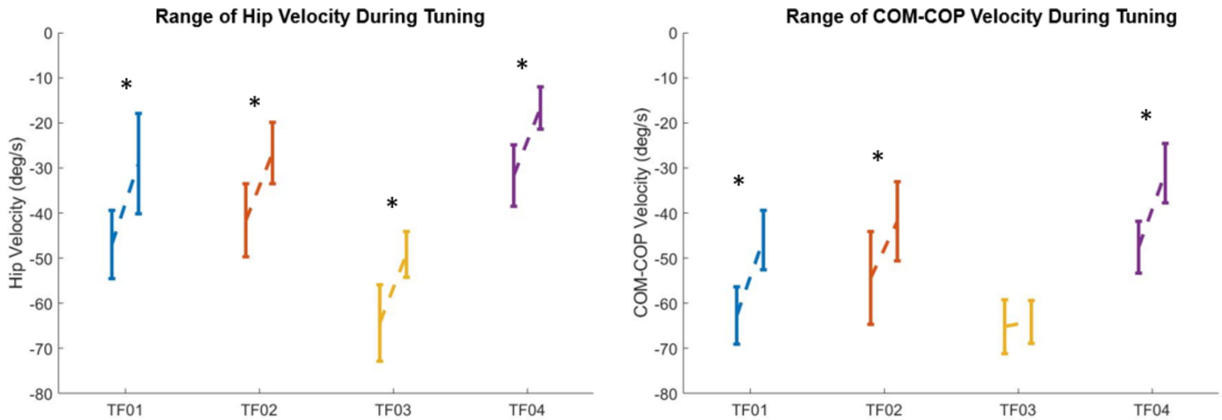

Figure 4.2: Range of hip velocity and COM-COP velocity during tuning. The maximum and minimum values were calculated from 20 steps corresponding to the five tuning iterations (comprised of 4 steps each) with the highest or lowest average hip or COM-COP velocity. In other words, we averaged the hip and COM-COP velocity every 4 steps to explore the effects of the prosthesis parameters for each iteration and then sampled highest five and lowest five iterations to define the range.

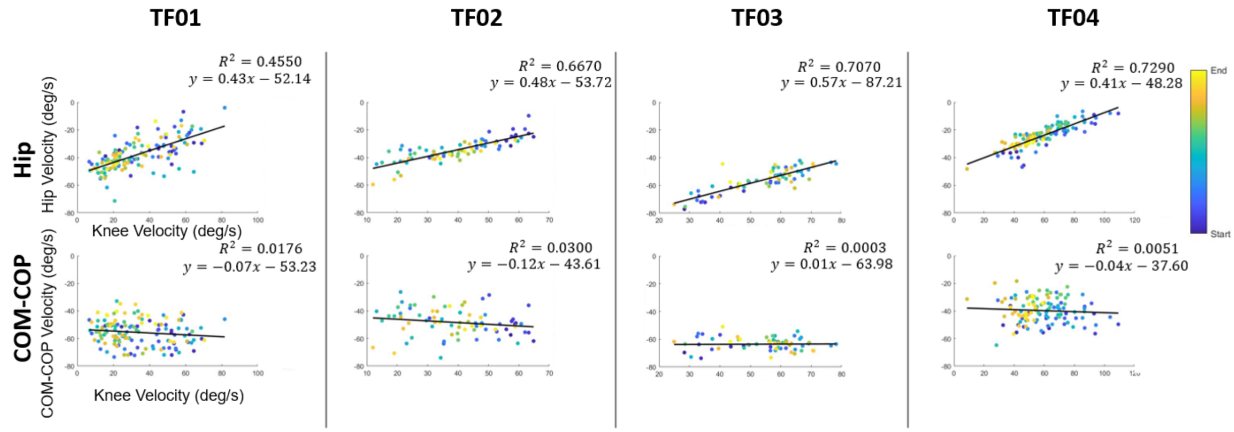

Figure 4.3: Knee velocity vs. Hip velocity during tuning (top) and Knee velocity vs. COM-COP velocity during tuning (bottom) for each participant. Each point is one step and the colors indicate the change in time during tuning from start (blue) to end (yellow). All participants observed a strong relationship between knee and hip kinematics and no relationship between knee and COM-COP kinematics.

CHAPTER 5: CONCLUSIONS, SIGNIFICANCE, AND FUTURE WORK

5.1 Conclusions and Summary of Findings

The results from Chapter 2 highlighted the need to investigate of human-prosthesis coordination at the local joint level. The preliminary analysis with five individuals with a transtibial amputation highlighted that even when walking with a powered ankle prosthesis, not all participants experienced positive net ankle work. In other words, the device did not successfully transfer it's added energy to the user. This may have been due to the timing of push-off and the user's limb position at push-off. Therefore, studying how the human and device coordinate and work together is imperative when designing future tuning and training paradigms.

In Chapter 3, an effort was made to improve the coordination between the prosthesis user and the powered prosthesis by combining visual feedback to the user with autonomous prosthesis tuning. This study set the groundwork for the future of tuning wearable robotics. From a technology point of view, adding feedback to the user may be able to speed up the tuning algorithm, assuming the feedback does not destabilize the user. Additionally, from the user's point of view, visual feedback may allow for performance to be maintained after feedback is removed, at least in the short term. Although these effects were modest, this is the first study, to our knowledge, to combine prosthesis tuning and user training and there is room to expand different combinations of the control goals for each system in the future.

Finally, Chapter 4 further explored how the tuning parameters influence the behavior of the prosthesis user. Preliminary analysis from four individuals with an amputation revealed that both local and global measurements could be used for optimization – they both changed significantly with respect to the knee control parameters. However, one of the four participants did not experience a significant difference in their global measurement (in this case, COM-COP velocity). In addition, it is not clear if the difference is directly related to the knee control alone because the relationship between the knee kinematics and COM-COP kinematics was not strong – the user's other joints on the intact limb or upper body may influence the global variable.

5.2 Clinical and Engineering Significance

In the standard clinical practice, individuals with an amputation are prescribed a prosthesis, they work with their prosthetist to achieve proper alignment and socket-fit, and then participate in physical therapy to learn how to walk with their prosthesis. However, when prescribed a new powered prosthesis, there is an added layer of tuning the control parameters to customize them to the user as well as training the user to coordinate their control system with that of their prosthesis. Therefore, understanding how to tune the device quickly and in a way to maximize performance and understanding how to train the user to use their device to get the most out of it is vital for the clinical success of these powered prostheses.

For example, in the case of the powered ankle prosthesis used in Chapter 2, the device has onboard sensors to measure net ankle work per step and display that ankle work to the prosthetist tuning the device, but our results demonstrated that the work is not necessarily transferred to the user. To assist this clinical process, something like a simple IMU sensor on the prosthetic shank could add additional information about the user's limb position to further adjust the timing parameter or instruct the user to adjust their timing of push-off. This would be a relatively inexpensive and quick method to measure at the device level how the human and prosthesis are working together.

As another example, this clinical practice could be further automated with the use of autonomous tuning algorithms and feedback to the user similar to that used in Chapter 3. The clinician could save time by letting the algorithm tune the initial parameter set while the user is in a steady-state as determined by the feedback goal and then later work with their patient to determine if the final parameters do indeed result in the best performance possible.

5.3 Future Work

As stated above, the work presented here is a first glimpse into better understanding how the human and prosthesis or wearable robot interact and coordinate with one another. Future studies with more individuals with an amputation are needed to corroborate the results from the preliminary study in Chapter 2. In addition, it would also be beneficial to add more global measures, such as metabolic cost, during the data collection although it is noted this is no small feat given the amount of time, equipment, and participant willingness to collect this data. Future studies are also encouraged to explore more combinations of prosthesis tuning goals and user performance goals – will tuning to optimize stance time symmetry coupled with stance time symmetry feedback result in lasting improvements in stance time symmetry? Finally, future work needs to also future explore the effect of changing prosthesis or wearable robot parameters on the user – when the device is changing, which parameters are sensitive to the change? These future studies will hopefully allow for the full potential of powered prostheses and wearable robots to be realized to allow individuals to reach their full potential.

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APPENDIX

Appendix A: Supplementary Information for Chapter 2

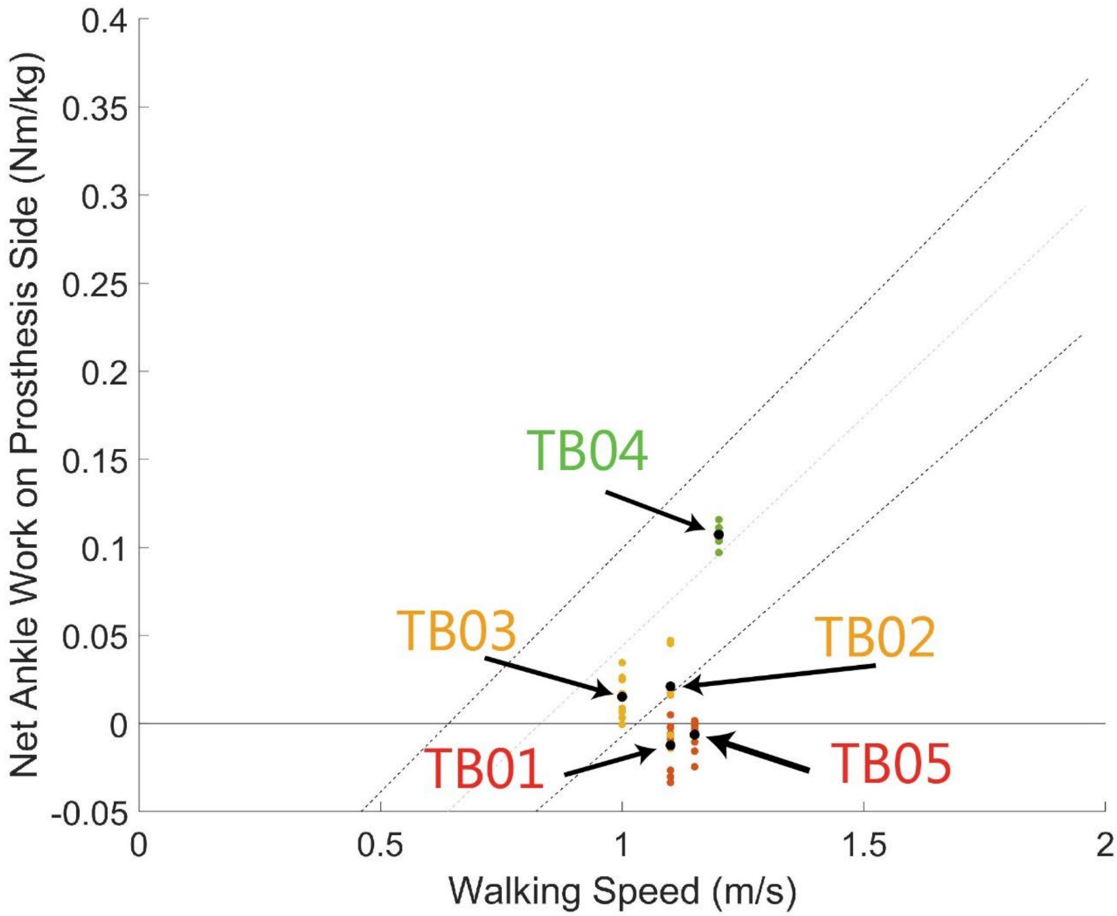

Figure S1: Net prosthesis side ankle work for each individual stride as a function of treadmill speed. Each colored point character corresponds to the net ankle work calculated for one stride via inverse dynamics. Each black point character represents the mean of 10 strides used for each participant in this study. The black dashed lines represent the range of net ankle work typical of a biological ankle in walking and the gray dashed line represents the midpoint. The point characters and participant labels are colored by their average net BiOM work, compute by inverse dynamics. These colors are used throughout the results. Green indicates net ankle work in the upper 50% of the range, yellow indicates net ankle work in the lower 50% range, and red indicates net ankle work below the expected biological net ankle work.

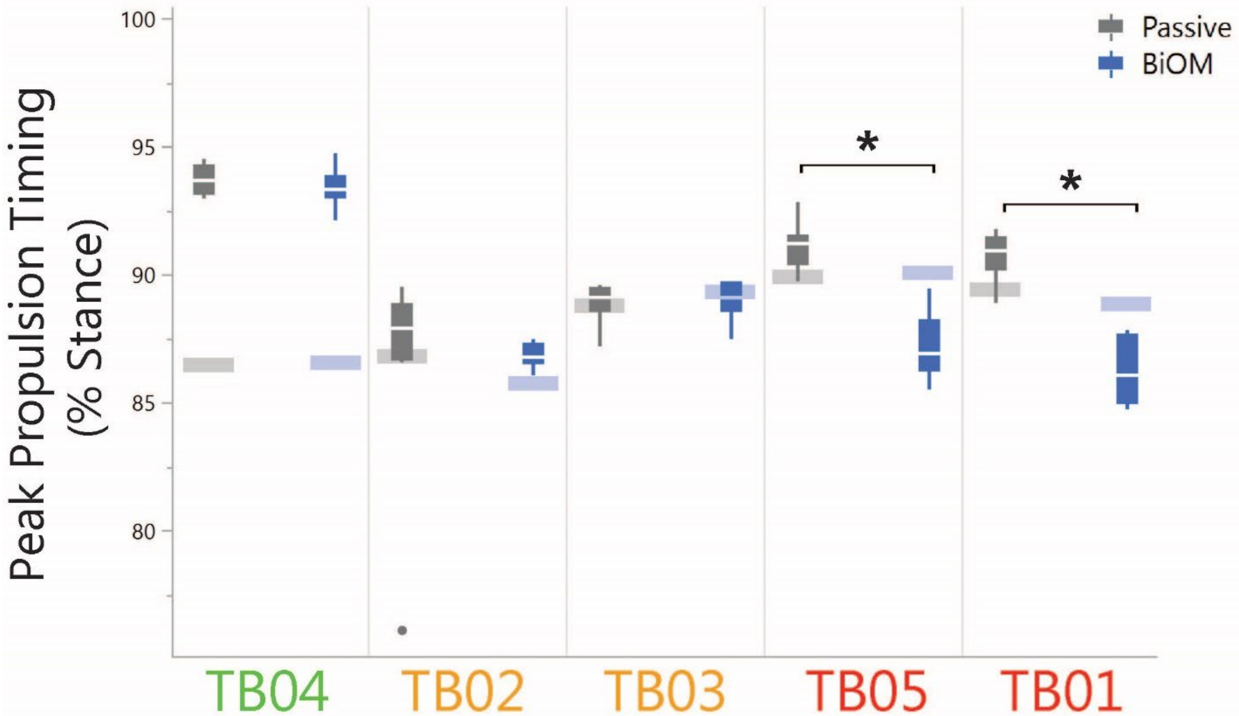

Figure S2: Boxplots of timing of peak propulsion with their prosthesis as a percentage of stance time for each participant. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Shaded bands indicate the timing of peak propulsion on the intact side. Outliers are 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. Additionally, TB01 and TB05 were more symmetric in their timing symmetry with their passive and moved away from symmetry with the powered prosthesis while the other three participants maintained their symmetry across the two conditions.

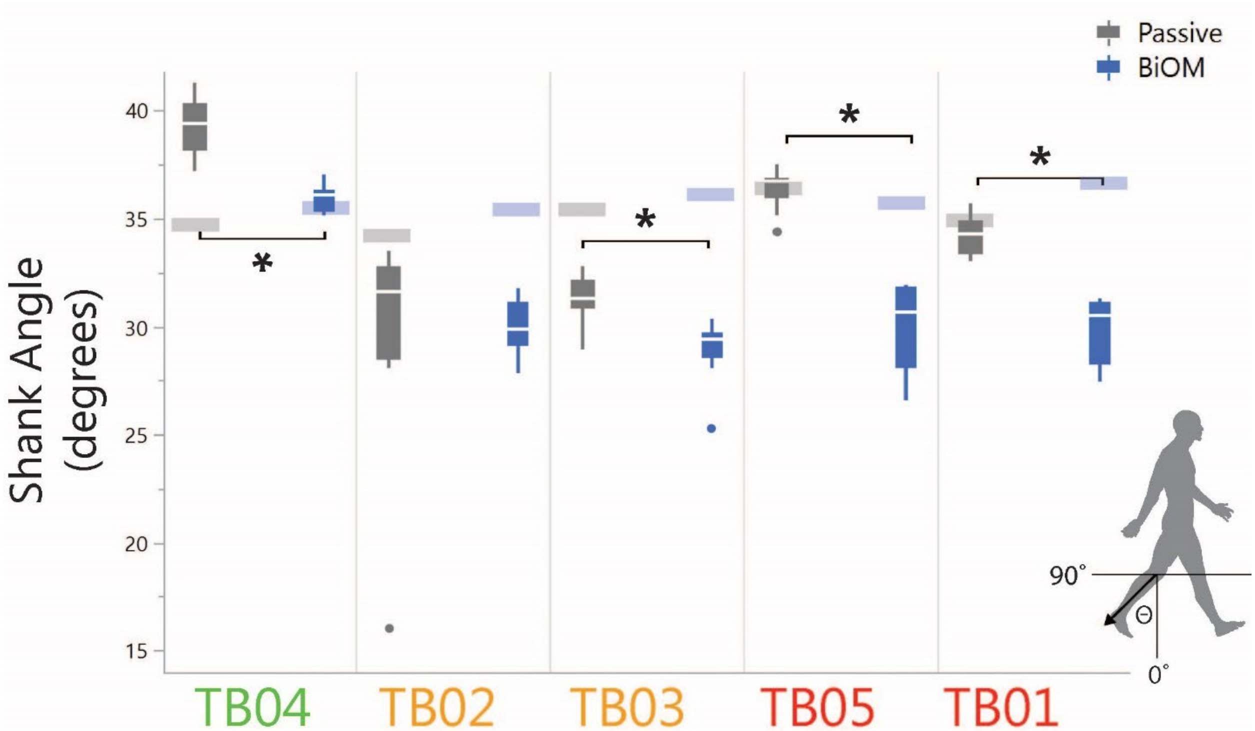

Figure S3: Boxplots of prosthesis side shank angle for each participant. Colors indicate the device condition: passive prosthesis (gray) and powered prosthesis (blue). Shaded bands indicate the timing of peak propulsion on the intact side. Outliers are defined as 1.5 times the interquartile range from 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. Interestingly, TB01 and TB05 were symmetric with their passive prosthesis and moved away from symmetry when walking with the powered prosthesis. TB04 on the other hand, experienced better symmetry in shank progression during push-off with the powered prosthesis.

Table S1: Net Work of the ankle, knee, hip, and composite limb work for the prosthesis side for both the BiOM and Passive conditions. Values are averaged over 10 steps.

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