Improving Amputee Endurance over Activities of Daily Living with a Robotic Knee-Ankle Prosthesis: A Case Study

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Abstract— **Robotic knee-ankle prostheses have often fallen short relative to passive microprocessor prostheses in time-based clinical outcome tests. User ambulation endurance is an alternative clinical outcome metric that may better highlight the benefits of robotic prostheses. However, previous studies were unable to show endurance benefits due to inaccurate high-level classification, discretized mid-level control, and insufficiently difficult ambulation tasks. In this case study, we present a phase-based mid-level prosthesis controller which yields biomimetic joint kinematics and kinetics that adjust to suit a continuum of tasks. We enrolled an individual with an above-knee amputation and challenged him to perform repeated, rapid laps of a circuit comprising activities of daily living with both his passive prosthesis and a robotic prosthesis. The participant demonstrated improved endurance with the robotic prosthesis and our mid-level controller compared to his passive prosthesis, completing over twice as many total laps before fatigue and muscle discomfort required him to stop. We also show that time-based outcome metrics fail to capture this endurance improvement, suggesting that alternative metrics related to endurance and fatigue may better highlight the clinical benefits of robotic prostheses.**

I. Introduction

In order to advance robotic leg prostheses out of research laboratories and into mainstream clinical practices, the field must demonstrate their benefits over the current standard of care for people with above-knee (AK) amputation, specifically passive microprocessor knee prostheses (MPKs). Robotic prostheses differ from MPKs in that they have motors at the joints capable of providing net positive work [1], which is especially important during sit-to-stand, ramp ascent, and stair ascent [2]–[11]. Evidence is needed to convince reimbursement agencies, who often pay a large proportion of prosthesis device and servicing costs [12], that the additional cost of robotic prostheses is justified by improvements in patient well-being and potentially reduced medical costs over the long-term [13]. Unfortunately, much of the powered prosthesis literature does not focus on clinical outcomes [14], motivating the need for more outcome-focused investigations.

One area where robotic prostheses have the potential to excel relative to MPKs is in user ambulation endurance. Endurance is an important component of well-being, as it impacts a person's ability to participate in the community for a sustained period of time. It is well understood that users of MPKs require more energy to walk compared to able-bodied people [15], [16], which likely leads to a comparably earlier onset of fatigue. Further, due to the lack of net positive joint work in an MPK, the user must compensate by overexerting other muscles, such as in the back and contralateral leg (*e.g.,* vaulting and hip-hiking) or in the hip joint (*e.g.,* circumduction) [17]. These compensations can lead to overuse injuries, pain, osteoarthritis, and other comorbidities [18]–[20] that may also limit endurance. Many articles cite these limitations of MPKs as motivation for developing robotic prostheses [21]–[28].

Perhaps one of the most comprehensive comparisons between robotic and MPK knees was presented by Hafner et al. in 2015 [29]. In this 14-month randomized crossover trial of 12 individuals with AK amputations, three classes of prostheses—mechanical knees, an MPK (Össur Rheo Knee II), and a robotic knee (Össur Power Knee II)—were compared using standard indoor clinical timed-based tests (timed up-and-go, timed stair ascent, etc.), outdoor walking course tests, surveys, and fall incidence. The robotic knee showed significantly worse results than the MPK in the indoor timed-based tests. During the half-mile outdoor walking test, which included uneven terrain, steep ramps, and stairs, no significant differences in time or perceived exertion were detected between knee conditions. Finally, fall incidence was highest with the robotic knee and many participants requested to end the robotic knee condition early, citing prosthesis weight, noise, increased pain, unexpected behavior, and difficulty changing ambulation modes as primary factors. When given the chance to keep the robotic knee at the end of the study, no participants elected to do so.

These results contrast the field's common assumption that robotic prostheses will improve endurance and decrease pain for people with an AK amputation. Particularly with respect to the outdoor walking course, one would assume that the additional energy provided by the prosthesis would reduce the person's energy expenditure and allow them to complete the course faster. We hypothesize that these negative results are more indicative of limitations of the early-generation robotic prosthesis used in the study and in the study design itself, rather than of robotic prostheses in general.

Specifically, we suggest three primary explanations for the negative results: 1) *Inaccurate/Cumbersome High-Level Classification:* The article noted that users had difficulty getting the prosthesis to change ambulation modes when desired. This may have limited their ability to quickly change modes during the obstacle course and resulted in

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both a slower time and lower satisfaction with the device. 2) *Discretized Mid-Level Control:* Due to the phase and task discretization likely used in the controller of the robotic knee [1], the prosthesis behavior may not have been optimized for every scenario encountered during the walking course. Because the appropriate joint kinematics and kinetics vary with walking speed and incline [10], [11], there may have been instances during the course when the robotic knee was at the wrong angle, added the wrong amount of energy, or mistimed energy injection. 3) *Insufficiently Difficult Task:* During the obstacle course in [29], the participants were instructed to "walk at a comfortable pace" and allowed to rest "as long as needed" between repetitions. These instructions may have not sufficiently challenged the participants to the point of fatigue with the MPK knee, hiding the potential endurance benefits of the robotic knee.

Our research group has aimed to mitigate the discrete mid-level control limitations by developing continuous, phase-based controllers for robotic knee-ankle prostheses. These controllers have enabled walking across variable speeds and inclines [30]–[33], stair ascent and descent at different step heights [34]–[36], and transitioning to and from sitting [37], [38]. To develop these controllers, we leverage large datasets of able-bodied (AB) individuals performing various activities of daily living (ADLs) [9]–[11]. Through convex optimization, we fit continuous models to the datasets, allowing our controllers to produce appropriate joint impedance and kinematics across continuums of gait phase, speed, and incline. In this work, we integrate our isolated ADL controllers into a unified controller capable of sit-to-stand motions, walking at variable speeds and inclines, and stair traversal across variable step-height staircases. We hypothesize that, by using our new continuous mid-level control approach, we will be able to demonstrate improved endurance with a robotic knee-ankle prosthesis.

In this case study, we focused on a young, active male participant with an AK amputation of his left leg. He had moderate tendinitis in his right quadriceps and right hip dysplasia, making him an excellent candidate to benefit from a robotic prosthesis. We challenged the participant to complete rapid and repeated laps of an ADL circuit comprising sitting, standing, walking on level ground and ramps, and climbing and descending stairs. While overall speed was not increased with our robotic prosthesis, the participant was able to comfortably complete over twice as many total laps with it compared to his regular MPK prosthesis. These findings support the hypothesis that, given appropriate prosthesis control and a sufficiently challenging task, a robotic prosthesis can improve a user's endurance.

II. Methods

A. Multi-ADL Prosthesis Controller

In this article, we present a multi-ADL prosthesis controller that enables sit-to-stand, stand-to-sit, walking at variable speeds and inclines, and stair traversal at varying step heights. This controller, which integrates controllers from our research group's preliminary work [33]–[35], [38], uses stance-phase impedance control and swing-phase kinematic control. By using impedance control during stance, the controller can produce kinematic and kinetic profiles that resemble AB data. Kinematic control during swing ensures foot clearance and provides the user a sense of volitional control over foot placement. We implemented the controller on the quasi-direct drive knee-ankle prosthesis presented in [25], which has been validated for accurate open-loop impedance control and high-bandwidth position control.

The controller comprises three high-level modes: $\mu \in \{\text{Sit/Stand, Walk, Stair}\}$. For this case study, the control mode was manually selected by a researcher. While many options for classifying the appropriate control mode have been proposed [39]–[43], we elected to perform manual classification in order to isolate the effects of the mid-level control. Likewise, while the controller can estimate user's task (walking speed and ground incline/step height) in real-time with reasonable accuracy [32], [33], these were also selected by the researcher to reduce the number of uncontrolled experimental variables. A time-based interpolation between modal torque commands was used to smooth transitions between control modes over 100 ms. This time window was heuristically chosen during controller development to balance response time with jerk.

Each control mode calculates a phase variable s that defines the user's progression through either the gait cycle (for walking and stairs) or the motion (for sitting and standing). We calculate s based on algebraic mappings of the user's thigh angle θ_{th}, giving them indirect volitional control over the prosthesis's behavior. The parameters used to calculate the phase variables adjust to the given tasks, allowing them to appropriately parameterize the motion regardless of the varying thigh kinematics associated with different speeds, inclines, and step heights [10], [11].

While the user is in the stance phase (which includes the entirety of Sit/Stand mode), the knee and ankle torques τ are calculated based on the joint angles θ and velocities $\dot{\theta}$ using impedance control:

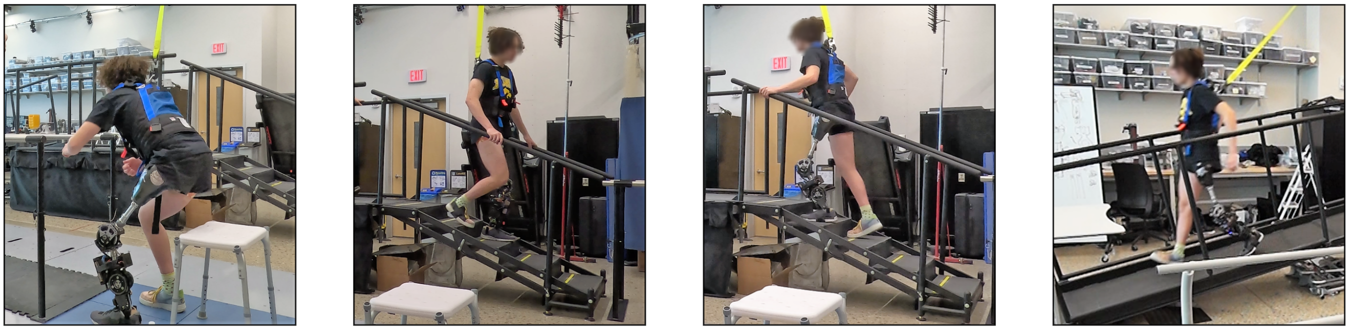
$$\tau = K(s,\chi,\mu)\left[\theta_{\text{eq}}(s,\chi,\mu) - \theta\right] - B(s,\chi,\mu)\dot{\theta}, \quad (1)$$

where the stiffness K, damping B, and equilibrium angle θ_{eq} are functions of phase s, task χ, and mode μ. The task variable χ contains the walking speed ν and the incline γ, and it is only applicable in Walk and Stair modes. Moderate values of γ represent the slope of a ramp in Walk mode whereas larger magnitude values of γ indicate the steepness of a staircase in Stair mode.

Similarly, we use kinematic control during swing-phase, where desired joint angles θ_{d} and velocities $\dot{\theta}_{\text{d}}$ are functions of s, χ, and μ. Joint torques are calculated through a proportional-derivative kinematic controller as

$$\tau = k_{\text{p}}[\theta_{\text{d}}(s,\chi,\mu)+\theta_{\text{s}}(z)-\theta]+k_{\text{d}}[\dot{\theta}_{\text{d}}(s,\chi,\mu)-\dot{\theta}]-\beta\dot{\theta}, \quad (2)$$

where k_{p} and k_{d} are proportional and derivative gains, respectively. An additional term with a constant viscous damping coefficient β helps improve stability of the low-impedance drivetrain of the prosthesis (see [25], [32]). An



(a) (b) (c) (d)

Fig. 1. Photos of the AK amputee participant using the robotic prosthesis while (a) standing from the chair, (b) descending the stairs, (c) ascending the stairs, and (d) descending the ramp throughout lap 13 during the second day of experiments with the robotic prosthesis.

additional joint offset $\theta_s(z)$ can supply additional flexion while climbing stairs if an ultrasonic distance measurement z indicates that a toe stub may be likely, as proposed in [35]. For more detailed information on the toe-stub prevention, phase calculations, impedance models, and kinematic models for each mode, please reference [33]–[35], [38].

B. Endurance Experiments

We enrolled one participant with an AK amputation of his left leg for this case study (sex: male, age: 19 years, mass: 71 kg with his MPK prosthesis, etiology: congenital). His every-day prosthesis is an Ottobock Genium X3 MPK with an Össur Pro-Flex LP foot, and he suffers from chronic tendinitis in his right quadriceps and hip dysplasia. The participant wore a ceiling-mounted safety harness while wearing the robotic prosthesis (Fig. 1), and the experimental protocol was approved by the Institutional Review Board of the University of Michigan (HUM00166976). This participant was an experienced user of the robotic prosthesis and had participated in prior studies.

A custom ambulation circuit was designed comprising tasks commonly encountered in a home or office setting, including sitting, standing, stair climbing, and variable-incline walking (Fig. 2). Each lap of the circuit consists of the following sequence of ADLs: First, the participant begins sitting in a chair (54.0 cm seat height). Then, the participant stands up and walks 7.5 m, turns right, and walks another 2.7 m to the base of a ramp. There, the participant turns right again and ascends the ramp (3.7 m, 10 deg slope). The participant then walks across the top of a level platform for 2.5 m before descending 5 stairs with a step height of 12.7 cm. Once off the stairs, the participant walks another 2.5 m before turning around and sitting in a second chair (44.5 cm seat height). Finally, the participant returns to the original chair by completing the same activities in the opposite sequence (this time including stair ascent and ramp descent). In total, a single round-trip lap of the circuit includes two sit-to-stand and stand-to-sit sequences, three bouts of level walking, one ramp ascent, one ramp descent, one stair ascent, and one stair descent for a total of 41.3 m.

The participant completed four data collection sessions for this case study; two were completed using the participant's every-day MPK prosthesis (PAS) and two were completed with the robotic prosthesis (ROB). The experimental protocol was identical for each prosthesis, and the participant was allowed to practice on the circuit with each prosthesis prior to the experiments until he felt comfortable. There were five days between data collections with the PAS device while only two days between data collections with the ROB device due to scheduling constraints, thus providing more recovery time between PAS sessions.

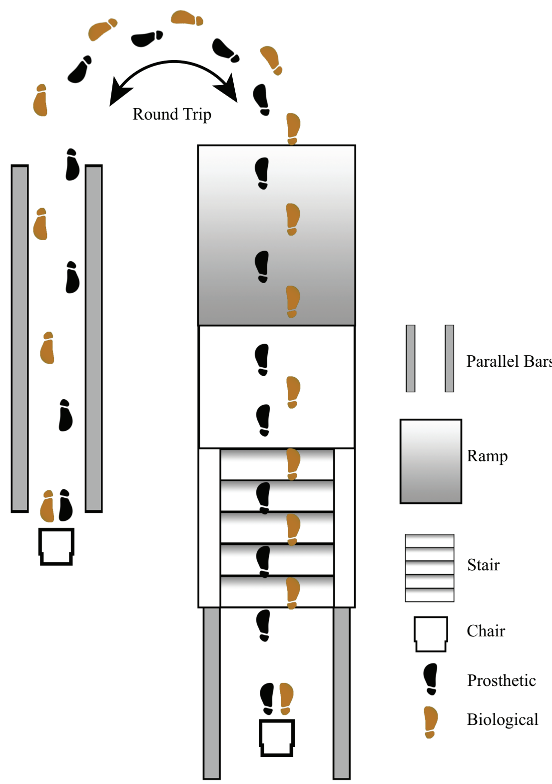


Fig. 2. Diagram of the ambulation circuit, where the footprints depict the approximate return path of one round-trip lap. During each lap, the participant began in a chair (left), stood and walked across the room to a ramp, ascended the ramp, crossed a level platform, descended the stairs, walked to a different chair (right), and sat in the chair before returning to the start in the reverse sequence. The total distance covered in one lap of the circuit is 41.3 m. Note that the diagram is not drawn to exact scale.

During each experimental session, the participant was asked to complete up to fifteen laps of the circuit "as fast as comfortably possible" with either the PAS device or the ROB device. The participant was allowed to end the session prematurely if he felt significant discomfort or excessive fatigue. The experiment was designed to be sufficiently challenging such that the effects of fatigue became evident. Water was provided, and the participant was encouraged to take rest breaks at any point if needed.

Four video cameras were mounted throughout the laboratory to allow continuous recording of the experiments. Data from the robotic leg were also recorded, including joint angles, joint torques, and ground reaction force measurements. Videos of both the PAS and ROB experiments are available for download in the supplementary materials.

C. Data Processing

The experiment videos were time-aligned and segmented into the different ADLs for each lap. Each segment was classified as sitting (SIT), standing (STA), ramp ascending (RA), ramp descending (RD), level walking (LW), stair ascending (SA), stair descending (SD), 180 degree turning (TUR), or resting (RST). For the locomotion ADLs (LW, RA, RD, SA, SD), transitions between segments occurred when the participant heel-struck on a different part of the circuit. For example, a transition from LW to SA was defined as when the participant's foot contacted the first stair after walking. Non-locomotion ADLs (SIT, STA, TUR, RST) were segmented by visual inspection of the motion's completion. Any rest time between laps was included with the following lap's total time (*i.e.,* the timer for lap i started as soon as the participant sat down at the end of lap $i-1$). The same researcher performed the segmenting for both the PAS and ROB experiments to control for researcher dependence.

The data recorded from the robotic leg were further segmented into individual gait cycles to examine the kinematics and kinetics of each joint. HS and TO events were detected based on ground reaction force measurements from a load cell integrated in the foot of the ROB prosthesis. Each gait cycle was classified based on its parent segment defined by the video segmenting above.

III. Results

The participant demonstrated initially faster lap times with the PAS prosthesis, but quickly fatigued. Fig. 3 shows the participant's lap times over the course of each experimental session for both the PAS and ROB devices. We fit exponential functions to each data series to visualize the fatigue trends, yielding exponential growth in lap time for the PAS device but nearly constant lap times for the ROB device. With the PAS device, the participant completed seven laps on Day 1 and only three laps on Day 2. With the ROB device, the participant completed eleven laps on Day 1 and thirteen laps on Day 2. In all four cases, the participant cited quadriceps discomfort as his reason for stopping before reaching the fifteen-lap goal. Though the fastest lap times were completed with the PAS device, the participant quickly fatigued and

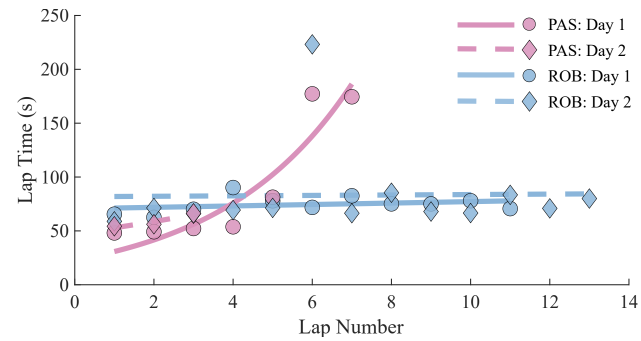


Fig. 3. Fatigue plots showing the duration of each lap of the circuit as a function of lap number for each experiment day along with exponential fit lines. While the PAS case started faster, the participant rapidly fatigued and took lengthy breaks (Day 1) or ended the experiment after only 3 laps (Day 2). In the ROB case, the participant held a consistent pace for 11 (Day 1) and 13 (Day 2) laps. Note that lap 6 on ROB Day 2, the participant took a lengthy break to re-don his prosthetic liner due to an air pocket forming.

began taking long rest breaks between laps. Contrastingly, the lap times with the ROB device remained fairly constant across a much greater number of trials.

One outlying data point occurred during lap 6 of ROB Day 2 (Fig. 3). During this lap, the participant took a long break to re-don his prosthetic liner due to an air pocket forming between it and his residual limb. We categorized this time as resting, though it was not motivated by fatigue. However, because the participant likely did recover during this period, we chose to keep the data point for completeness.

Fig. 4 depicts the time spent performing each ADL during each lap with the PAS and ROB devices based on the video segmentation. In this plot, experimental days were lumped together. On average, the participant was noticeably faster with the PAS device for LW, SA, SD, and TUR. Very similar average speeds (within 0.25 s) were observed between the PAS and ROB devices for SIT, STA, RA, and RD. Excluding the anomalous data point for re-donning his prosthetic liner, the participant took two rest breaks with his PAS device that were much longer than any taken with the ROB device, suggesting either more fatigue or muscle discomfort.

Finally, the ROB device produced biomimetic kinematic and kinetic behavior for each ADL. The researcher triggered the mode changes at the appropriate moments with no noteworthy errors, enabling the participant to smoothly transition between sections of the circuit. We averaged the joint angle and torque trajectories across representative non-transition strides for each ADL, which are shown in Fig. 5 along with normative AB trajectories from the data in [9]–[11]. As in our isolated ADL controllers [33], [34], [38], the integrated controller produced highly biologically-similar trajectories, suggesting that the prosthesis provided appropriate torque assistance and motion across the ADLs.

IV. Discussion

A. Time-Based Metrics

Across all ADLs, the ROB device did not outperform the PAS device in the time-based metrics (Fig. 4), similar to the findings observed in [29]. Though the participant completed

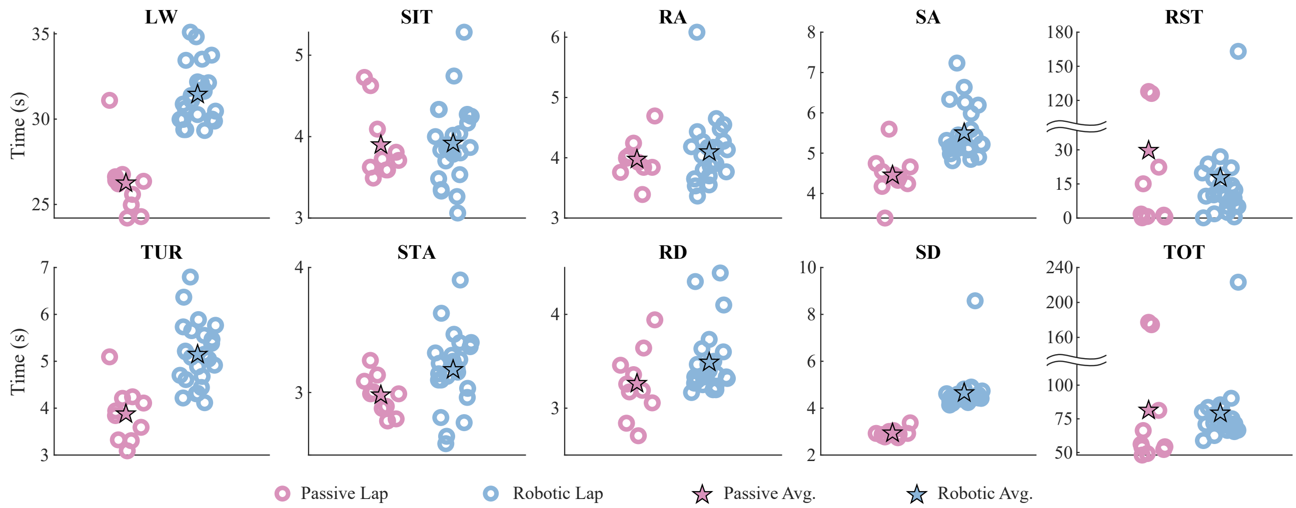


Fig. 4. Swarm charts showing the time spent performing each ADL (SIT sitting down, STA standing up, RA ramp ascending, RD ramp descending, LW level walking, SA stair ascending, SD stair descending, TUR turning, RST resting, and TOT total) during each lap across all experimental sessions. Circles denote individual lap times, and stars denote the average for both the passive prosthesis (PAS) and robotic prosthesis (ROB). The vertical axes for RST and TOT are broken to better visualize the data across large magnitude differences.

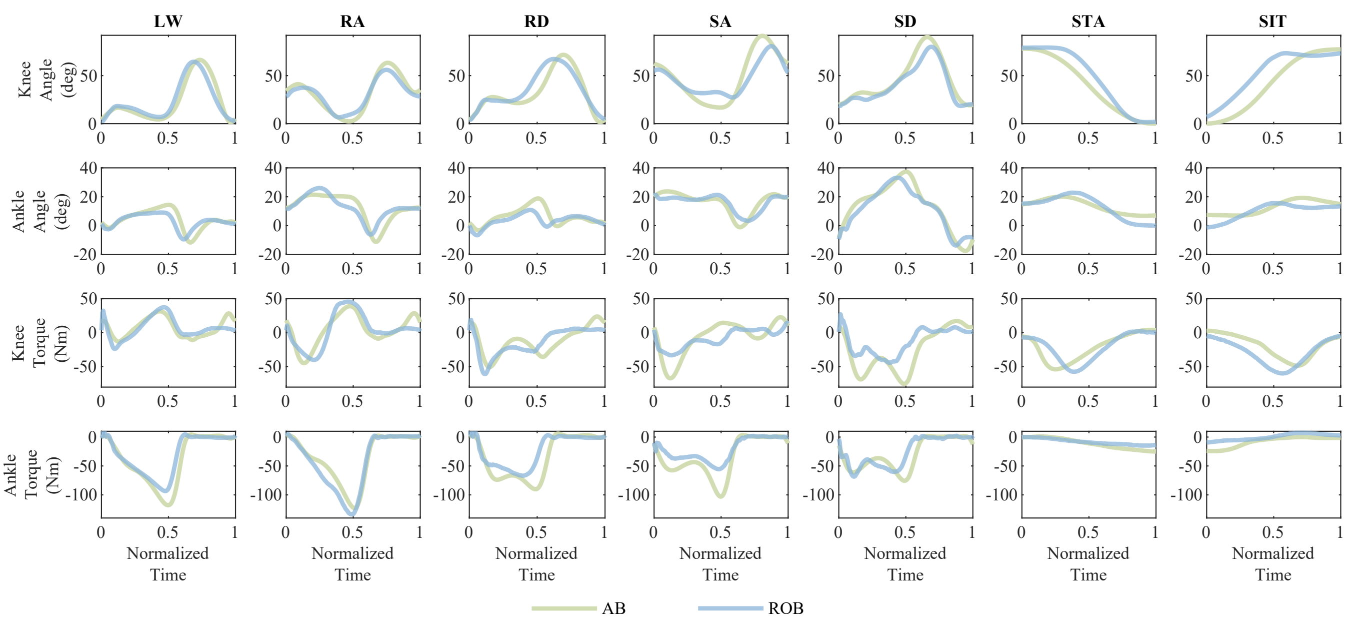
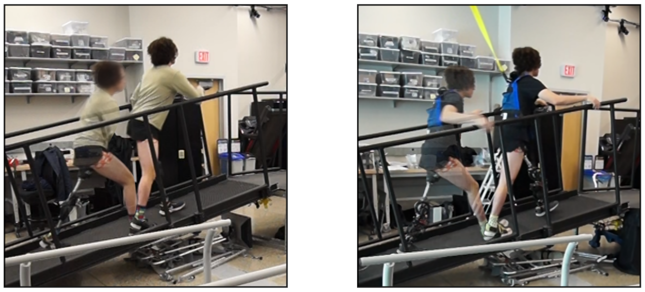


Fig. 5. The ROB device's mean kinematic and kinetic trajectories for representative non-transition strides of each ADL (LW level walking, RA ramp ascending, RD ramp descending, SA stair ascending, SD stair descending, STA standing up, and SIT sitting down) plotted against normalized stride time. Joint angles are defined relative to the proximal limb segment and positive in flexion/dorsiflexion. The resemblance to the able-bodied (AB) trajectories from [9]–[11] (scaled by the participant's weight) suggests appropriate mid-level control behavior.

some ADLs (SIT, STA, RA, and RD) in comparable amounts of time with both prostheses, he was noticeably faster with the PAS device for LW, SA, SD, and TUR. This reduction in speed with the ROB prosthesis occurred even with accurate high-level classification (performed by a researcher) and appropriate mid-level control, indicated by the biomimetic kinematic and kinetic similarity to able-bodied references (Fig. 5). It is unsurprising that the ROB device provided no speed benefit during LW, SD, SIT, RD, and TUR, as these ADLs require little to no net-positive joint work at the knee and ankle [6], and thus the PAS device may have been sufficient. However, PAS devices cannot produce biomimetic joint mechanics during STA, RA, and SA due to positive work requirements [1], [6], and it is therefore surprising that the biomimetic joint mechanics of the ROB device did not produce a time advantage for these ADLs. This result suggests that providing appropriate joint kinematics and kinetics alone is not in itself sufficient for improving participant speed.

The participant was also likely slowed by his reduced balance and agility with the ROB device, primarily due to its larger mass and relative unfamiliarity. This decrease in balance was apparent through the participant's increased use of the handrails for medial-lateral stability with the ROB



(a) (b)

Fig. 6. Photos of the participant performing RA with (a) the PAS device and (b) the ROB device at the instants of both biological heelstrike (shadowed) and prosthetic heelstrike (opaque). Notable differences with the PAS device include a decreased prosthetic step length, increased vaulting, and less natural upper body posture. These asymmetries in the PAS gait likely contribute to the fatigue felt during RA.

device (see the supplemental video and Fig. 1). During SA specifically, we observed that the participant ascended the stairs slower with the ROB device, with noticeable delays during late swing. We believe that lifting the heavier prosthesis and locating the foot at the proper location on the stair was especially difficult for our participant because of his short residual limb ($\sim$18 cm). This difficulty likely negated the speed benefits of the positive knee work, resulting in an overall slower SA with the ROB device.

B. Participant Endurance

Despite a lack of improvement in time-based metrics, we observed a sizeable increase in the participant's endurance with the ROB device compared to the PAS device (Fig. 3). This improvement suggests that the participant benefited from the ROB device in ways not captured by traditional time-based metrics. The participant specifically noted that RA was much easier for him with the ROB device than with the PAS device, and that this ADL was primarily to blame for the soreness in his quadriceps.

A visual comparison of the participant performing RA showed strong spatial gait asymmetry with the PAS device that was not present with the ROB device (Fig. 6). With the PAS device, the participant's prosthetic-side step length was very short, resulting in excess effort required by his contralateral leg to propel him up the ramp. The participant also demonstrated increased ankle vaulting and a less upright posture with the PAS device, likely also contributing to his increased fatigue. These visual observations suggest that quantitative kinematic symmetry measures may be more similar to able-bodied [44] with the ROB device. Future investigations with motion capture analyses should be performed to enable this comparison.

The effects of fatigue induced by the PAS device also persisted for several days after the experiment. On PAS Day 2, the participant noted that, although five days had passed since PAS Day 1, he still felt residual quadriceps soreness, explaining why he could only complete three laps. In contrast, the participant noted that he felt "fully recovered" after only two days between ROB data collection sessions. Though we did not explicitly test recovery time in this experiment, this result suggests an additional potential benefit of ROB prostheses that should be investigated further.

C. Limitations

A major limitation of our results is that we off-loaded the task of high-level classification to a researcher rather than having the prosthesis itself decide the proper control mode. While we did this in an effort to isolate the mid-level control problem, it is possible that, in practice, any errors made by the high-level classifier could offset the fatigue benefits of a ROB device and make a user still prefer an MPK. Future studies including online classification will be required to fully understand this trade-off. Additionally, a larger study is required to know if similar results hold across a broader population with AK amputations. For example, it may be true that endurance benefits of robotic prostheses are evident for people with chronic discomfort in the contralateral leg, such as our participant, but not for others. Finally, there are many other considerations of powered prostheses that this study did not address, such as noise and weight. While the prosthesis used in this study is very quiet, it is also quite heavy [25]. The added mass of ROB devices may be problematic in extended use and may have contributed to the prosthetic liner problems encountered by our participant on ROB Day 2. It is possible that a similar study with a different, lighter ROB prosthesis such as [26], [28] could show improved results in the timing-based metrics in addition to endurance benefits.

V. Conclusion

In this case study, we presented an integration of our continuous, data-driven prosthesis controllers that enabled a participant with an AK amputation to complete repeated, rapid laps of an ADL circuit. Although the participant was slower with the ROB device than with his PAS device for all ADLs, he completed over twice as many total laps with the ROB device before needing to stop due to fatigue and muscle discomfort. Our results suggest that there are endurance benefits to robotic prostheses, and that standard, timing-based clinical tests may be insufficient to capture them.

Acknowledgments

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