Impact of a Portable Autonomous Exoskeleton on Human Exertional Metrics During Sloped Walking

IMPACT OF A PORTABLE AUTONOMOUS EXOSKELETON ON HUMAN EXERTIONAL METRICS DURING SLOPED WALKING

by

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Abstract

The Biomechatronics group of the MIT Media Lab has developed an autonomous powered exoskeleton capable of providing a significant metabolic benefit to the user when walking on level ground. To gain a better sense of the practicality and versatility of this system, an assessment of metabolic and biomechanical performance on surfaces other than level ground is warranted. While other groups have evaluated aspects of exoskeleton performance on inclined terrain, no group has provided a direct quantitative comparison of sloped walking without any device to sloped walking wearing a powered autonomous ankle exoskeleton.

We hypothesized that a leg exoskeleton capable of providing a significant metabolic benefit on level ground through substantial positive mechanical power and minimal added distal mass would continue to provide a significant metabolic benefit on an incline, and that such a system would also provide no significant metabolic cost on a decline. In this study, the effect of the MIT exoskeleton on human exertional metrics was evaluated through measurement of respiratory metabolics while walking on an instrumented treadmill at +/- 15% grade. At the same time, electromyography was performed to evaluate differences in muscle activation.

In four healthy male subjects, the exoskeleton was found to impart no net metabolic benefit nor net cost to the user during inclined walking or declined walking, with a trend suggestive of metabolic benefit on an incline and a trend toward added metabolic cost on a decline. The lack of significance is likely the result of the low sample size, however these trends are likely attributable to the much higher overall metabolic cost of inclined walking relative to the amount of mechanical power applied by the exoskeleton, and a net destabilizing effect of powered plantarflexion during careful ramp descent. In demonstrating lack of cost on an incline and highlighting a possible added metabolic cost during powered decline walking, this study further characterizes both the the practical utility as well the limitations of an autonomous powered exoskeleton for military, recreational, rehabilitative, or other use.

Lay Summary

The terms “exoskeleton”or “exosuit” may be applied to a variety of devices worn on the human body which are designed to provide additional structural support, to augment load-carrying capability, to add strength or power across joints, or perhaps to replace muscle function in the case of aging, disease, or paralysis. Many exoskeletons aim to reduce the degree to which the human body must exert itself using its own native muscles to achieve a certain task, described as the “metabolic cost” of the activity. Simply donning most exoskeletons increases the baseline metabolic cost of walking due to the effect of carrying added weight on the body. Therefore, to demonstrate a true metabolic benefit relative to walking without any device at all, an exoskeleton design must reduce the metabolic cost of walking more than enough to account for its own added weight. While multiple designs have demonstrated a true metabolic benefit using large cable- or pneumatic-driven actuators tethered to a central lab power supply, these systems effectively limit the user to walking on a treadmill.

The MIT Media Lab has developed a portable autonomous powered ankle exoskeleton capable of providing true metabolic benefit during level ground walking. This system consists of a pair of modified boots or shoes, a robotic actuator mounted on the front of each shin, and a battery pack worn at the waist. This portability allows the exoskeleton to be worn in a variety of environments, however to date the efficacy of the exoskeleton while walking uphill or downhill has yet to be determined.

This study demonstrates the MIT exoskeleton is able to deliver consistent added power to the ankle joint across a range of slopes. Data calculated from test subjects’ intake of oxygen and output of carbon dioxide hints at a possible metabolic benefit during uphill walking and a possible added metabolic cost during downhill walking, though these results failed to reach statististical signficance (likely due to an insufficient number of test subjects). Electrical measurements of muscle activity while walking with the exoskeleton provide some explanation for these preliminary findings, however further testing will be required to elucidate the true metabolic effect of this device during inclined and declined walking.

Introduction

Background

Since the first leg exoskeleton was patented in the late 19th century, exoskeleton designers have sought to augment human physical capability[1]. The primary goal of many designs is to reduce the metabolic energy consumed by the human body during periods of strenuous or extended walking [2,3]. Furthermore, assistive devices have the potential to reverse the loss of functional capacity associated with aging and chronic disease, potentially compensating for losses in propulsion, weight support, and balance due to decreased limb torques and forces in the lower extremity [4]. Historically, many attempts at leg exoskeleton design have resulted in additional metabolic cost to the user; however, in the past decade multiple groups have been able to show significant metabolic benefit [5,6,7]. Notably, some of these metabolically successful devices require tethering the exoskeleton to an industrial power supply not worn by the user, which limits their practicality outside of the lab setting.

The Biomechatronics group of the MIT Media Lab has developed an autonomous powered exoskeleton capable of providing a significant metabolic benefit to the user when walking on level ground. [8,9] The original intent of the Mooney et al. device was to reduce the energetic costs of carrying loads without a tethered power source, especially in circumstances where the practicality of wheeled transport is limited and the flexibility and versatility of bipedal locomotion is advantageous. For instance, soldiers often must carry loads of up to 35kg for distances of more than 10km over uneven terrain [10]. To date, this device has demonstrated the greatest published metabolic benefit for a self-contained powered robotic exoskeleton at 11.6% [11], even outperforming some exoskeletons that rely on a tethered power supply [5,6,12,13,14].

However, the Mooney et al. device has thus far only been tested on flat, even terrain, which proves the viability of the system but does not assess its potential to perform when vehicle use is limited – for example, when ascending and descending a steep hill. Though our exoskeleton demonstrates significant benefit on flat ground, it is rendered conditionally impractical if the mechanical assistance provided by the exoskeleton is insufficient to overcome the burden of carrying added mass uphill and therefore against gravity.

Although walking is more metabolically efficient than running or cycling at a 15% incline [15], the metabolic cost of walking increases dramatically with increasing gradient [16]. This is likely due to the increase in the amount of positive mechanical work required to raise the body's center of mass. Although this increased work is derived largely from increased muscle activity across the hip and knee joints [17], an increased contribution from the ankle plantarflexors is also observed [18,19,20]. At a 15% incline, an increase in native ankle moment of up to 18.3% has been reported [18]. This provides an opportunity for meaningful augmentation by an assistive device, particularly because power supplied about the ankle has been shown to lead to proximal joint adaptations leading to the most metabolically economical gait [21].

Though energy consumption on a decline is more complicated due to the assist of descent provided by gravity, additional energy is required for careful control of descent at downhill gradients >8%, possibly leading to a net increase in metabolic energy consumption relative to level ground [16]. Muscle and joint dynamics are also significantly altered on a decline, leading to a substantial decrease in ankle moment and contributions to both positive and negative work [18,22]. Though it has been hypothesized that the human nervous system employs distinct control strategies to handle level ground and sloped terrain [18,20], it remains to be seen whether the control strategy developed for the Mooney et al. exoskeleton on level ground is robust to positive and negative changes in grade.

State of the Field

Three previous evaluations of ankle-exoskeleton-assisted incline walking have been performed: one by Sawicki and Ferris [23], and two by Galle et al. [21,24]. Both groups use pneumatic, tethered ankle exoskeletons. Sawicki and Ferris evaluated subjects at positive 0%, 5%, 10%, and 15% gradients, noting an increase in mechanical power provided by the exoskeleton and an increasing absolute metabolic reduction roughly commensurate with the increasing metabolic demand of incline climbing (however, the percent metabolic reduction remained roughly constant) [23]. In one study, Galle et al. maintained a constant 15% gradient with alterations made to the timing of mechanical actuation, observing similar reductions in metabolic cost [21]. The other Galle et al. study tested subjects on an incline with increasing amounts of added weight, noting that subjects tolerated more weight and more trial time when the exoskeleton was powered than when it was unpowered [24]. In all cases, testing was performed using a tethered exoskeleton, i.e. the mass of the power supply and motor controller were not carried on the subject against gravity as would be the case with our autonomous exoskeleton. None of these studies assessed declined walking. Furthermore, the authors did not compare powered exoskeleton walking to walking without a device; rather, they compared powered walking to unpowered but loaded walking, wherein the user bears the burden of the added exoskeletal mass without receiving the benefit.

Due to the dominance of more proximal muscle groups during inclined walking, other groups have investigated the effect of knee and hip exoskeletons on graded walking energetics [25,26,27]. Seo et al. demonstrated a true metabolic benefit with hip exoskeleton assistance compared to walking without a device, however testing was only performed at 5% and 10% inclines [25]. Kim et al. also reported a metabolic benefit relative to walking without a device at 10% incline, however they only report results for one subject as the focus of their study was level-ground walking [26]. Lee evaluated the performance of a unilateral knee exoskeleton and both 15% incline as well as 15% decline, noting wide variability in metabolic costs among users despite relatively consistent EMG response [27]. Although all three of these studies do evaluate metabolic cost relative to walking without any device worn (rather than walking with an unpowered device as dead weight), they do not adequately fill gaps in what is known about exoskeleton assistance specifically at the ankle joint. Furthermore, two of these studies are limited to 10% incline, and only one addresses decline walking.

Purpose of the Inquiry

This study builds directly on the original design evaluation of the Mooney et al. ankle exoskeleton. The experimental procedures used in testing that device must be mirrored as closely as possible to provide as direct a comparison as possible between the device's original performance on flat ground and its performance on sloped terrain. More important than simply assessing the performance of the mechanical device itself under new conditions, this study aims to evaluate the response of the human-machine system to changing terrain by focusing on exertional metrics.My hypothesis is that a leg exoskeleton capable of providing a significant metabolic benefit on level ground through substantial positive mechanical power and minimal added distal mass will continue to provide a significant metabolic benefit on an incline, and also provide no significant metabolic cost on a decline. Evaluation of this hypothesis involves assessing the performance of the Mooney et al. exoskeleton by measuring respiratory metabolics while participants walk on an instrumented treadmill, as well as collecting lower-limb electromyographic data to gain insight into the biological basis of observed metabolic changes.

Methods

Portable Autonomous Exoskeleton

The bilateral ankle exoskeleton used in this study was previously designed and described by Mooney et al. [8,9,11] and consists of three main components: a pair of boots with attached fiberglass struts, a pair of unidirectional actuators with onboard controllers mounted at the anterior shank, and a single battery pack worn at the waist (Figure 1 and Figure 2). By applying an anterior force to the fiberglass struts which serve as an extension of the ankle-foot complex, the actuators provide an overall torque across the ankle joint, as demonstrated in Figure 3.

The exoskeleton configuration employed in this study differs from the previously described exoskeleton in multiple respects. Whereas the previously described system used bulky high-top boots designed for military use, the current system instead uses lightweight, low-profile Adidas running shoes (Figure 2). The struts are pinned firmly into the rubber sole of the running shoes both medially and laterally at approximately the level of the metatarsophalangeal joints of the foot. A posterior nylon strap secures the struts to the heel of the shoe. As previously described, the effective strut moment arm was set at 230mm [9]. To avoid painful contact of moving struts against the wearer's calves, guards were installed on the medial and lateral sides of each shank-mounted actuator complex.

The actuators used in this system were similar in design to previous iterations of the exoskeleton, consisting of a brushless DC motor system acting as a winch mechanism to pull the proximal ends of the fiberglass struts via 1mm polyethylene cord. The overall transmission ratio for this system was calculated at approximately 160:1. Whereas the previous system mounted both the motor controllers and batteries at the waist, choice of a lightweight, low-profile motor controller system allowed for migration of the motor controller boards distally to the shank actuator complex. Although the metabolic cost of walking is particularly sensitive to increases in distal mass [28], the lightweight design of the controller board coupled with a lighter, more streamlined actuator complex made this migration reasonable (765g compared to 749g prior). This limited the amount of cables required to travel from shank to hip, as all sensor data and motor output could be managed locally at the shank. Once again, Maxon model 305015 motors with attached encoders were used for position-based motor control, however this system also included input from a strain sensor embedded in the actuator complex to enable closed-loop force control rather than the open-loop control previously based on a linear motor model.

As mentioned above, a significant change in this exoskeleton system compared to that previously described is the choice of motor controller. As a part of a lab-wide migration to the unified FlexSEA bionics actuation system [29], this generation of exoskeleton employed a FlexSEA-Execute board to run the motor control loop at 1000 Hz. Using this system of shared coding architecture and modular standardized circuitry means it is easier for lab members to collaborate on projects, as the programming language and circuit design will be familiar across very different projects and applications. With this migration came a significant amount of debugging, however the end result is a more streamlined system which encourages collaboration, is easier to troubleshoot, and offers a fine degree of control over experimental parameters not afforded by the prior system. The FlexSEA-Execute board also contains onboard inertial measurement units (IMUs) as well as a current sensor. As with the previous design, two 24V battery packs were mounted at the waist. The battery packs were connected in series to provide 48V of power. This single power supply was then wired in parallel to be shared between the two actuator systems. The total mass of the system was calculated at 2.75 kg: an 816g battery pack mounted at the waist, 204g struts attached to each athletic shoe, and a 765g actuator worn on each shank.

A similar biomechanically-inspired control strategy was used to maintain consistency with previously published flat-ground work, however the move to a new motor control framework necessitated a complete rewrite of the control code. The controller employs a state-machine-based controller for each phase of the gait cycle. During stance, the actuator behaves like a virtual spring, with neutral position set at an upright standing angle. With progressive ankle dorsiflexion, the system applies an increasing resistive moment to the ankle, either returning the wearer to the neutral angle or else providing a force pre-load in anticipation of powered plantarflexion. If the wearer dorsiflexes to a 0.2 radian deviation from the neutral angle, this is interpreted by the system as intent to initiate powered plantarflexion. Typically, such a transition would occur in late stance. In this case, the actuator delivers a power stroke which is built around a parabolic torque profile with a target peak of approximately 30-40 Nm. The power delivered with each cycle is calculated and tuned by the system to apply approximately 2.3 W/kg peak power. This is approximately 70% of the peak ankle power typically reported for normal walking, and was previously chosen as a desirable limit to prevent thermal overload of the exoskeleton motors [9,30]. Scaling factors are adjusted with each step to achieve this desired peak power. At the conclusion of the parabolic torque profile (or after .275 radians of plantarflexion), the system enters the swing phase, where tension on the winch system is rapidly relieved to allow for the wearer to dorsiflex their ankle unimpeded by the system. This is an important step not only to avoid tripping the wearer, but to eliminate unnecessary fatigue on the wearer's native ankle dorsiflexors. The end of the swing phase is sensed by the system's onboard IMUs, and slack is removed from the winch in order to return to the virtual spring of stance phase. Notably, prior designs employed a foot switch to sense the end of swing phase with heel contact, however the foot switch complex has been removed from the current design.

System Characterization

Although the overall design principles of the Mooney et al. exoskeleton were unchanged, a preliminary test of the system's mechanical performance was warranted given the changes described above. Furthermore, the actual mechanical output of the system had yet to be evaluated under conditions other than level ground. I sought to determine that the system's mechanical performance is robust to changes in slope, so as to ensure investigation of the effect of ramp grade rather than unexpected changes in device dynamics. Applied moment and ankle angular velocity were recorded by the exoskeleton's onboard strain sensor and encoder as a single 68kg male subject walked on the Bertec instrumented treadmill at -15%, 0%, and +15% slope for 35, 38, and 54 steps, respectively. From these recordings, applied joint power was calculated for each step. Applied motor voltage and current were also measured to monitor the electrical requirements of the exoskeleton under these three conditions.

Experimental Protocol

The metabolic effect of the exoskeleton was assessed in a fashion similar to previous experiments using the Mooney et al. exoskeleton. Four healthy male subjects weighing 73±7 kg and exhibiting no gait abnormalities were recruited. Testing was performed on a Bertec instrumented treadmill at 1.25 m/s, slightly lower than the average adult walking speed on level ground [31], to avoid exhaustion or imbalance on graded trials. Participants were evaluated during one lab session each. Before walking, 5 minutes of standing metabolic data were collected to assess the participant's baseline metabolic rate, followed by three walking trials of 10 minutes each. This time was chosen to balance maximization of human-machine adaptation and avoidance of anaerobic respiration during strenuous uphill trials [32]. These three trials consisted of i) walking without any device components worn, ii) walking with the device donned but unpowered, and iii) walking with the device donned and powered. The order of these three trials was randomized for each subject. These three trials were repeated once for inclined walking at +15% grade and once for declined walking at -15% grade, the order of which was also randomized. After the first set of three trials, another 5 minute standing trial was performed to obtain an additional measurement of resting metabolic rate before switching grades. After three more 10-minute walking trials, a final 5-minute standing trial was performed. All of the participants were asked to give their informed consent after the nature and possible consequences of exoskeleton-based experiments were explained.

Data Collection and Analysis

Participants wore a portable pulmonary gas exchange measurement device (COSMED K4b2) during the trials to collect data on the rate of oxygen inspiration and carbon dioxide expiration. The average flow rates of the last four minutes of each trial were converted into metabolic power using the equation developed by Brockway et al. [33]. The standing, baseline metabolic rate for each subject was subtracted from the metabolic rates of each walking trial to determine the metabolic cost of activity. Metabolic costs was then normalized by body mass and averaged for each of six trial conditions: incline and decline with no device, incline and decline with exoskeleton assistance, and incline and decline with the exoskeleton unpowered.

Participants were also outfitted with wireless EMG electrodes at the tibialis anterior, medial gastrocnemius, lateral gastrocnemius, rectus femoris, biceps femoris, and gluteus maximus of one leg according to SENIAM guidelines for EMG placement [34]. The instrumented treadmill is outfitted with a force plate which measured the ground reaction forces from walking. Both EMG and force data were sampled at 2000 Hz. EMG data from only three of the four participants was included for analysis due to inadequate electrode adhesion. EMG data was captured for each subject as 30-second windows distributed throughout each walking trial. The RMS envelope was calculated with a 150-sample smoothing window and subsequently averaged across all steps within a given trial condition. For each subject, EMG RMS amplitude was normalized to the peak value of the device-free walking trial and time-normalized from heel contact to heel contact.

Personal Role

I was personally responsible for effecting all aspects of the current study, including device control code rewrites, subject recruitment, data collection, and final analysis.

Results

System Characterization

Across all three slope settings (-15%, 0%, and +15%), the exoskeleton was able to achieve its peak 2.3 W/kg target during most strides. While on average the exoskeleton exceeded this performance goal on a decline (2.46 W/kg) and on flat ground (2.48 W/kg), the exoskeleton fell short at 2.12 W/kg average peak on an incline. This appears to be due to a failure of the exoskeleton to deliver an adequately rapid plantarflexion stroke, as evidenced by the blunted peak ankle angular velocity on an incline (4.38 rad/s compared to 5.66 rad/s and 5.31 rad/s on decline and flat ground, respectively). However, the exoskeleton does exhibit an increased peak ankle moment on an incline (38.56 Nm) compared to that seen on a decline (33.33 Nm) or flat ground (31.57 Nm). Average power across the full stride is calculated as 0.17 W/kg on a decline, 0.16 W/kg on flat ground, and 0.20 W/kg on an incline. Results are summarized graphically in Figure 4.

Electrical requirements of the exoskeleton are fairly consistent across testing conditions. The peak electrical power required is greatest on an incline at 230 W, followed by 211 W on flat ground and 208 W on an incline. Peak current draw is highest on an incline at 11.2 A, compared to 10.5 A peak on a decline and 9.7 A peak on flat ground. Results are summarized graphically in Figure 5.

Metabolic Cost

Evaluation of exoskeleton-assisted walking on both a +15% incline and -15% decline demonstrated no statistically significant metabolic cost or benefit. A 5.86% decrease in metabolic cost was observed while walking with a powered exoskeleton on an incline relative to walking without an exoskeleton (9.11±1.60 W/kg vs. 9.68±0.609 W/kg), however this result was not statistically significant ($p=0.21$). Conversely, a 24.97% increase in metabolic cost was observed while walking with a powered exoskeleton on a decline relative to walking without an exoskeleton (3.29±0.81 W/kg vs. 2.63±0.33 W/kg). This result also failed to achieve statistical significance ($p=0.12$). Across all three walking conditions, inclined walking required substantially greater metabolic activity than declined walking (9.36±1.14 W/kg vs. 2.97±0.65 W/kg, $p=1.31E-06$). Incline data with exoskeleton assistance was lost for one subject due to recording device power failure. Results are summarized graphically in Figure 6.

Electromyography

Normalized data for six muscles, averaged across three subjects, are visualized separately for walking on an incline (Figure 7) and on a decline (Figure 8). On an incline, a reduction in muscle activity was noted in the lateral gastrocnemius, rectus femoris, and gluteus maximus while walking with exoskeleton assistance compared to walking without a device. Medial gastrocnemius and biceps femoris activation were similar. An increase in tibialis anterior activity is noted during the loading response, pre-swing, and terminal swing.

On a decline, a reduction in muscle activity can be seen in the rectus femoris during the loading response with exoskeleton assistance. Activity in the medial gastrocnemius and gluteus maximus are similar between conditions. Tibialis anterior activity is again increased during the loading response, pre-swing, and terminal swing, along with activity in the lateral gastrocnemius during these periods. Increased activity in the biceps femoris can also be observed during pre-swing.

Discussion

System Characterization

On level ground, the mechanical performance data demonstrates near functional equivalence between the newest iteration of the MIT ankle exoskeleton compared to previously published results [9]. The peak power target of 2.3 W/kg was approximated via dynamic adjustments of scaling factors within the control algorithm, resulting in an average peak power of 2.48 W/kg on flat ground, slightly higher than desired. The average power provided per stride (0.16 W/kg) was also consistent with previous results on flat ground [9].

While the exoskeleton was able to provide similar results during declined walking, the device failed to achieve its average peak power target of 2.3 W/kg on an incline, instead offering only a peak 2.12 W/kg on average. This was largely driven by a relatively sluggish angular velocity, for which the system attempted to compensate with increased ankle torque. Although this increased applied torque is desirable in the context of the increased native ankle moment required during inclined walking without an exoskeleton [18], recent analyses of ankle exoskeleton control strategies have revealed that added positive work is more effective than added torque at reducing metabolic cost [35,36]. Interestingly, although peak power measures were decreased during inclined walking, average power over the entire stride was measured at 0.20 W/kg, an increase relative to average power delivered during decline or flat-ground walking, resulting in a larger net work per stride. This average power per stride on an incline is higher than the 0.12 - 0.13 W/kg reported by Galle et al. with a tethered pneumatic system [21], and lower than the 0.37 W/kg reported at a 15% by Sawicki and Ferris, also with a tethered pneumatic system [23].

As a portable autonomous system unable to benefit from a central lab power supply, the MIT exoskeleton appears to be constrained to approximately 200 - 250 W peak electrical power across all slopes to ensure the system is not damaged by overdraw. However, the highest average electrical power consumption per step (measured at 27.3 W during decline walking) is lower than the average 45 W reported for flat-ground walking by Mooney et al. [9], suggesting increased efficiency of this device and perhaps a potential to liberalize the mechanical power restrictions in future testing.

One limitation of this study is the fact that exoskeleton performance data was obtained asynchronously with subject testing. Data obtained during subject walking trials would provide extra utility in that this data would allow for direct correlation between device performance and the human subject response.

Metabolic Cost

Within a given ramp condition (i.e. incline vs. decline), no significant differences in metabolic rates were observed between device-free walking, powered exoskeleton assistance, and the unpowered exoskeleton as “dead weight”.

A nonsignificant benefit was observed when using a powered exoskeleton on an incline, although the validity of this result is undermined by the fact that walking with an unpowered “dead weight” exoskeleton was also metabolically less costly than unencumbered walking (9.30±1.42 W/kg vs. 9.67±0.61 W/kg). One plausible explanation is that the drastic increase in the metabolic cost of walking on an incline dwarfs the small effect this portable autonomous ankle exoskeleton is able to impart. While Mooney et al. reported a 0.43 W/kg decrease in metabolic cost attributable to ankle plantarflexion assistance, this level of assistance is much more impressive in the context of that study’s 3.71 W/kg walking trial, as compared to the 9.67 W/kg metabolic cost observed here on an incline [11]. While Sawicki and Ferris reported a 10% reduction in metabolic cost between their powered exoskeleton and unpowered “dead weight” trials, their pneumatic, tethered exoskeleton was also capable of delivering higher ankle power (as noted above), albeit limited by an immobile power supply and cumbersome pneumatic system [23]. Furthermore, the authors did not report the metabolic cost of walking without their device, therefore true metabolic benefit cannot be calculated.

Whereas inclined walking with the powered exoskeleton demonstrated a nonsignificant trend toward metabolic benefit, an unexpected nonsignificant trend toward added metabolic cost was observed during declined walking. Interestingly, this preliminary data suggests it is less costly to wear an unpowered 'dead weight' exoskeleton on a decline than to receive powered plantarflexion assistance. I hypothesize this may be due to a destabilizing effect and lack of balanced control during careful ramp descent. Whereas subjects may aim to take great care with each step to land in a balanced, coordinated fashion, the unfamiliar forward propulsion of a powered exoskeleton may give subjects the uncomfortable sensation of being launched downhill. Though this effect lacks statistical significance, a potential biomechanical basis for this phenomenon is explored below.

The major limitation of this study is sample size. While prior studies with the MIT ankle exoskeleton system evaluated the system on at least six test subjects [8,9,11], only four subjects were tested in the current study. Furthermore, data from the powered incline testing condition for one subject was lost, further contributing to small sample size.

The most recent testing of the Mooney et al. exoskeleton demonstrated an effect size of 0.23 W/kg with a standard deviation of 0.14 W/kg when comparing powered walking to walking without an exoskeleton [11]. Prior to the current experiment it was determined based on this previous data that at 80% statistical power and a level of significance $p = 0.05$, three subjects should afford the ability to detect such an effect size. However, in practice these new testing conditions yielded substantially more variable results among subjects, leading to 25.76% calculated statistical power on the incline condition and 22.85% statistical power on the decline.

In the future studies, decreasing the walking slope to +10% and -10% may alleviate subject displeasure with walking 60 minutes along challenging gradients and lead to easier subject recruitment. Furthermore, the decreased metabolic cost of walking on a 10% incline relative to a 15% incline should allow for enhanced visualization of effect size at the current electromechanical safety limits which have been placed on the MIT exoskeleton.

Electromyography

Although care must be taken when interpreting the absolute magnitude of electromyographic signals due to muscle crosstalk and a non-linear, fiber-length dependent relationship of voltage to force production [37], an analysis of the relative magnitudes of signals during the phases of gait may explain the trends identified in metabolic data.

During inclined walking, active exoskeleton assistance did appear to reduce the activity of the lateral gastrocnemius. This is consistent with findings by Steele et al. and Sawicki and Ferris suggesting that plantarflexion assistance can reduce the EMG activity of the ankle plantar flexors of the triceps surae [23,36]. Curiously, medial gastrocnemius activity was not significantly reduced, suggesting either improper electrode placement, significant crosstalk, or perhaps an asymmetrical effect of assistance on a given leg. The reduction in activity noted in the rectus femoris and gluteus maximus is consistent with the findings of Galle et al., suggesting that power applied to the ankle joint on an incline impacts the more proximal muscle groups which dominate during uphill walking [21]. While these authors describe a reduction in vastus lateralis and biceps femoris activity, here we present reductions in activity of two other proximal muscles involved in incline gait. Should the ankle exoskeleton system continue to demonstrate a metabolic benefit over a larger sample size, it may be the case that this is best explained by reductions in proximal muscle activity rather than simply ankle plantar flexors.

Discouragingly, the marked increase in tibialis anterior activity across the gait cycle is suggestive of unfavorable co-contraction. This may be due to a combination of factors, including lack of user comfort with exoskeleton assist causing the wearer to resist joint torques with their native antagonist dorsiflexors (all but one subjects were exoskeleton-naïve), or perhaps inappropriate actuation timing leading users to force dorsiflexion against resistance in early swing. If the latter is the case, there may be a role for predictive sensing of ramp incline within the exoskeleton control algorithm, as is under investigation for the purpose of enhanced powered prosthetic limb control [38].

During declined walking, a similar increase in tibialis anterior activity was observed, however increases in lateral gastrocnemius and biceps femoris activity were also noted. This may be to an overall effect of destabilizing gait perturbation caused by inappropriate powered plantarflexion during carefully controlled ramp descent. The only muscle to exhibit substantial reduction in activity during declined walking is the rectus femoris during the loading response of gait. This may be due to the anterior propulsion of the shank caused by powered plantarflexion against a ramp, resulting in a knee extension moment that otherwise would have been performed in part by the rectus femoris [20].

Conclusions

The Biomechatronics group of the MIT Media Lab has developed an autonomous powered exoskeleton capable of providing a significant metabolic benefit to the user when walking on level ground. To date, this device has demonstrated the greatest published metabolic benefit for a self-contained robotic exoskeleton, even outperforming some exoskeletons that rely on a tethered power supply. While other groups have evaluated aspects of exoskeleton performance on inclined terrain, no group has provided a direct quantitative comparison of normal, unassisted sloped walking to sloped walking wearing a powered autonomous exoskeleton.

We hypothesized that a leg exoskeleton capable of providing a significant metabolic benefit on level ground through substantial positive mechanical power and minimal added distal mass would continue to provide a significant metabolic benefit on an incline, and that such a system would also provide no significant metabolic cost on a decline. In four healthy male subjects, the exoskeleton was found to impart no significant net metabolic benefit nor net cost to the user during inclined walking or declined walking. The lack of significance is likely the result of the low sample size, however early trends demonstrate a diminished but intact metabolic benefit during inclined walking, and an unexpected added metabolic cost during declined walking. This is likely attributable to the much higher overall metabolic cost of inclined walking relative to the amount of mechanical power applied by the exoskeleton, and possibly a net destabilizing effect of powered plantarflexion during careful ramp descent. Although the metabolic results are nonsignificant, EMG data may explain this preliminary data, with reductions in proximal leg muscle activity during ramp ascent and increases in a variety of muscle groups during ramp descent.

Importantly, the results of this study suggest that the metabolic cost associated with the added mass of the ankle exoskeleton may at the very least be evenly balanced by the mechanical power imparted to the user by the ankle exoskeleton, resulting in no net burden to the user while walking uphill at a 15% incline. In demonstrating lack of cost on an incline and illustrating a trend of added metabolic cost during powered decline walking, this study further characterizes both the the practical utility as well the limitations of an autonomous powered exoskeleton for military, recreational, rehabilitative, or other use.

Supplement

The following represents unpublished work completed in collaboration with Roman Stolyarov and Jen Dawkins. As the body of my thesis highlights a role for an altered control strategy during sloped walking, a method of recognizing changes in terrain will become necessary as the exoskeleton control strategy evolves.

Terrain prediction in lower limb wearable devices using anticipation of foot strike timing

Introduction

Previous work has explored terrain prediction in the BiOM prosthesis using rigid body equations and linear discriminant analysis (LDA) and has achieved high accuracy [38]. Terrain prediction is necessary because ankle angles and torque are altered on different terrain causing the system to provide a force at the wrong point in the walking cycle. However, we found that some error arises from using an imperfect 'prediction window' (cutting off data gathering too soon to give enough time for device actuation). Thus, we looked into developing a system that could accurately estimate the time remaining in swing. In applying the BiOM terrain prediction algorithms to the Mooney et al. exoskeleton, the nature of the exoskeleton (namely, data from two feet instead of one) gave an opportunity to explore additional measures of predicting terrain with higher accuracy. In addition to exploring stride time remaining estimation and using data from two feet instead of one, we looked at other measures of building upon past work by exploring deep learning methods and comparing their accuracy.

Objectives

To further our prior work and analyze the possibilities for new improvements, we performed 3 experiments to:

(1) assess the accuracy of alternate learning methods,
(2) assess any improvements in using data from two walking feet, and
(3) assess any improvements by making more 'timely' predictions.

Objective 1: Compare Learning Methods

Methods

While wearing the plantarflexion exoskeleton, 1 subject completed ~ 600 strides of flat ground walking, upstairs walking, and downstairs walking. The prediction signals fed to machine learning algorithms were x/y acceleration, x rotational velocity and the integrals and derivatives of these signals. 3 models were compared that took in a full stride of data and predicted the terrain from the 3 classes:

(1) Convolutional Neural Network (CNN) with 1 convolution layer and 1 fully connected layer
(2) Long Short-Term Memory (LSTM) with 1 hidden layer
(3) Linear Discriminant Analysis (LDA)

Because LDA is not a 3D method like CNN or LSTM, the time dimension was discarded. Input data included the mean, max, min, and range over each stride for each of the 9 prediction signals.

Results

Despite the fact that LDA required summing over the time dimension, LDA largely out-performed the deep learning methods (Figure 9).

Objective 2: Comparing Prediction Accuracy in 1 vs 2 Feet

Methods

While wearing the plantarflexion exoskeleton, 1 subject completed ~ 1200 strides of flat ground, upstairs, downstairs, up-ramp, and down-ramp walking. Similar models to those described above were constructed, as well as an LSTM model that makes predictions over 50, 90, or 120 milliseconds for the terrain of the next stride (rather than over the ~1000ms of the whole stride).

Results

There was little difference in using data from one foot versus two feet of data using an LDA or an LSTM trained over the whole stride (Figures 10 and 11). However, a change was seen when using a smaller section of data to train an LSTM (Figure 12).

Objective 3: Improve Terrain Prediction Through Anticipation of Foot-Strike Time

Methods

While wearing a BiOM prosthesis, six subjects completed 8200 strides over five terrains including level ground and ascending/descending stairs and ramps while telemetry was logged. To allow for enough actuation time for the system to respond to changing terrain, the terrain predictions must be made with a several milliseconds to spare. If predictions are made too late, the system will not have time to respond. If predictions are made too early, data is lost.

Linear discriminant analysis (LDA) models were trained and cross-validated offline under three conditions:

1. Predictions were made a set time after foot-off (windowing by elapsed time), leading to lost data but ensuring enough actuation time.

2. Predictions were made at the "perfect" time before foot-strike where the most data can be seen by the LDA with enough time for actuation (windowing by remaining time).
3. An ensemble learning algorithm with 100 trees and least-squares boosting was used to estimated remaining time and applied it in combination with the LDA classifier to determine if any an improvement exists in prediction accuracy (windowing by predicted remaining time)

Results

Estimations of remaining time in stride were fairly accurate in the training set, as expected (Figure 13), but tended to have a much larger variance in the test set (Figure 14). Figure 15 shows that error rate does indeed increase when data is lost to ensure enough actuation time (windowing by elapsed time), as compared to the ideal case. However, due to the large variance in our time remaining predictions, using such predictions did not lead to any apparent decrease in error. Interestingly, error does not increase in the ideal case when max allotted actuation time increases, meaning the data in the last 50 to 300ms of the stride is not critically important to predicting terrain. However, error does increase in the other two cases as the chance of predicting too late increases.

Conclusions

While most methods explored did not lead to increased accuracy, several interesting results were obtained that warrant further investigation, such as the interesting difference between continuous and per-stride LSTM predictions and ways to decrease variance of predicted time remaining. Possible further routes for the latter are using filtering (such as Kalman filtering) to restrict future predictions based on past predictions to more likely values or modeling the entire system as a Hidden Markov model and using state estimation techniques.

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Figures and Tables

Figure 1: The original Mooney et al. portable autonomous powered ankle exoskeleton, 2014

Figure 2: A later iteration of the MIT Media Lab portable autonomous powered ankle exoskeleton featuring low-profile metal struts not utilized in this study. However, the presently described exoskeleton system does feature athletic shoes rather than military boots.

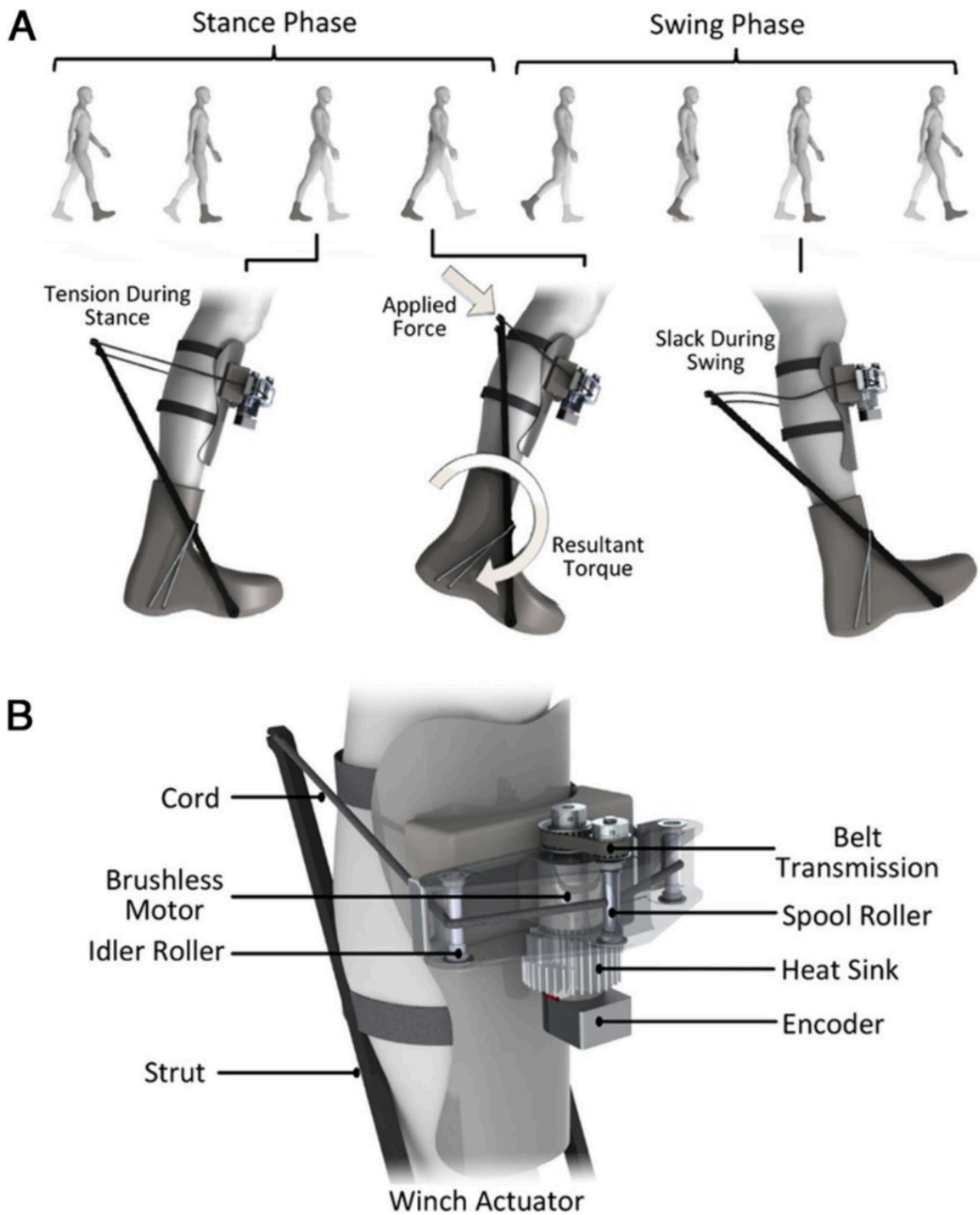

Figure 3: (From Mooney et al. 2014) A plantarflexion torque is applied about the ankle in late stance just prior to the toe-off phase of gait. Force is applied to struts which act as an extension of the foot-ankle complex rather than directly to the foot.

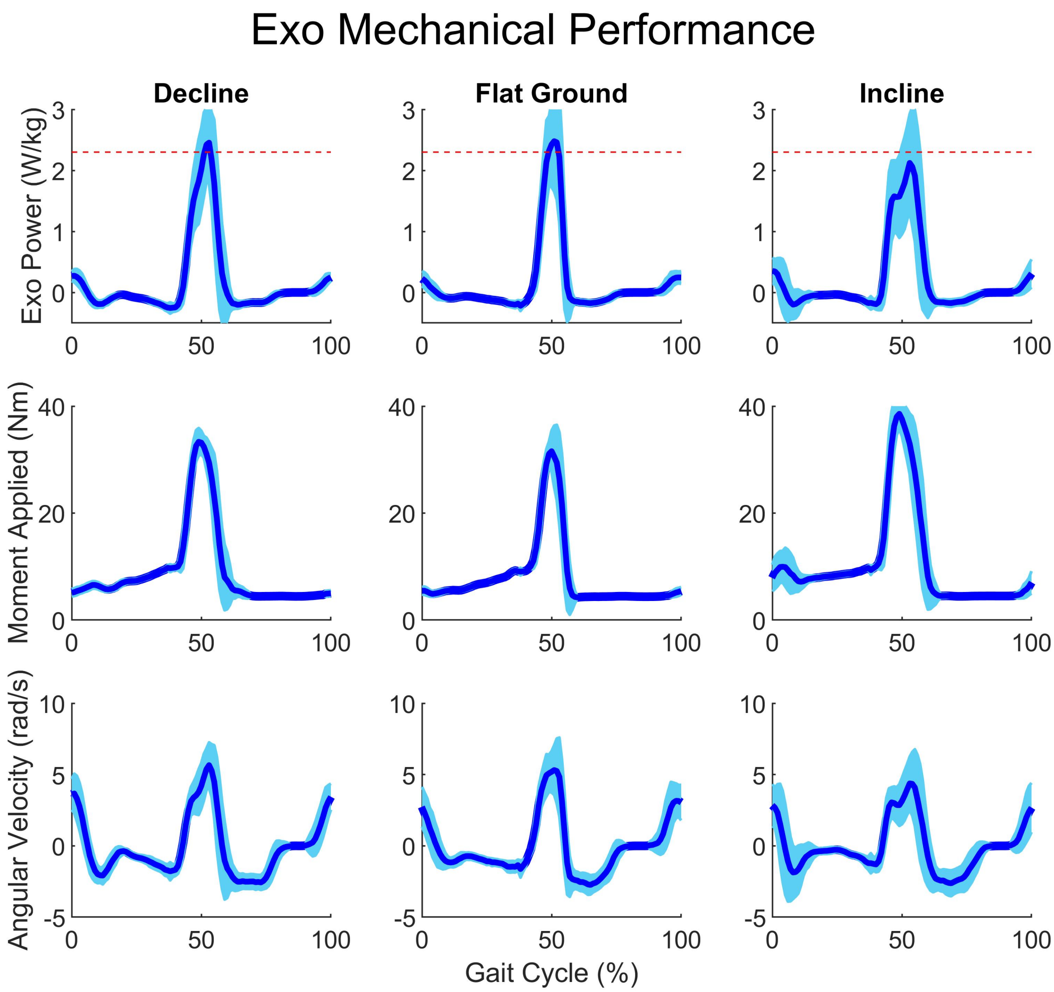

Figure 4: Mechanical performance of the exoskeleton on a -15% decline, flat ground, and a +15% incline. Average curves are displayed in bold blue with one standard deviation indicated by light blue shading. Appropriate power is delivered on a decline and flat ground, though the system struggles to reach its peak power target on an incline. The exoskeleton attempts to deliver a higher ankle moment on the incline, however it struggles to keep up a consistent rate of plantarflexion as indicated by the blunted ankle angular velocity curve.

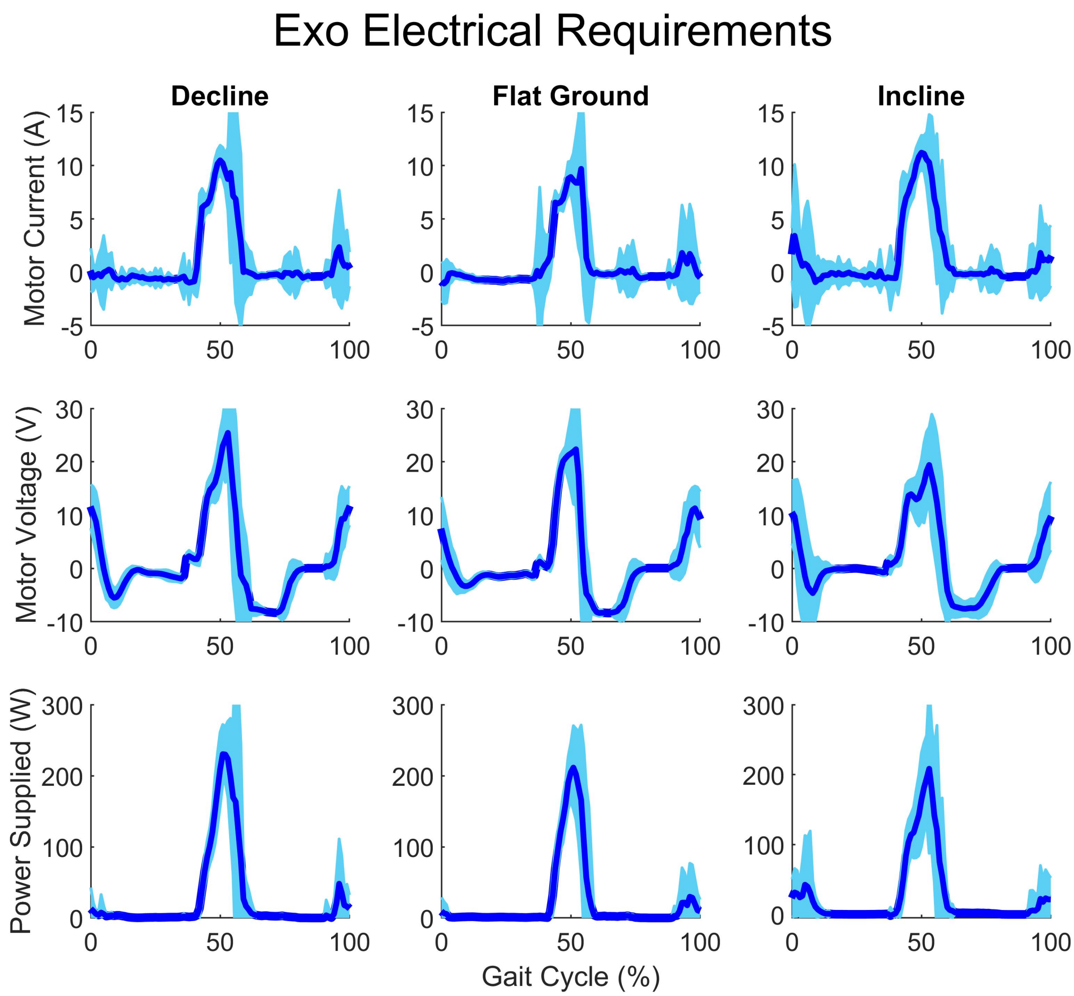

Figure 5: Electrical requirements of the exoskeleton on a -15% decline, flat ground, and a +15% incline. Average curves are displayed in bold blue with one standard deviation indicated by light blue shading. Peak power is roughly equivalent across all conditions, though inclined walking draws a higher peak current (11.2A) compared to declined (10.5A) or flat-ground (9.7A) walking due to higher torques required to maintain adequate ankle power.

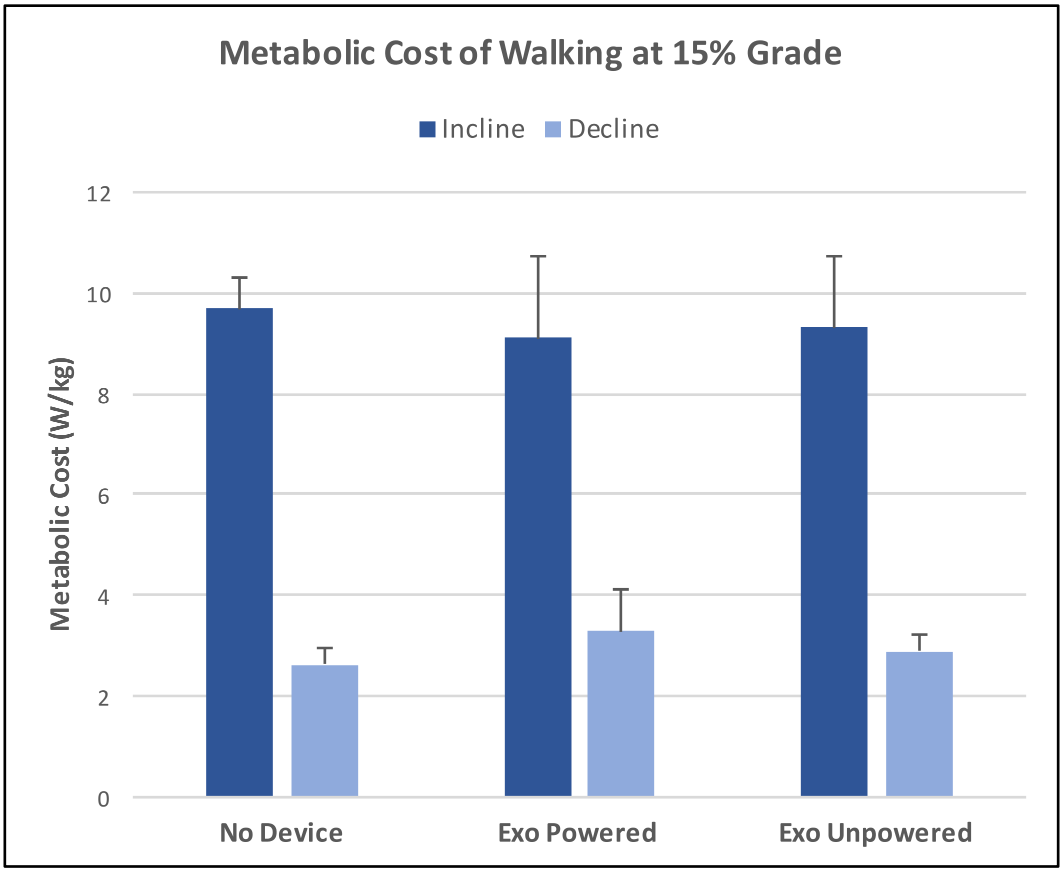

Figure 6: Within a given ramp setting (i.e. incline or decline), no significant difference in metabolic cost was observed. Obvious and expected differences in metabolic cost exist between incline and decline walking.

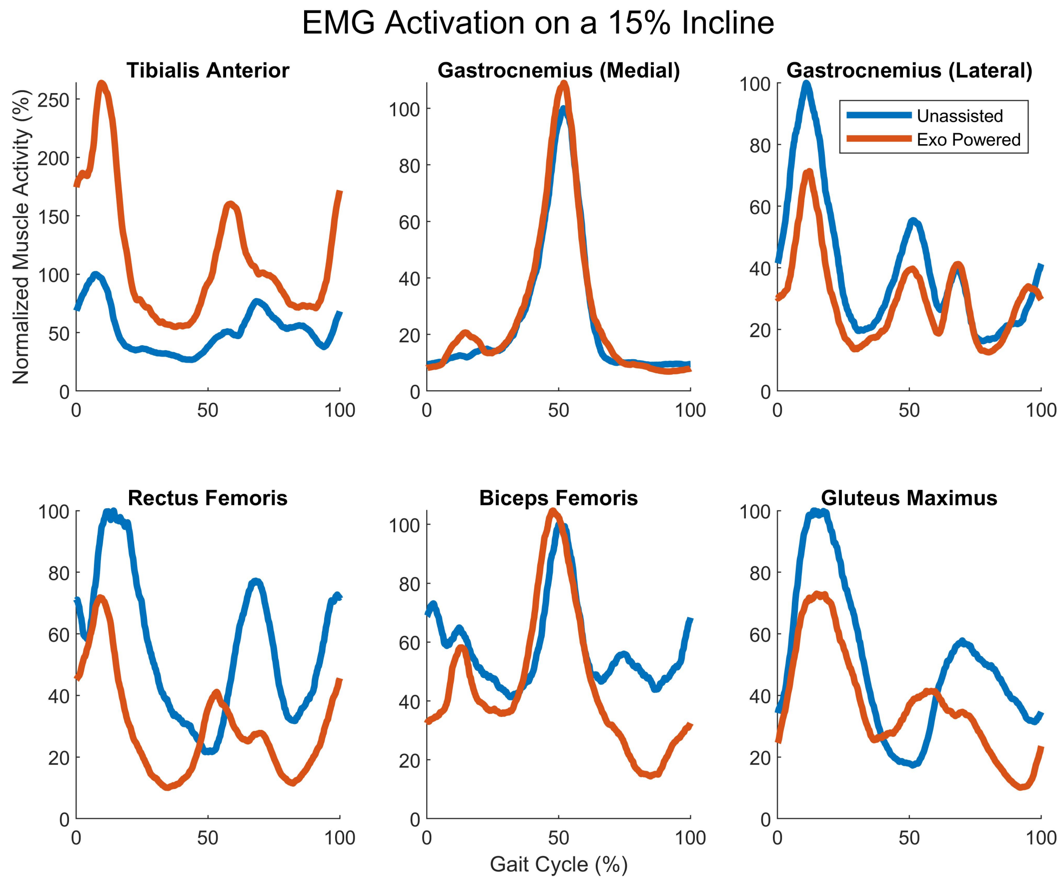

Figure 7: Electromyography of three subjects walking on a 15% incline. With an active exoskeleton, decreased activity is seen in the lateral gastrocnemius, rectus femoris, and gluteus maximus relative to walking with no device. Increased muscle activity is observed in the tibialis anterior.

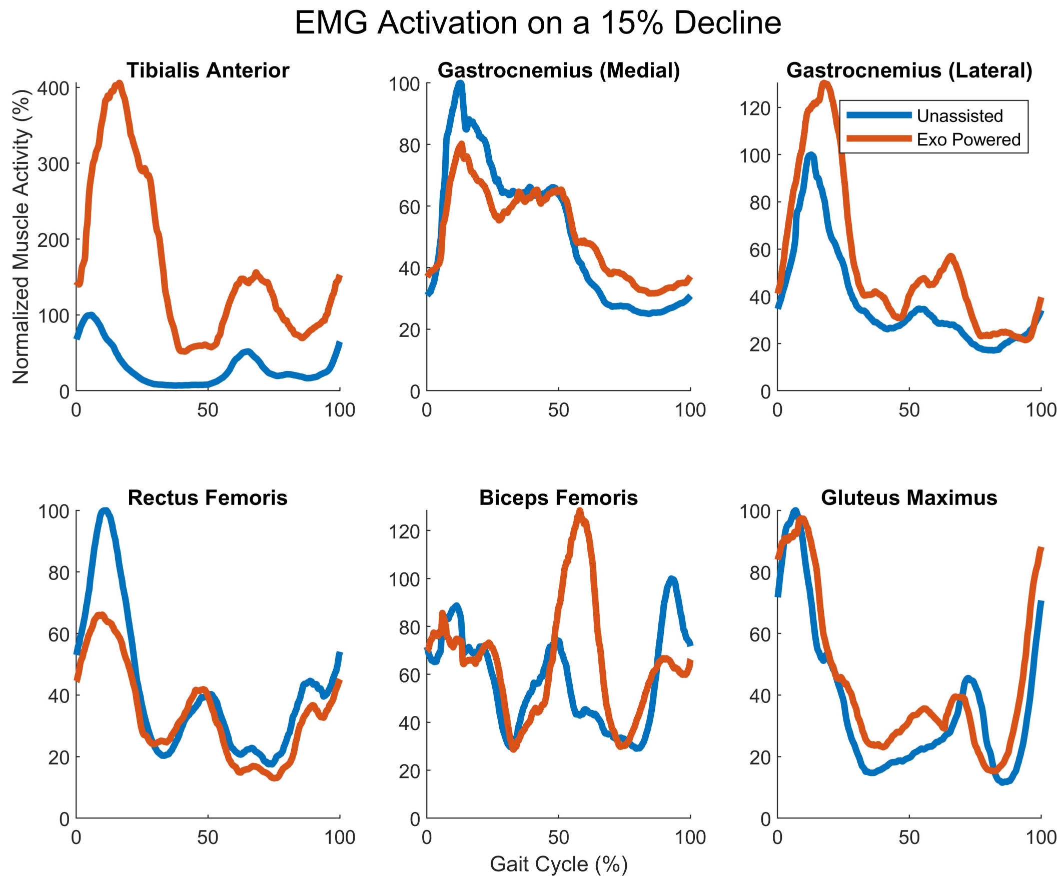

Figure 8: Electromyography of three subjects walking on a 15% decline. With an active exoskeleton, decreased activity is seen in the rectus femoris. Increased muscle activity is observed in the tibialis anterior, lateral gastrocnemius, and biceps femoris.

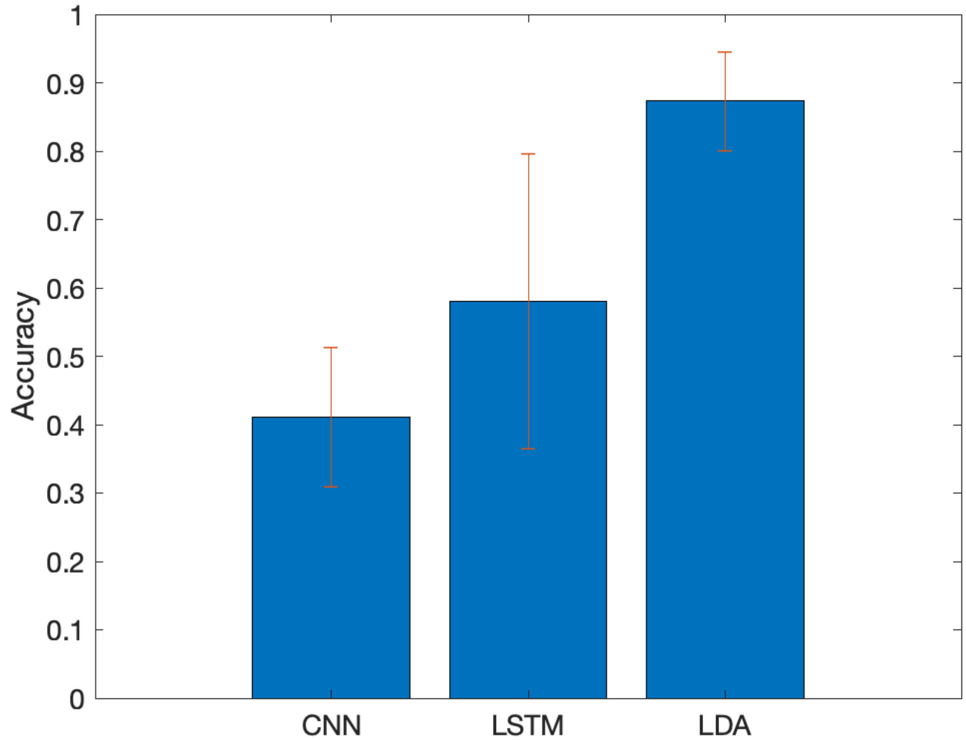

Figure 9: Comparison of accuracy of deep learning methods (CNN, LSTM) compared to a classical machine learning method (LDA).

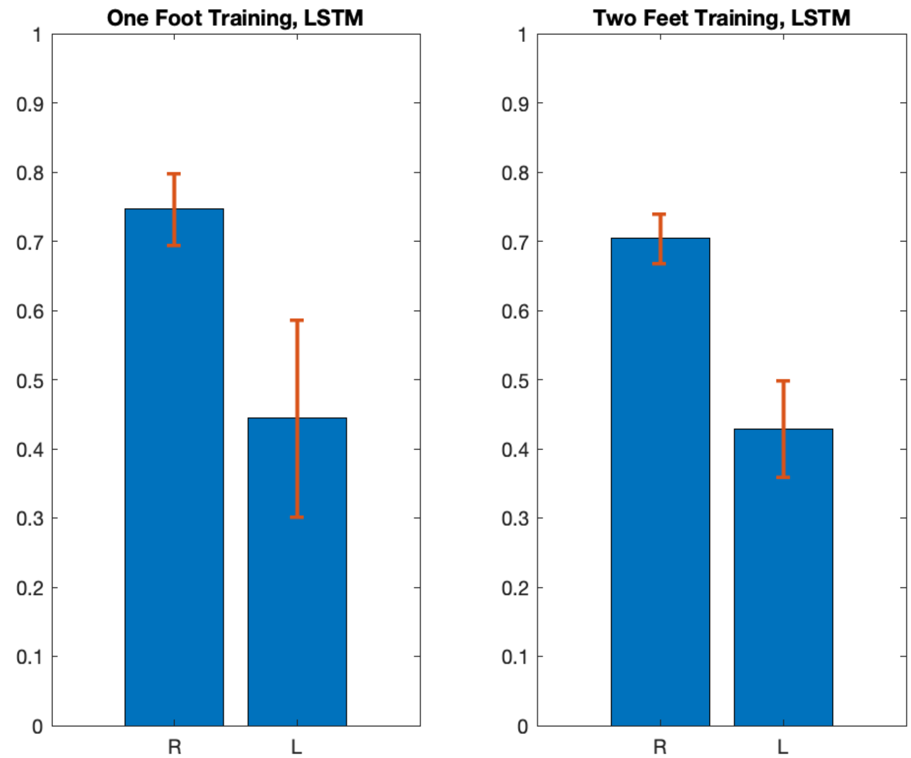

Figure 10: Comparison of terrain prediction accuracy when using data from one foot versus using data from two feet. Here an LSTM is trained by data one whole stride at a time (~1000ms).

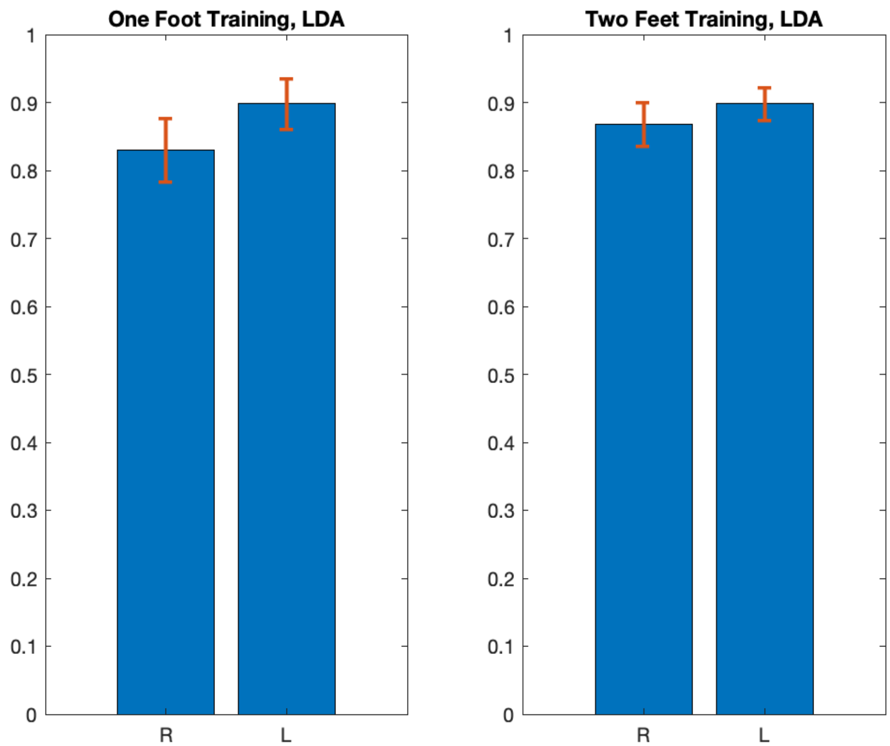

Figure 11: Comparison of terrain prediction accuracy when using data from one foot versus using data from two feet. Here, an LDA is trained by data one whole stride at a time (~1000ms).

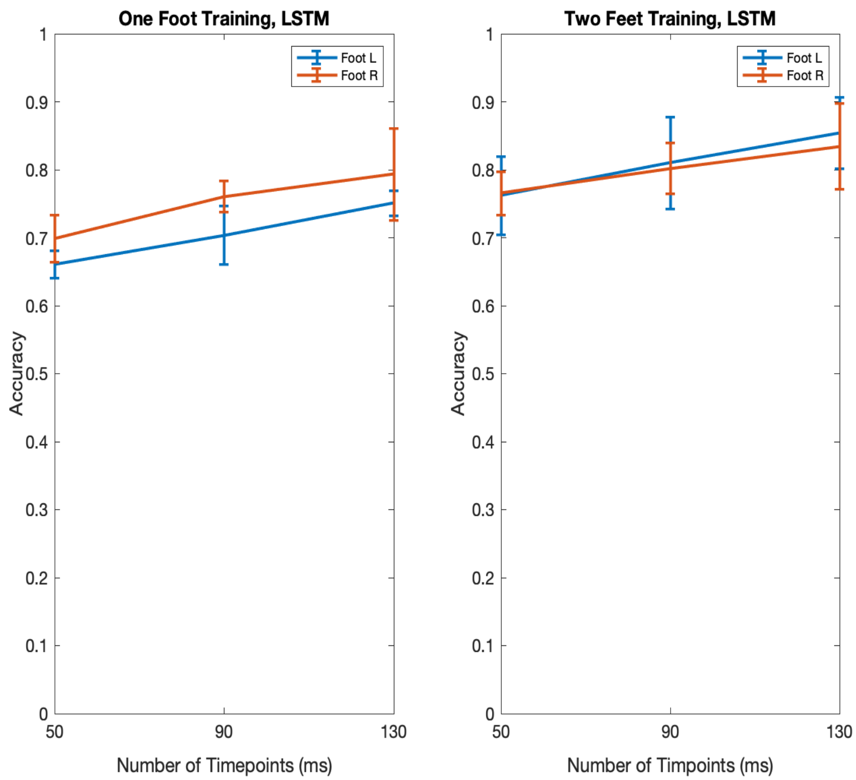

Figure 12: Comparison of terrain prediction accuracy when using data from one foot versus using data from two feet. Here, data is trained over 50 ms, 90 ms, or 120 ms.

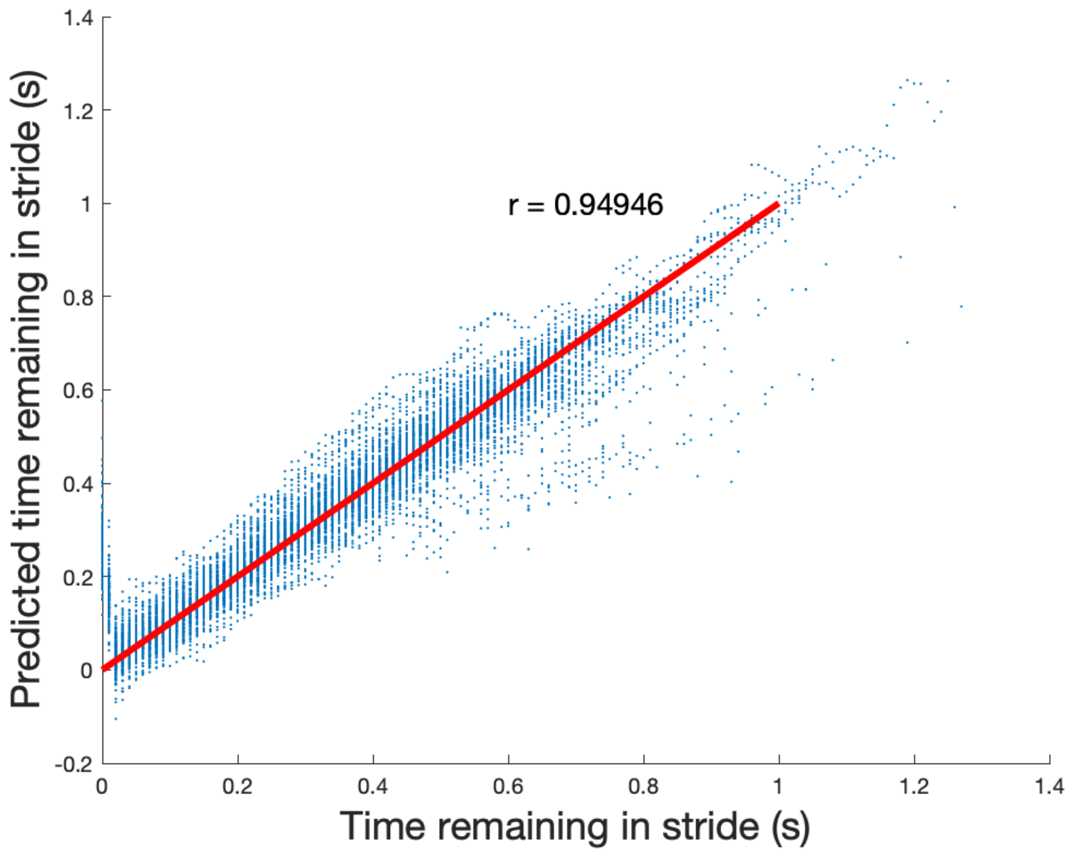

Figure 13: Estimation of remaining time in stride from ensemble learner for training data.

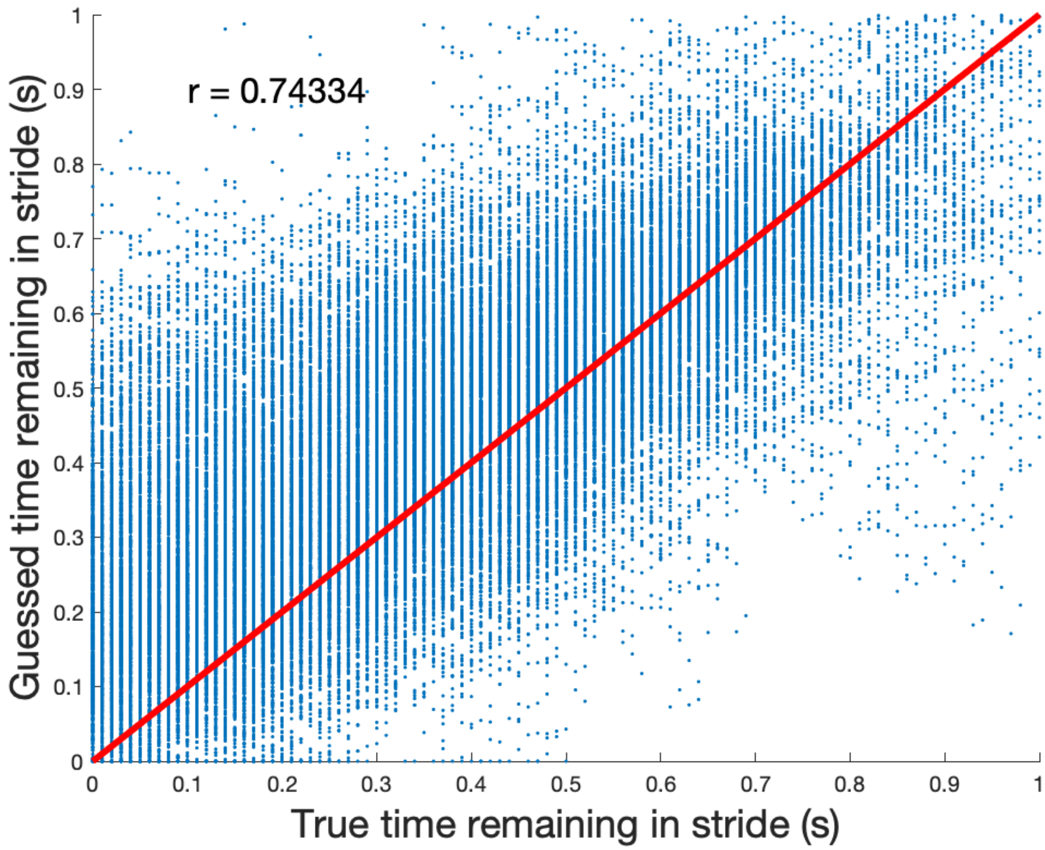

Figure 14: Estimation of remaining time in stride from ensemble learner for test data.

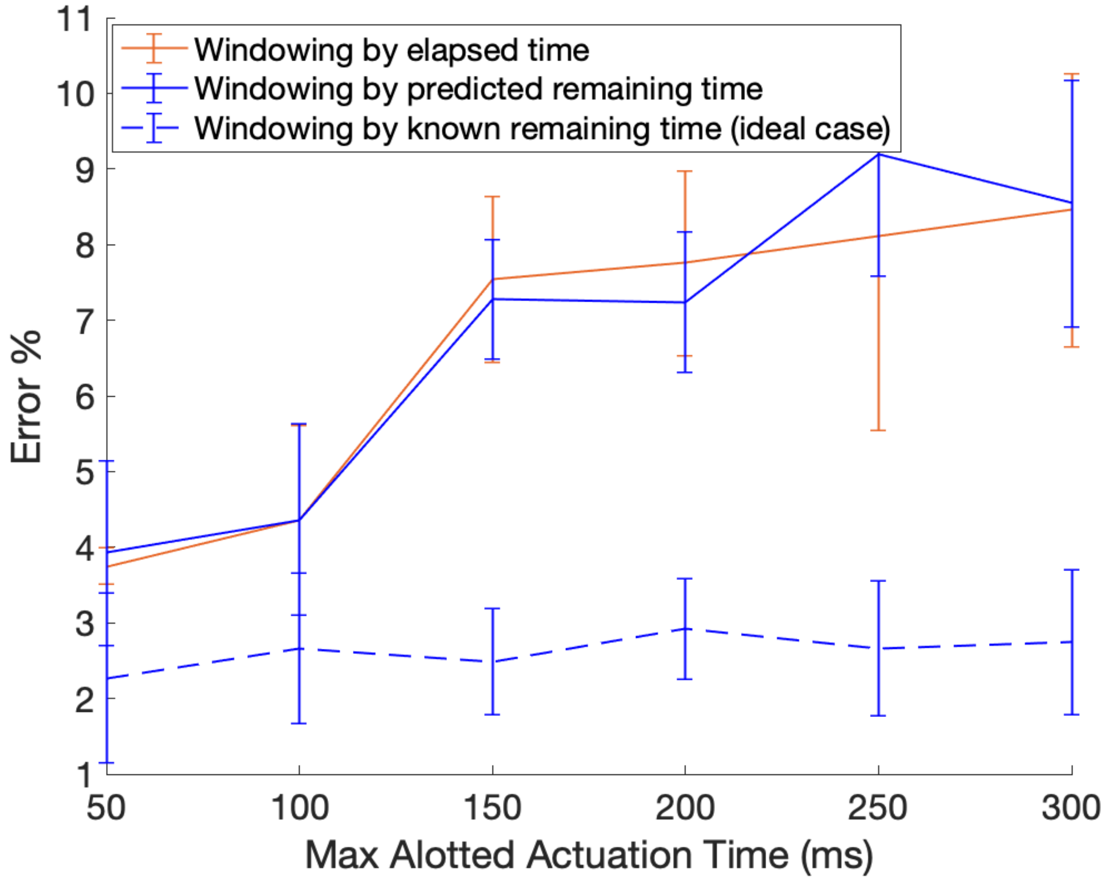

Figure 15: Error rate of terrain predictions when predicting at the right time (ideal case), predicting at a safe time for actuation (windowing by elapsed time), and predicting when to predict terrain (windowing by predicted remaining time)

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