Improving the Energy Cost of Incline Walking and Stair Ascent with Ankle Exoskeleton Assistance in Cerebral Palsy

Ying Fang, Greg Orekhov, Zachary F. Lerner

Abstract—**Objective: Many individuals with cerebral palsy (CP) experience gait deficits resulting in metabolically-inefficient ambulation that is exacerbated by graded walking terrains. The primary goal of this study was to clinically-validate the accuracy and efficacy of adaptive ankle exoskeleton assistance during steady-state incline walking and stair ascent in individuals with CP. Exploratory goals were to assess safety and feasibility of using adaptive ankle exoskeleton assistance in real-world mixed-terrain settings. Methods: We used a novel battery-powered ankle exoskeleton to provide adaptive ankle plantar-flexor assistance during stance phase. Seven ambulatory individuals with CP completed the study. Results: Adaptive controller accuracy was 85% for incline walking and 81% for stair-stepping relative to the biological ankle moment. Assistance improved energy cost of steady-state incline walking by 14% (p = 0.004) and stair ascent by 21% (p = 0.001) compared to walking without the device. Assistance reduced the muscular demand for the soleus and vastus lateralis during both activities. All participants were able to safely complete the real-world mixed-terrain route, with adaptive ankle assistance resulting in improved outcomes compared to walking with the device providing zero-torque; no group-level differences were found compared to walking without the device, yet individuals with more impairment exhibited a marked improvement. Conclusion: Adaptive ankle exoskeleton assistance can improve the energy cost of steady-state incline walking and stair ascent in individuals with CP. Significance: As the first study to demonstrate safety and performance benefits of ankle assistance on graded terrains in CP, these findings encourage further investigation in free-living settings.**

Index Terms—**Incline walking, Stair ascent, Metabolic of locomotion, Cerebral Palsy, Exoskeleton.**

I. Introduction

GRADED walking terrains (i.e., stairs and slopes) pose significant mobility challenges for individuals with movement disorders, including those with cerebral palsy (CP), the most common childhood movement disorder [1], [2]. Individuals with CP have gait deficits that result in slow, energy-expensive ambulation on level ground [3]–[5]. These walking performance deficits in ambulatory individuals with CP are often exacerbated during locomotor tasks requiring greater positive joint work. For example, incline walking and stair ascent require greater activation of the lower limb extensor muscles, particularly the ankle plantar-flexors [6]–[8]. Elevated energy cost during activities of daily living can lead to a less active lifestyle and lower quality of life for those with Gross Motor Function Classification System (GMFCS) level I to III [9].

The ability to readily complete challenging walking conditions is paramount to increasing and normalizing activities of daily living for ambulatory children and young adults with physical disabilities [10]. Participation in many school and social activities can require navigation of stair and sloped terrains; encountering these terrains can be highly disruptive to activities of daily life for individuals with CP. Walking speed of individuals with CP is 66% of that of unimpaired individuals across level ground, while individuals with CP ambulate at only 55% and 50% of unimpaired speeds during incline and stair ascent, respectively [6], [11]. Additionally, biomechanical efficiency on stairs is over 6 times lower for individuals with CP compared to their unimpaired counterparts [12]. Difficulty walking on graded terrains likely contributes to reduced physical activity levels and quality of life for this patient population [13], [14].

Impaired strength, control, and function of the ankle musculature is often considered to be a major contributor to mobility deficits across many terrains and neurological conditions [15][16]. Plantar-flexor strength explained 50% of the variance in a timed stair test [17], while elevated calf muscle spasticity negatively affects stair ascent speed [18]. During incline walking, children with CP generated only 1/3rd of the

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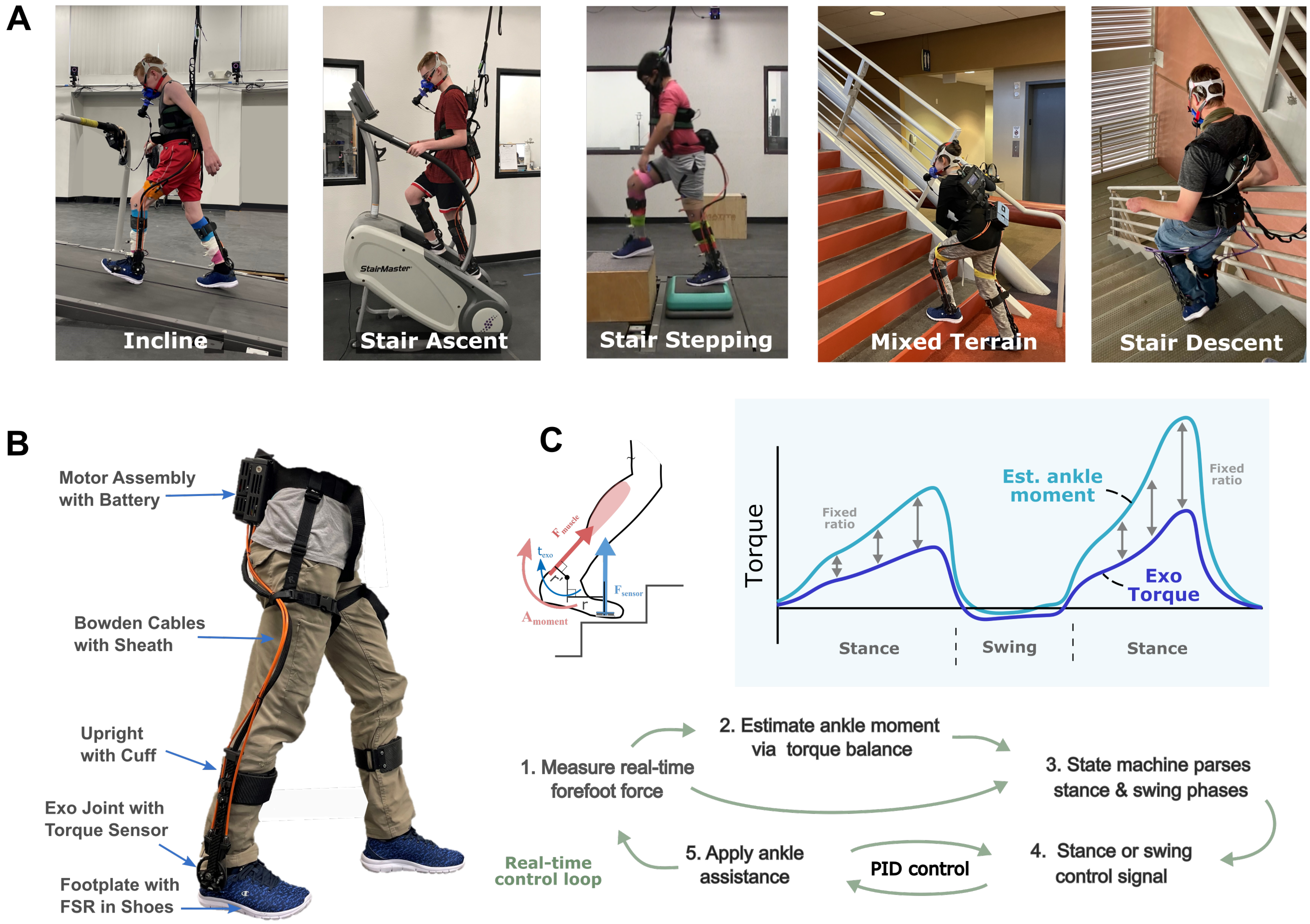

Fig. 1. A) Pictures of the experimental setup for 5° incline walking on a treadmill, stair ascent on a stepmill, instrumented stair stepping, real-world mixed-terrain walking, and stair descent. B) Pictures of the ankle exoskeleton components while worn on a model. C) Schematic depiction of the control scheme, real-time ankle moment estimation, and resulting adaptive torque profile.

push-off power relative to unimpaired children (0.48 vs. 1.32 W/kg), resulting in slower speeds and shorter steps [19]. AFOs, including leaf-spring, solid, and hinged designs, appear unable to improve stair climbing ability in CP [20].

Lower-limb exoskeletons may be able to improve ambulatory ability in individuals with or without gait impairments, which is a prominent goal for wearable robotic technology [21]. In healthy adults, ankle exoskeleton assistance reduced metabolic cost and soleus EMG during incline walking [22]. Knee exoskeleton assistance can reduce the metabolic cost of incline walking in healthy adults during load carriage [23], while hip exoskeleton assistance can reduce the metabolic cost of stair climbing in elderly adults [24]. We previously demonstrated that battery-powered ankle exoskeleton assistance can improve walking efficiency, speed, step length, and posture on level terrain [25]–[27]. However, we are not aware of any ankle exoskeleton research studies reporting outcomes on inclines or stairs in individuals with gait impairment.

Working towards our overarching objective of augmenting mobility in challenging free-living settings, the primary goals of this study were to (a) clinically-validate the safety, accuracy, and efficacy of adaptive ankle exoskeleton assistance on graded terrain in CP and (b) investigate the biomechanical mechanisms underlying changes in ambulatory performance. We hypothesized that walking with battery-powered ankle assistance would reduce metabolic energy consumption and muscle activity, and increase positive ankle power and power contribution across the lower-limb during these graded walking tasks compared to ambulating without wearing the device. An exploratory goal was to assess the feasibility of battery-powered ankle exoskeleton assistance in (a) a real-world mixed-terrain setting that combined level over-ground walking and stair ambulation, and (b) sustained staircase descent.

As the first evaluation of wearable exoskeleton assistance on graded and real-world mixed-terrain walking in individuals with neurological deficits, the primary contributions of this work include establishing (a) the feasibility of battery-powered exoskeleton assistance for improving performance across challenging terrain, (b) safety and benchmark performance results for real-world mixed terrain walking, (c) an optimal range of peak torque magnitudes for stair ascent and descent, (d) a clinically-validated and efficacious adaptive exoskeleton control scheme, and (e) mechanistic biomechanical factors influencing assisted walking performance, providing insight into future study and device designs.

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TABLE I
PARTICIPANT CHARACTERISTICS

Participant	Age (Years)	Sex	Height (m)	Mass (kg)	GMFCS Level	Incline Speed (m/s)	Stair Speed (floor/min)	Functional Ankle RoM (°)	Functional Knee RoM (°)
P1	11	M	1.5	48.4	I	1.05	3	21.4	59.7
P2	33	M	1.7	71.4	II	1.15	3	26.0	54.5
P3	15	M	1.65	57.2	I	1.1	2.7	20.8	64.1
P4	25	F	1.47	47.4	III	0.5	1.2	16.7	45.8
P5	14	M	1.48	39.5	II	0.9	3	26.4	39.2
P6	12	M	1.41	37.7	II	1	3	24.4	35.5
P7	14	M	1.65	55.8	II	0.9	2.7	23.1	54.5

GMFCS: Gross Motor Function Classification System; Functional Ankle and Knee RoM: range of motion during shod walking.

II. Methods and Procedures

A. Ankle Exoskeleton

Our battery-powered ankle exoskeleton, customized to the anthropometrics of each participant, was used to provide ankle untethered plantar-flexor and dorsi-flexor assistance. Details of the design can be found in our prior work [26], [27]. In short, waist-mounted motors (Maxon) remotely actuated a pulley at the ankle joint via a Bowden cable transmission (Fig. 1B). Carbon fiber footplates were attached to a torque sensor attached to the pulley. The waist assembly included a custom PCB with motor drivers, microcontroller, signal processing components, and a Bluetooth transceiver. A finite state machine was used to identify stance and swing phase of a gait cycle using custom embedded fore-foot sensors (Tekscan) (Fig 1C). During stance phase, the device provided adaptive plantar-flexor torque proportional to the real-time estimate of the biological ankle moment informed from the fore-foot sensors [28]. The torque profile was unique for each person, and the control signal adapted instantaneously to changes in each user's walking pattern, walking speed, and locomotor conditions because the ankle moment was estimated in real-time. To maximize the accuracy of biological ankle moment estimation, we administrated a second calibration immediately after the device began delivering torque. In this way, the influence of device actuation on the user's interaction with the force sensor was minimized. During swing phase, we provided dorsi-flexor assistance per user preference (~0.06 Nm/kg). The magnitude of dorsiflexor assistance was tuned by sequentially increasing and then decreasing the magnitude until (a) participants indicated a preference and (b) the research team observed adequate toe-clearance. The mass of the device was ~2.2 kg. Peak torque production capability was 30 Nm.

B. Assistance Level Validation

Using results from a prior investigation of level walking with the same controller [29], we specified 0.30 Nm/kg (roughly 30% of the biological moment) as the nominal peak setpoint for adaptive plantar-flexor assistance during incline walking. The nominal peak setpoint corresponded to the peak torque provided, usually at push-off, when each participant is walking at the speed and in the same pattern as they were when the calibration period was completed. The controller was adaptive, so if participants walked faster or slower, they would receive more or less assistance, respectively. To confirm that the same general nominal peak torque setpoint was appropriate for stair ambulation, we had our first participant complete a torque setpoint validation experiment by ascending the stair stepper (Stairmaster) and descending 4 flights of stairs (Fig. 1A, Fig. 5D) with low (0.20 Nm/kg), medium (0.30 Nm/kg), and high (0.45 Nm/kg) nominal levels of assistance. We calculated the net metabolic power for the last 3 minutes of the 5-minute stair ascent trials and the entire duration of the stair descent trials. The comparisons provided initial confirmation on the use of the 0.30 Nm/kg nominal setpoint for stair ambulation (See Results, below).

C. Participants and Study Design

Our device was designed to provide adaptive ankle exoskeleton assistance to the 71.3% of individuals with CP who can walk with or without an assistive device (GMFCS Levels I-III) [30]. Seven individuals with CP between 11 and 33 years of age participated in this study (Table 1, Supplemental Table 2). Inclusion criteria included diagnoses of CP; the ability to walk on a treadmill and stair stepper with or without a walking aid for at least 6 minutes; gross motor function classification system (GMFCS) level I, II, or III; at least 20° of passive ankle plantarflexion range of motion; no knee extension or ankle dorsiflexion contractures greater than 15°; no orthopedic surgery completed in the prior 6-month; and the absence of any medical condition other than CP that would affect safe participation. The study was approved by the Institutional Review Board of Northern Arizona University (NAU) under protocol #986744-27 on 12/03/2020 as a part of NCT04119063. Participants over 18 years old read and signed an informed consent document. For each minor, we obtained assent and informed written consent from a parent.

Following the consent process, we determined each participant's preferred "brisk" 5° incline and stair ascent speeds and had participants walk with exoskeleton assistance on the incline and stairs for 5-6 minutes each to acclimate to the condition and device (Fig. 1A). The 5° incline angle was based on American Disability Act (ADA) guidelines and selected because it is likely the most common ramp angle encountered during daily life. During formal data collection, each participant completed two separate sessions: 6-minute walking on 5° incline on a treadmill (Bertec) and 5-minute stair ascent (Stairmaster). For each session, participants completed two conditions at the same speed in a randomized order (Supplemental Table 4): Shod – walking wearing shoes, and Exo – walking with the exoskeleton as it provided bilateral plantar- and dorsi-flexor assistance.

A portable metabolic system (K5, Cosmed) was used to

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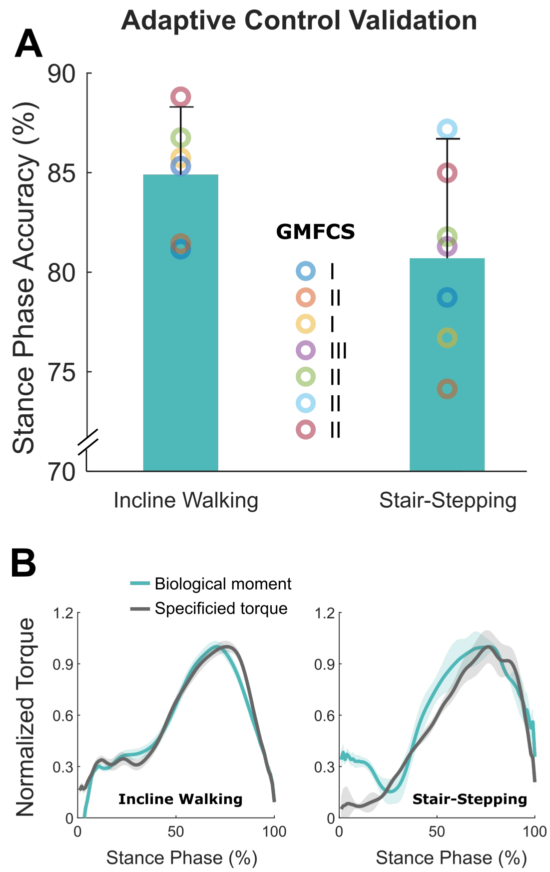

Fig. 2. Clinical validation of user- and terrain-adaptive ankle exoskeleton assistance. (A) Average controller accuracy relative to peak normalized biological moment during incline and stair ascent. (B) Profiles of the stance-phase biological ankle moment and specified exoskeleton torque during incline walking and stair-stepping with ankle assistance for participant CP1. Shading depicts mean ± standard deviation.

collect O_2 and CO_2 volume data during each task. Muscle activity was recorded from the tibialis anterior, soleus, and vastus lateralis using a wireless electromyography (EMG) system (Trigno, Delsys). We did not record gastrocnemius activity because the device's shank cuff blocked the necessary EMG sensor location, particularly for children with small muscle volumes.

To evaluate joint mechanics, we collected motion and force data for 20 seconds starting at the third minute of each 6-minute incline walking trial. Because we were unable to instrument the Stairmaster with force plates, we did not collect motion or force data within the 5-minute stair ascent trials. Instead, we asked participants to complete 6 trials of stair-stepping (3 trials leading with left leg and 3 leading with right leg) on blocks positioned over the instrumented treadmill's force plates (while treadmill was stationary, Fig. 1A) while we collected motion and force data. A ten-camera motion capture system (120 Hz; Vicon) with a custom marker set was used to record kinematic data [31], where markers were placed bilaterally on the torso, pelvis, and lower-extremity. We collected ground reaction force using the instrumented treadmill (960 Hz; Bertec). Experimental data were recorded simultaneously in the Vicon system and synchronized with exoskeleton data (100 Hz) using a trigger signal.

D. Biomechanical Data Analysis

Metabolic data and muscle activity were analyzed for the last 3 minutes of each trial [23]. Metabolic power was calculated from flow rates using Brockway's standard equation [32]. We subtracted the standing metabolic power from the average metabolic power of each condition [33], and then divided by body mass to obtain net metabolic power.

EMG data were band-pass filtered between 15 and 380 Hz, rectified, and low-pass filtered with a 7 Hz cutoff to generate the linear envelope [34]. We normalized the filtered EMG signal for each muscle by the peak value from walking with just shoes (Shod). EMG data were then segmented and normalized to percent gait cycle. The area under the normalized EMG curve (integrated EMG, iEMG) was summed across the gait cycle. EMG outcomes were averaged within limbs and across all gait cycles of the last 3 minutes of each trial.

Gait events (heel strike and toe-off) were identified in Vicon Nexus. We used OpenSim 3.3 [35] to derive joint angles and moments. We first scaled a generic musculoskeletal model [36] for each participant, and then computed the joint angles and moments across the lower-extremity using the inverse kinematics and inverse dynamics analyses, respectively. The biological ankle moment was found by subtracting the exoskeleton torque from the total ankle moment determined from inverse dynamics. Joint power (W) was calculated as the product of the joint moment (Nm) and the respective joint angular velocity (rad/s). We calculated stance-phase average positive power by integrating the positive area of the joint power curve and dividing stance time. The total limb average positive power was calculated by summing the average positive power at each joint (total = hip + knee + total ankle). Next, each individual joint's percent contribution to the total limb average positive power was calculated by dividing that joint's average positive power by the total limb average positive power. We focused our analyses on average positive power and positive power distribution of ankle, knee, and hip because they were closely related to energetics results. Biological (muscle-tendon and ligament contributions minus the exoskeleton's contribution) and total (combined biological and exoskeleton contributions) ankle moment and power at the ankle were computed for the exoskeleton-assisted trials [25]. Outcomes were normalized and averaged across all gait cycles during the 20 s of incline walking and by across the 3 trials for stair-stepping, and then averaged across limbs.

E. Controller Validation

To evaluate the accuracy of the adaptive joint-moment controller and confirm that providing torque does not decouple user intent, we computed the average stance-phase accuracy (1-root mean square error (RMSE)) between the peak normalized desired torque profile and peak normalized biological ankle moment for incline walking and stair-stepping.

F. Mixed-Terrain Study Design and Data Analysis

To explore the effects of adaptive ankle exoskeleton assistance on walking performance in a controlled real-world environment, we designed an indoor route that consisted of 330

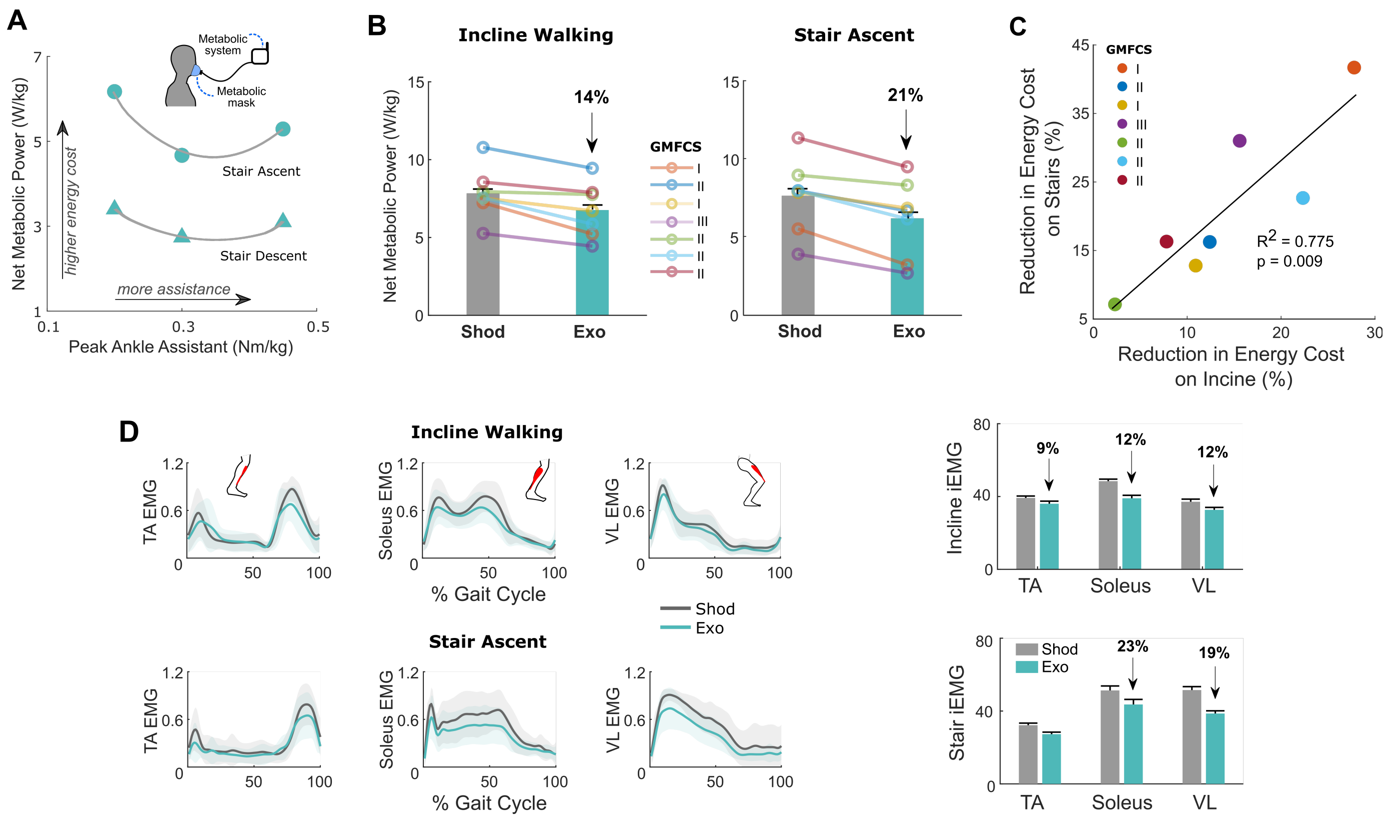

Fig. 3. A) Net metabolic power of our first participant with cerebral palsy (CP1) when walking with three levels of plantar-flexor assistance during stair ascent. B) Group and individual net metabolic power during incline walking and stair ascent with exoskeleton assistance (Exo) or shoes (Shod). Each line represents one participant. GMFCS: Gross Motor Function Classification System. C) Relationship between reduction in metabolic power between incline walking and stair ascent. D) Group averaged normalized tibialis anterior (TA), soleus, and vastus lateralis (VL) activity during incline walking and stair ascent with exoskeleton assistance or shoes. Shading depicts mean ± standard deviation. Bar charts indicate TA, soleus, and VL integrated muscle activity (iEMG) of the whole gait cycle. Error bars represent standard error of the mean.

meters of level ground and 3 sets of stairs (24 steps each, 72 steps total). Participants walked under three conditions in randomized order: Shod – walking wearing shoes, Exo – walking with the exoskeleton as it provided bilateral plantar- and dorsi-flexor assistance, and Zero-Torque – walking with the exoskeleton when the footplates were disconnected from the pulleys. We had participants complete walking bouts prior to testing under each condition, only starting a trial once participants indicated that they were (re)acclimated to that walking mode and following a 15-minute rest. During testing trials, we instructed participants to walk at their natural comfortable walking speed under each condition. We measured the time each participant took to complete the route and metabolic data of each condition. Walking speed was calculated by dividing the distance by the completion time. Net metabolic cost of transport was calculated by dividing the net metabolic power by the average walking speed.

To explore the effects of adaptive ankle exoskeleton assistance on stair descent, we conducted a pilot test on three participants within the cohort by having them walk down 4 flights of stairs (24 steps each, 96 steps total) in two conditions: Shod – walking wearing shoes, and Exo – walking with the exoskeleton as it provided bilateral plantar- and dorsi-flexor assistance. We measured the time each participant took to complete the route and metabolic data of each condition. Walking speed was calculated by dividing the total steps by the completion time. Net metabolic cost of descent was calculated by dividing the net metabolic power by the average walking speed. The stair stepping machine did not operate in the direction for descent, thus requiring the use of the staircase.

G. Statistical Analysis

We used paired two-tailed t-tests to compare outcomes between the Shod and Exo conditions for incline walking, stair ascent, and stair-stepping. For mixed-terrain walking, we used repeated measures one-way ANOVA to compare across Shod, Exo, and Zero-Torque conditions. Pairwise comparisons were then used if there was a significant condition effect. Kolmogorov-Smirnov tests were used to check the normality of the outcomes in each comparison. For parameters that were not normally distributed, we used Wilcoxon signed-rank tests to evaluate statistical significance. In post-hoc exploratory analyses, we used linear regression to assess the relationship between change in energy cost of incline walking and stair ascent, between soleus iEMG and energy cost, and between peak hip moment and energy cost for incline walking, and between baseline energy cost and change in energy cost of mixed-terrain walking. We checked each regression analysis dataset for outliers in using Cook's distance. Cohen's d was used to calculate effect size (ES), where 0.2 was considered a small effect, 0.5 a medium effect, and 0.8 a large effect [37]. A

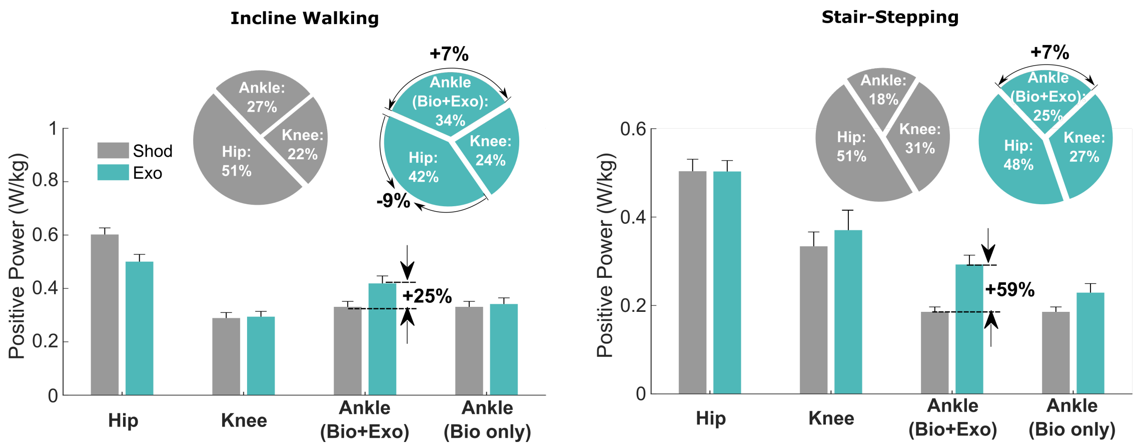

Fig. 4. Average positive power and power contribution of the ankle, knee, and hip during incline walking and stair climbing while walking with and without the exoskeleton. Error bars represent standard error of the mean.

fixed significance level of $\alpha < 0.05$ was used for this pilot study to allow individual interpretation of p-values.

III. Results

A. Controller Performance

All participants safely completed all assisted walking trials without falling or injury. The average accuracy of prescribed exoskeleton plantar-flexor assistance relative to the biological ankle moment was 84.9 ± 3.4% for incline walking and 80.7 ± 6.0% for stair ascent (Fig. 2).

B. Torque Setpoint Comparison

For stair ascent, the net metabolic power for low, medium, and high assistance was 6.17 W/kg, 4.67 W/kg, and 5.29 W/kg, respectively (Fig. 3A). For stair descent, the net metabolic power for low, medium, and high assistance was 2.13 W/kg, 1.88 W/kg, and 2.45 W/kg, respectively (Fig. 3A).

C. Energetics and Muscle Activity

Walking with exoskeleton assistance reduced net metabolic power by 14.2 ± 8.6% during incline walking (p = 0.004, ES = 1.7) and 21.1 ± 11.8% during stair ascent (p = 0.001, ES = 2.5) compared to walking without the device (Shod) (Fig. 3B). There was a significant positive relationship between the metabolic reduction from walking with exoskeleton assistance during the incline and stair conditions; participants who had greater reduction in incline walking exhibited greater reductions in stair ascent ($R^2 = 0.775$, p = 0.009; Fig. 3C).

During incline walking, wearing the exoskeleton reduced muscle activity across the gait cycle of the tibialis anterior (9.4 ± 10.5%, p = 0.043), soleus (12.2 ± 5.2%, p = 0.006, ES = 1.9), and vastus lateralis (11.9 ± 10.5%, p = 0.029, ES = 1.1) compared to walking without the device (Fig. 3D). During stair ascent, exoskeleton assistance reduced iEMG of the soleus (23.0 ± 12.8%, p = 0.01, ES = 1.6) and vastus lateralis (18.8 ± 28.3%, p = 0.028) muscles compared to walking without the device; tibialis anterior iEMG was similar between walking with and without the exoskeleton (Fig. 3D). EMG profiles are reported in Supplemental Fig. 2.

D. Joint Mechanics

During incline walking, walking with assistance increased stance phase average positive total ankle power by 24.7 ± 14.3% (p = 0.007, ES = 1.5) compared to Shod, and increased ankle joint contribution to positive lower-limb power by 7.4 ± 5.9% (p = 0.016, ES = 1.3) compared to Shod; the hip joint contribution to positive lower-limb power decreased by 8.9 ± 8.4% (p = 0.031, ES = 1.1) (Fig. 4).

During stair-stepping, walking with assistance increased stance phase average positive total ankle power by 59.2 ± 36.2% (p = 0.014, ES = 1.3) compared to Shod, and increased the ankle joint contribution to positive lower-limb power by 7.2 ± 3.4% (p = 0.001, ES = 2.1) compared to Shod (Fig. 4).

E. Mixed-Terrain Performance

During the mixed-terrain walking experiment, there was a significant condition effect on velocity (p = 0.024); pairwise comparisons indicated a significant difference between Exo and ZT with participants walking 11.2 ± 13.5% faster (p = 0.034, ES = 1.0) with exoskeleton assistance compared to with the device providing zero torque (Fig. 5A, Supplemental Fig. 3). There was a significant condition effect on metabolic cost of transport (p = 0.016); pairwise comparisons indicated significant differences between Shod and ZT (p = 0.016, ES = 1.3) and between Exo and ZT (p = 0.035, ES = 1.0); compared to walking with the device providing zero torque, net metabolic cost of transport was 11.4 ± 8.6% lower with exoskeleton assistance and 10.2 ± 8.0% lower without the device. At the group level, there was no difference in walking speed or cost of transport between walking with exoskeleton assistance vs. without the device (Fig. 5B). There was a strong positive relationship between change in cost of transport with exoskeleton assistance and a participant's cost of transport while walking without the device (shod); the more impaired participants exhibited reductions in cost of transport, while the opposite was true for the less impaired participants (Fig. 5B).

During the pilot stair descent experiment, participants at each GMFCS level were able to safely descend the four flights with adaptive assistance; there were with minimal changes in walking speed or energy cost (Fig. 5D). The energy cost of stair

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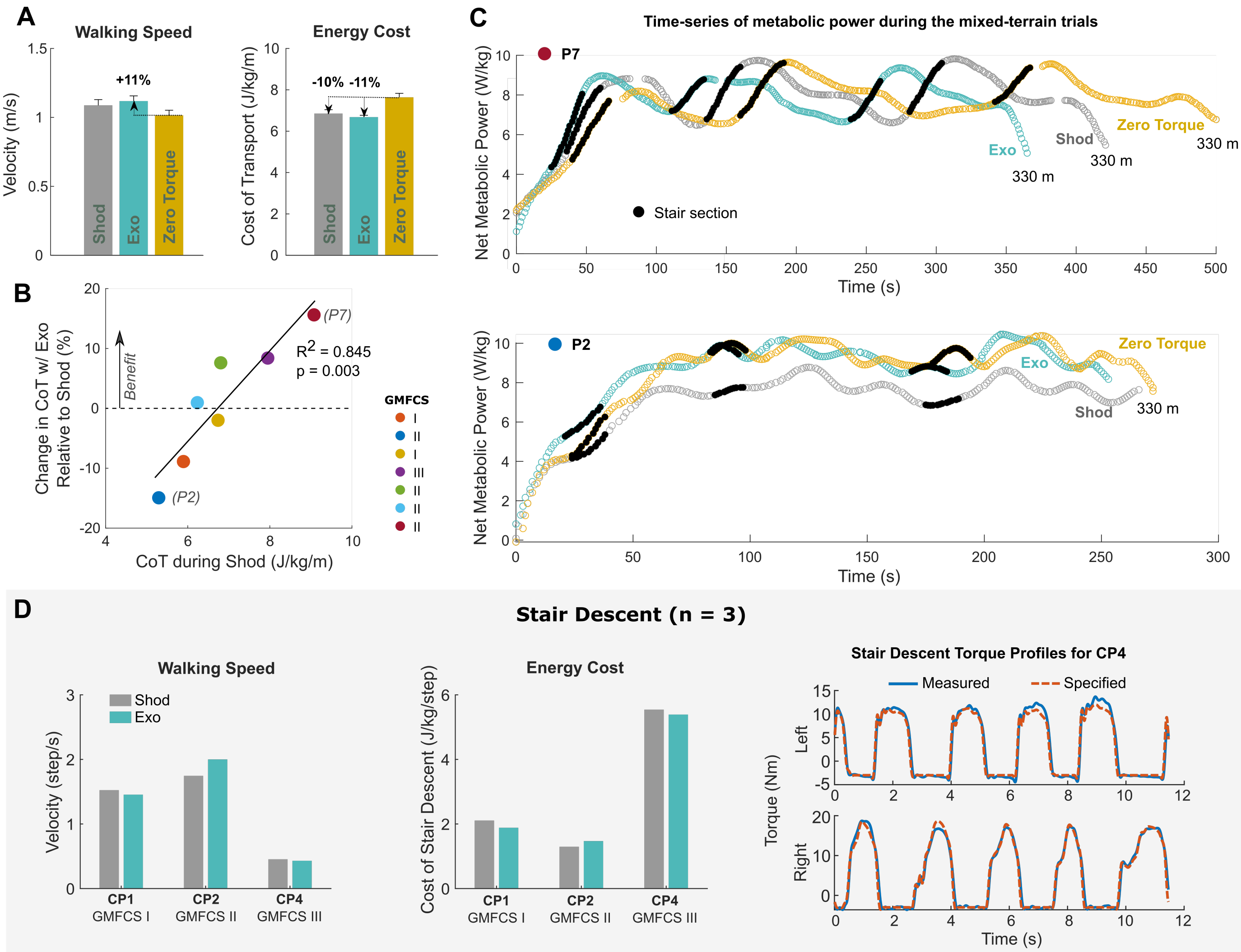

Fig. 5. A) Speed and energy cost of transport during mixed-terrain walking in Shod, Exo, and Zero Torque conditions. B) Relationship between the shod (i.e., baseline) cost of transport (CoT) for the mixed-terrain route and change in CoT with exoskeleton assistance relative to shod for the mixed-terrain route. C) Net metabolic power profiles of the best (CP7) and worst (CP2) responders showing the data collected across the mixed-terrain walking route in Shod, Exo, and Zero Torque conditions. Each point represents a data input from the breath-by-breath module of the metabolic system (K5, Cosmed). Filled data points indicate when the participant ascended the stair portions of the course. D) Walking speed and energy cost during stair descent in Shod and Exo conditions in three participants. Error bars represent standard error of the mean.

descent (with or without assistance) was ~50% lower compared to stair ascent.

IV. Discussion

We fulfilled the overarching goal of this study, which as to clinically-validate and demonstrate efficacy of battery-powered adaptive ankle exoskeleton assistance on controlled and real-world graded terrain in CP. Our adaptive controller achieved 81-85% ankle-moment-estimation accuracy and improved stead-state energy cost on inclines and stairs. The device and control system was safely used in real-world settings to improve mixed-terrain walking speed and energy cost compared to walking with the device providing zero torque. Our results suggest that more impaired individuals can quickly benefit from adaptive assistance in a mixed-terrain environment.

To the best of our knowledge, this is the first study to evaluate the effects of ankle exoskeleton assistance during graded terrain among people with CP. In the present study, the group level improvement in the energy cost of incline walking at 5° was 14%. This result is consistent with previous studies among unimpaired individuals, where providing ankle plantar-flexor assistance when walking on a similar incline grade (5.7°) contributed to a ~12% reduction in energy cost [22], [38]; however, these gains were from tethered devices, unlike the present study. Similar to incline walking, we found that ankle assistance reduced the energy cost of stair ascent by 21%. We found that participants who had greater metabolic reduction by wearing the device during incline walking also benefitted more from the assistance during stair ascent (Fig. 3C). Walking with the device improved ankle function by assisting in the generation of greater total ankle power (25% more for incline walking and 59% more for stair stepping), which likely contributed to the observed gait efficiencies. Limiting our ability to compare this work to prior research, we are not aware of any study that has reported the effects of ankle exoskeleton assistance or AFOs on measures of walking performance (e.g.,

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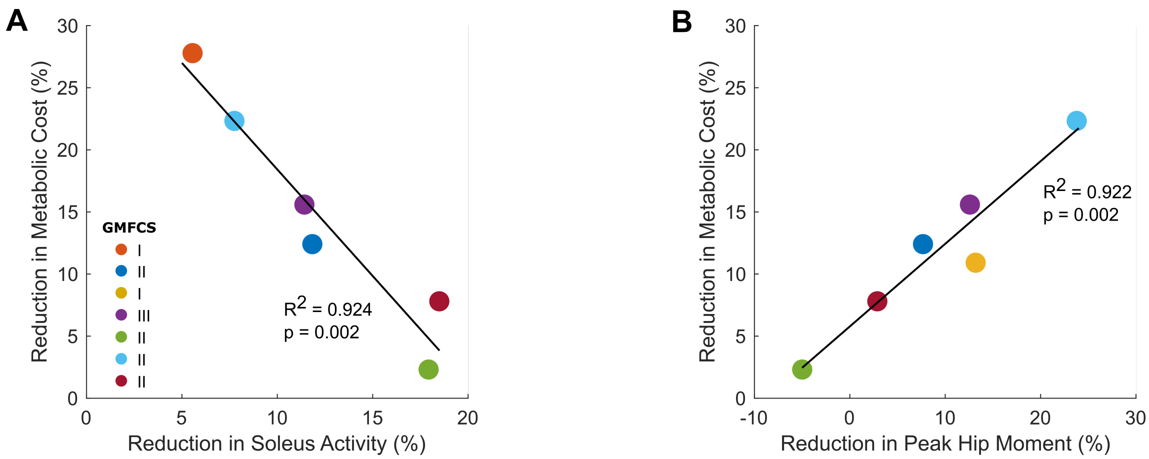

Fig. 6. A) Relationship between reduction in metabolic power and reduction in soleus activity with exoskeleton assistance compared to Shod during incline walking. Soleus activity data from one participant (P3) were not available because of a hardware issue (EMG sensor ran out of battery charge) during the experiment. B) Relationship between reduction in metabolic power and reduction in peak hip power with exoskeleton assistance compared to Shod during incline walking. Data from one participant (P1) were flagged as an outlier and excluded based on Cook's distance.

energy cost) across graded and mixed terrains in CP.

Ankle assistance reduced soleus activity by 12% during incline walking and 23% during stair ascent. Among unimpaired individuals, soleus activity was decreased by 16% while walking on an incline (5.7°) with an ankle exoskeleton compared to shoes [22]. Together, these results suggest that external assistance can replace some of the muscular demand on the plantar-flexor muscles for unimpaired individuals and individuals with neurological impairment alike. Somewhat surprisingly, our participants that exhibited larger reductions in soleus activity had less of a reduction in energy cost during incline walking ($R^2 = 0.924$, $p = 0.002$, Fig. 6A). Additionally, we found a significant positive relationship between the amount of energy reduction and the amount of hip moment reduction during incline walking ($R^2 = 0.921$, $p = 0.002$, Fig. 6B). This highlights the influence of ankle assistance across the entire lower-extremity, and speaks to the complex interconnectedness of lower-extremity joint and muscle function across graded terrain in this patient population.

Many individuals with CP, including many of our participants, walk in a crouched posture [39]. Ankle assistance affected the activity of the knee extensor muscles in addition to the plantar-flexor muscles. A plausible explanation for why ankle assistance reduced activity of the knee extensor muscles in addition to the plantar-flexor muscles could be that ankle actuation while the foot is firmly planted on the ground may help drive the tibia backward and assist in knee extension, particularly for individuals in crouch gait. We found that the device significantly reduced vastus lateralis activity during steady-state incline walking (12%) and stair ascent (19%). This finding corroborates our previous findings that ankle exoskeleton assistance can alleviate the burden on the knee extensor muscles during level walking [25], [40], with perhaps even greater benefits during more intensive activities.

We provided dorsi-flexor assistance during swing phase because many of our participants exhibited toe-drop during that interval of the gait cycle. A prior study found that providing plantar-flexor assistance can cause undesirable activation of the tibialis anterior [38]. The reduction of tibialis anterior activity across the gait cycle and increased ankle dorsiflexion during swing phase (Supplemental Table 1, Supplemental Fig. 1) for incline walking provides justification for the timing and magnitude of plantar- and dorsi-flexor assistance from our device.

As the first step towards evaluating the performance benefits of adaptive ankle assistance in free-living environments, the focus of our exploratory mixed-terrain work was on feasibility and safety. All of our participants were able to safely complete the mixed-terrain walking route that consisted of 330 m of level ground and 72 steps with and without adaptive ankle assistance. Walking with adaptive assistance helped participants walk faster and with lower energy cost compared to walking with the device unpowered. These are encouraging results that demonstrate the potential for battery-powered ankle assistance to be safely used in free-living settings.

We did not observe a significant group-level benefit relative to walking without the device. However, we did observe a marked benefit for our more impaired participants (Fig. 5B), a finding consistent with our prior work [26], [27]. The relationship between our participants' Shod (i.e., unassisted baseline) energy cost and change in energy cost with assistance during the mixed-terrain trial provides valuable information on the participants' characteristics that can be used to determine the candidates most likely to benefit in future intervention studies and with real-world deployment.

We achieved our goal for the pilot stair descent experiment, which was to demonstrate that the device was safe to use when descending stairs across GMFCS levels I-III. We did not expect the device to reduce the energy cost of stair descent because ambulating down stairs is generally not a very metabolically intensive activity (descent was ~50% less demanding than ascent for this cohort). Our torque setpoint experiment demonstrated that 0.3 Nm/kg of peak nominal assistive torque was the most beneficial magnitude for both stair ascent and descent. This speaks to the benefit of our adaptive controller and indicates that there is no need to adjust the nominal setpoint across different terrains.

It is important to consider that our mixed-terrain results reflect a single snapshot in time. In prior work, we observed performance improvements during level over-ground walking following 120 minutes of acclimation across 4 sessions [26], [27]. Therefore, it is likely that all of our participants would similarly demonstrate performance benefits during the mixed-terrain walking following additional acclimation. Compared to stair ascent wearing a harness on a stair stepper in the lab, a repetitive activity in a well-controlled setting, ascending real steps may have caused participants to walk more cautiously out of consideration for the significant safety consequences. We expect that longer acclimation to real-world stair ascent with the device would have resulted in increased psychological comfort, translating to additional performance benefits for the mixed-terrain trials.

A limitation of this pilot study was the relatively small sample size. Readers should be aware that the consistent group-level changes apply to a small cohort of participants; our results should be interpreted with caution. With the success of this feasibility study, we encourage future investigations in larger cohorts. Another limitation was that we did not measure hip extensor muscle activity, which would be necessary to determine how ankle exoskeleton assistance affects hip compensations that may result from ankle deficiency; our future studies will include assessment of this muscle group. Finally, speed was kept constant for steady-state incline walking and stair ascent across both Shod and Exo trials. While we expect participants may have naturally walked faster with assistance during these conditions, keeping the speed constant was necessary to isolate the effect of the device on energy cost.

V. Conclusion

In conclusion, our results suggest that wearable adaptive ankle exoskeleton assistance can safely and effectively increase ankle function and improve the energy cost of steady-state incline walking and stair ascent in individuals with CP. With improvements recorded after only ~5-10 minutes of practice, we found that participants acclimated quickly to the adaptive control algorithm during the graded walking conditions. Assessment of changes in muscle activity while completing the assisted walking trials indicated reduction in muscular effort across both the ankle and knee joints. We demonstrated that our device and adaptive control scheme allowed all participants to safely navigate level ground and stairs in controlled real-world environments, with our more impaired participants exhibiting a marked benefit. Stair descent with adaptive assistance was safe, but few changes relative to walking without the device were noted. Future work will include exploring the potential factors influencing participant responses (e.g., motor learning), completing more comprehensive investigations on the effects of torque magnitude gains on stability during these tasks, and implementing longer duration mixed-terrain walking in controlled settings, and eventually longer-term testing in uncontrolled free-living environments.

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