Ankle Exoskeleton Assistance Can Affect Step Regulation During Self-Paced Walking

Santiago Canete, Elizabeth B. Wilson, and Daniel A. Jacobs

***Abstract*— Exoskeleton assistance can reduce metabolic cost and increase preferred walking speed in unimpaired and impaired groups, but individual outcomes are highly variable. Assistance may influence step regulation, leading to individual modulation of gait variability, energetic cost, and balance control. In this study, we aimed to understand the effects of a powered ankle exoskeleton on step regulation and its relationship to self-selected walking speed, cost of transport, and gait variability. We asked 12 unimpaired young adults to walk at their comfortable walking speed on a self-paced treadmill in their regular shoes, with the exoskeleton tracking zero torque, and in two trials using proportional myoelectric control. We measured preferred walking speed, cost of transport (COT), mean and standard deviation of gait parameters, (step length, step time, and step width) and computed long-term correlations via detrended fluctuation analysis (DFA). In all exoskeleton trials, subjects walked significantly slower than in their shoes. However, the COT was equivalent between shoes and both proportional myoelectric control trials. Subjects also increased medio-lateral balance control by increasing their mean step width and reducing both short-term variability and long-term auto-correlation for this parameter. In the second powered trial subjects returned to the levels of control over step width exhibited during regular shoe walking. During the unpowered condition subjects showed a significant association between step width regulation, walking speed, and COT. However, these parameters were not significantly associated when the assistance was turned on. Together, these results demonstrate that the response to assistance is closely related to the stepping strategy, especially in the initial stages of learning.**

***Index Terms*— Exoskeleton, walking speed, self-paced treadmill, cost of transport, step regulation.**

I. Introduction

EXOSKELETONS and powered assistive devices can enhance human performance and rehabilitation. Often, the goal with lower limb exoskeletons has been to reduce energy expenditure during walking, with several devices achieving substantial success [1], [2], [3], [4]. Although energy expenditure is an important aspect, it is not the only objective that humans consider during gait. Humans simultaneously balance the demands of walking speed [5], [6], temporal cost [7], and balance [8], [9]. These objectives are competing and are modulated by different gait parameters (i.e. changes in step length, step frequency, and step width). When presented with new stimuli (e.g., changing ground dynamics, visual input, or robot-assistance), people may adjust foot placement [10], [11] to maintain lateral balance or modulate variability to deal with the changing environment [12], [13]. Selecting the movement objective is person-specific and can vary greatly between individuals [7], [14].Thus, understanding the underlying step regulation strategy in gait is critical to unlocking the potential for assistive devices to augment movement in unimpaired populations and improve rehabilitation for clinical populations [15].

A fundamental challenge for understanding step regulation strategies arises from the multiple ways in which humans can coordinate stepping to achieve the same task. For example, if a person prioritizes only forward walking speed, they could maintain the same speed through a large combination of step lengths and step frequencies. How humans modulate this redundancy is directly related to the movement objective, where task-relevant gait parameters are tightly regulated while task-irrelevant ones are allowed to vary more freely [16], [17]. For instance, measurements of human gait demonstrate that energetic cost is roughly parabolic around the nominal, self-selected, values for step length, step frequency, and step width [5], [12], [18], [19]. However, when balance is included as an objective, a person may choose to widen their steps in response to dynamic changes in the environment [12], [20], even if it incurs an energetic cost penalty [14]. For individuals with impaired motor function, prioritizing balance over energetics is an effective strategy to reduce the chance of injury due to falling [8]. This suggests, that humans will modulate their control over gait parameters in accordance to their movement objective.

Exoskeleton devices have been used to reduce energetic cost [1], [2], [3], [4] or increase walking speed [21], [22], [23], [24] but the way they impact step regulation is still unclear. Currently, exoskeleton assistance leads to highly variable and conflicting outcomes even within a study population [25], [26], [27]. To use an assistive device, users must incorporate the tool into their internal model and identify a new

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movement strategy [28], but we do not fully understand how to separate changes in the user's prioritization of objectives from changes to the underlying cost landscape. Designing an exoskeleton to assist propulsion and lower energetic cost, may come at the expense of balance control. Exoskeleton assistance has been shown, in certain systems, to worsen dynamic stability [29], [30], suggesting a change in the cost landscape that requires a new gait strategy. As a result, it is important to consider if the user's gait modulation strategies align with the objectives of the exoskeleton assistance (e.g. lowering energetic cost).

To study the relationship between cost of transport, walking speed, and stepping regulation when assisted by an exoskeleton we used our previously developed self-paced treadmill (SPT) algorithm [31], where people can modulate their walking speed to explore their new cost landscape. SPTs are advantageous for studying step regulation because they permit rapid, user-controlled changes in gait that facilitate measurement of step regulation during gait. Fixed-speed treadmills have been shown to decrease variability and promote increased local stability during walking [32], which could obscure some of the effects of exoskeleton assistance over stepping regulation.

Similarly, the choice of controller in the exoskeleton may impact the step regulation strategy. Many exoskeleton studies use proportional myoeletric control [1], [33] or torque-based control [3], [24], [34] to provide assistance. While research involving direct comparisons between the controllers is sparse, there is evidence that the controller type impacts gait strategy [35]. For this study, we used a proportional myoelectric controller since it has the advantage of a direct physiological connection to the nervous system, and thus the timing and magnitude of the assistance are directly related to muscle activity [1]. It has been widely shown that within the range of walking speeds that humans generally adopt, the magnitude of muscle activity in the plantarflexors increases with walking speed [36]. However, in recent research using torque controllers and optimization, the torque profile has been scaled temporally to accommodate changes in stride timing, but the magnitude has been fixed or optimized to a constant value regardless of the user's chosen walking speed. [2], [24]. As a result, there is the potential for torque based controllers to anchor the gait strategies of the user to a set of predefined controller parameters.

To evaluate the underlying control modulating of gait parameters when assisted by the exoskeleton, we used a detrended fluctuation analysis (DFA) [16], [37], [38]. DFA can be used to measure the step-to-step auto-correlations observed in human walking, and thus provide key insights into the underlying gait strategies. Reduced DFA values, (i.e., increased anti-persistence) can be interpreted as an increase in control over a certain gait parameter [16], in other words, a deviation in one direction is more likely to be followed by a deviation in the opposite direction in the following step. By analyzing these shifts in control over stepping parameters, one can potentially identify temporal changes in the task priorities (e.g. lateral balance, energetic cost, or walking speed) when assisted by an exoskeleton.

The objectives of this study were: 1) to evaluate the effects of a myolectrically controlled ankle exoskeleton on self-selected walking speed and cost of transport, 2) to compare the long-term and short-term gait variability between walking with shoes and the exoskeleton on a self-paced treadmill (SPT), and 3) to investigate the relationship between changes in energetics, walking speed and step regulation. We hypothesized that when wearing the exoskeleton in zero-torque mode, the self-selected walking speed would be significantly lower than when wearing regular shoes. Furthermore, in the powered conditions, users would reduce the cost of transport compared to zero-torque but use individual preference to walk faster or slower than in their regular shoes. Second, we hypothesized that, when wearing the exoskeleton, subjects would increase balance priority by widening their steps and increasing their control over step width regulation. Lastly, there would be a positive relationship between cost of transport and changes in walking speed, and step width regulation when wearing the exoskeleton. In this case, more tightly controlling step width would result in an increase in cost of transport.

II. Methods

A. Data Collection

We recruited 12 young adults (4 female, 8 male, age: 23.58± 3.82, height (m): 1.77± 0.09, mass (Kg): 75.48± 15.25) with no history of neurological or musculoskeletal impairment to participate in this study. Prior to the experiment, subjects had no experience walking in a powered exoskeleton. All twelve subjects included for analysis were right-side dominant. Subjects provided informed written consent in accordance with the Temple University Institutional Review Board (IRB:28448).

We tracked the motion of the subjects using 16 motion capture cameras (sample rate: 120 Hz; Qualisys, Goteborg, Sweden) and 39 reflective markers (34 lower body, 5 upper body). We measured the ground reaction forces using a split-belt instrumented treadmill (sample rate: 1200 Hz; Bertec, Ohio, USA). We collected surface electromyography (sEMG) from muscles in the dominant leg (i.e soleus, tibialis anterior, lateral gastrocnemius, biceps femoris long head, rectus femoris, and vastus lateralis) using a wired amplifier system (Delsys, Massachusetts, USA). We fitted the subjects with a K5 portable respirometer (COSMED, Rome, Italy) to measure O_2 and CO_2 flow rates. A visual representation of the experimental setup is given in Fig. 1B.

B. Experimental Protocol

Subjects walked on a self-paced treadmill in four conditions performed in the following order: 1) no device i.e regular shoes (ND) for 8 minutes, 2) exoskeleton in zero-torque mode (ZT) for 8 minutes, 3) first exoskeleton trial with proportional myoelectric control (E1) for 15 minutes, and 4) second powered exoskeleton trial (E2) for 15 minutes. Both E1 and E2 were performed with the same exoskeleton and controller configuration. Before the first self-paced treadmill trial, each subject was given 5-minutes of acclimation to the self-paced

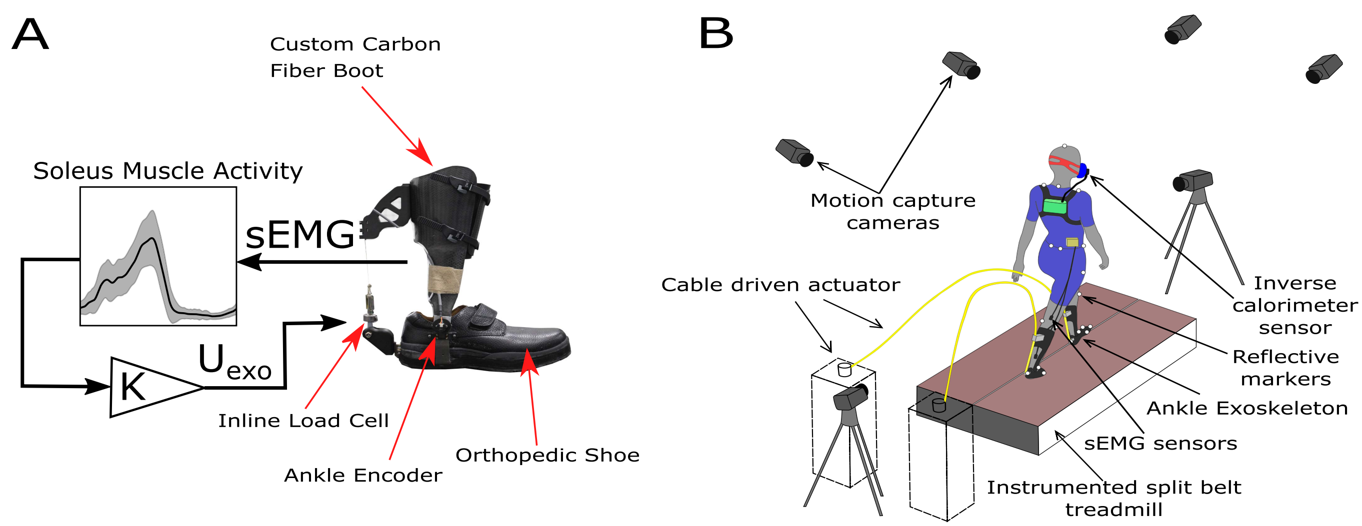

Fig. 1. Experimental setup: (A) During the assisted walking condition participants used a myoelectrically controlled ankle exoskeleton operating in closed loop with torque feedback. (B) Participants walked on a self-paced treadmill in three conditions (regular shoes, exoskeleton in zero-torque, and exoskeleton powered) where we measured their preferred walking speed, cost of transport, and collected motion capture data.

treadmill. In between trials, participants had a 5 minute resting period. Subjects were given standardized, explicit instructions: "walk at your comfortable walking speed during the entirety of the trial".

C. Self-Paced Treadmill

The self-paced treadmill algorithm used ground reaction forces for step detection, and kinematic data for walking velocity estimates. The algorithm used for the self-paced treadmill has been validated and described in previous work by the authors [31]. Briefly, the estimated walking speed of the user is calculated on a step-to-step basis by measuring step length and time, corrected for the portion of the step traveled during push-off before the leading leg contacts the ground. The steps were detected through the force load cells in the treadmill when the force measurement exceeded 5% of the subject's body weight.

D. Powered Ankle Exoskeleton

1) Hardware: We used a custom bilateral ankle-foot orthosis with one rotational degree of freedom assisting dorsoplantarflexion (Fig. 1A). The shank component is made out of carbon fiber and connects to the shoe's two pin-joints at the medial and lateral sides. The shoes have a steel plate inlaid in the midsole to transfer force directly to the ground. The total mass of each ankle-foot orthosis was 1.7 kg, and the medial distance from the ankle pin joint to the medial malleolus was approximately 3 cm.

The exoskeleton was actuated through an off-board cable-driven system (Humotech, Pennsylvania, USA) with a flexible Bowden cable transmission. The device provided plantarflexion assistance and the force exerted at the heel was measured by an inline load cell (Omega Engineering, Stamford, Connecticut). To measure the dorsiflexion angle of the exoskeleton, an absolute magnetic encoder was attached (MAE3, US Digital, West Virginia, USA) to the lateral side of the pin-joint at the ankle. The control signals to the actuators were generated through a Performance Real-Time target machine (Speedgoat, Maryland, USA) using Simulink 2020b (MathWorks, Maryland, USA).

2) Control: The myoelectric controller used raw sEMG signals of the soleus and tibialis and estimated the user's desired plantarflexion torque input as the difference in activity between the two muscles (Fig. 1A). The raw sEMG signals were processed through a linear envelope in real-time. The signal was high-pass filtered (2nd order Butterworth, cutoff frequency 80 Hz), full-wave rectified, and finally low-pass filtered (2nd order Butterworth, cutoff frequency 4 Hz). The signal was then multiplied by a static gain to achieve a 40% level of assistance based on the peak torques measured during a few steps of overground walking next to the treadmill. This was done only for tuning the controller gains and was not part of the experiment. The real-time controller used a proportional controller with damping injection [39] to compensate for the error between the user's desired torque and the measured ankle torque from the load cell and ankle encoder.

E. Data and Statistical Analysis

1) Walking Speed: The walking speed of the subjects was obtained from the recorded treadmill speed. Walking speeds were normalized to Froude number [40] by the following $FN = \frac{v^2}{lg}$. Where v is walking speed, l is the subject's leg length measured from the greater trochanter to the floor, and g is gravity.

2) Metabolic Cost: The instantaneous metabolic energy expenditure (MEE) [41] was calculated from O_2 and CO_2 flow rates. The MEE measurements were normalized for each trial by subtracting the measured MEE during quiet standing. To account for changes in speed during self-paced treadmill walking, the cost of transport (COT) was calculated with the following equation $COT = \frac{MEE}{v}$. Where MEE is the instantaneous metabolic energy expenditure, and v is the walking speed. The last three minutes of each trial were averaged to obtain a general estimate of COT.

3) Detrended Fluctuation Analysis: Detrended Fluctuation Analysis (DFA) was used to investigate the long-term correlation of the variability in step time, step length, and step width. [16], [37], [38]. DFA has been shown to avoid errors due to artifacts of nonstationaries in the time series [42]. DFA is used to calculate the value of the Hurst Exponent, $H_q(max)$ of the time series, where q is the order of the detrending polynomial. For monofractal time-series, the α value in DFA corresponds to the Hurst exponent $H_q(q = 2)$ [37]. However, monofractal and multifractal behaviors have been measured during gait with unimpaired and clinical populations [43]. Therefore, we evaluated each time series for multifractality by weighting the fluctuation functions in a range of q-order weights to avoid numerical errors in the tails of the multifractal spectrum [44]. We used the multifractal implementation of Ihlen [38] to calculate the width of the multifractal spectrum W, expressed as the difference between $H_q(5)$ and $H_q(-5)$. Small widths are considered monofractal. All of the gait parameters evaluated have spectrum widths less than 0.35 and were considered monofractal for the purposes of this study.

The time-series were separated in windows of sizes (s) ranging from 4 to N/4 (N: total number of steps) [16]. The integrated time-series was fitted by a polynomial of order 2. The logarithmic plot of the fluctuation function $log[F_q]$ vs $log[s]$ was fitted with a line where the slope corresponded to $H_q(2)$. When $H_q < 0.5$ it indicates anti-persistence (negatively autocorrelated), when $H_q = 0.5$ it indicates uncorrelated white noise, and when $H_q > 0.5$ it indicates persistence (positively autocorrelated). For this study, since the gait data appeared to be monofractal, only the $H_q(q = 2)$ value was analyzed and it is referred to as α. As α decreases closer to 0.5 the time-series is said to be uncorrelated, or in other words, a deviation in one direction is more likely to be followed by a deviation in the opposite direction.

A threshold was set to avoid the effects of local fluctuations that arise from marker tracking errors. It is common for the fluctuation function to have greater variability near the extremes of the sampling window range. Thus, the fitted polynomial may be skewed toward noise in the smaller sample windows [38]. Since fluctuations smaller than the equipment capabilities should not be considered, position measures less than 1 mm (marker position accuracy) and temporal measures less than 0.0083 sec (120 Hz capture rate) were set to 0.

Often, when computing α values in gait studies it has been done over strides instead of steps. The reason for choosing strides over steps is unclear. However, it has been shown that walking speed may not have a significant influence on α values for step length but may result in a significant difference for stride length [45]. For this study steps versus strides were used. The number of steps used varied across trials, averaging $N \approx 800$ per trial. 40 steps were removed from the beginning of the trials to ensure that the acceleration period was not included [31].

4) Statistical Analyses: All statistical analyses were performed in JMP® Pro 15 (SAS Institute Inc., Cary, NC, 1989-2022). All statistical tests were set to a significance level of 0.05.

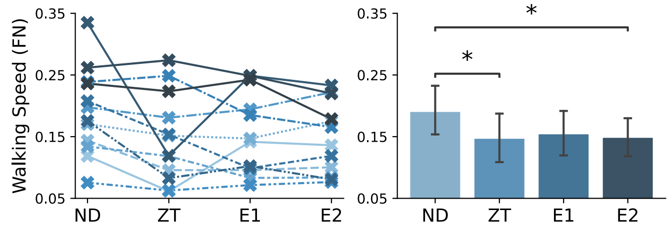

Fig. 2. Mean walking speed (normalized to Froude number (FN)) of 600 steps for 12 subjects at the four conditions: no Device (ND), exoskeleton in zero-torque mode (ZT), exoskeleton powered trial 1 (E1), and exoskeleton powered trial 2 (E2). The bar plot shows: mean ± std. 1) for ND the mean walking speed (FN) was 0.191±0.071, 2) for ZT 0.148±0.072, 3) for E1 0.155±0.067, and 4) for E2 0.149±0.058. The error bars indicate ±95% confidence intervals. The asterisks represent a significant difference (p<0.05). Overall, subjects chose to walk slower in all exoskeleton conditions, but there were no persistent patterns across individuals.

A linear, mixed-effect model was used to test the effect of the exoskeleton on COT, walking speed, and gait parameters. The fixed effect was the exoskeleton condition and the random effect was the subjects. When the fixed effect was significant, a Tukey's honest significant difference pairwise comparison was conducted between conditions (ND, ZT, E1, E2).

The relationships between COT, walking speed, and step regulation was evaluated through a linear regression and the linear correlations of the model. The slope, intercept, and correlation values were considered statistically significant if their p-value was smaller than 0.05.

III. Results

A. Walking Speed

There was a significant fixed effect on self-selected walking speed for the condition (DF=33, F-Ratio=4.24, p=0.012), and there was a significant random effect for the subject (p=0.032). Across subjects, the comfortable walking speed was significantly lower in ZT compared to the ND condition (p=0.02). Comfortable walking speed also significantly decreased in E2 (p=0.027), while in E1 it did not change significantly compared to ND (Table I). Although there were significant differences across subjects, when looking at the individual response to the exoskeleton per subject there were variable outcomes (Fig. 2). Some subjects walked faster when the exoskeleton was powered while others walked slower. One subject walked faster in the ZT condition than in any other condition.

B. Cost of Transport

There was a significant fixed effect on COT for the condition (DF=33, F-Ratio=37.9, p<0.0001), and there was a significant random effect for the subject (p=0.04). Across all subjects, the COT was significantly lowered with respect to the ZT condition (Fig. 3). Between the ND and ZT conditions, the difference in COT was −0.24 (p<0.0001). Between E1 and ZT conditions the difference was −0.22 (p<0.0001). Between E2 and ZT conditions the difference was −0.22 (p<0.0001)

TABLE I
COT AND MEAN WALKING SPEED BY CONDITION

		Cost of Transport		Mean Walking Speed (FN)	
Condition 1	Condition 2	p-value	Difference (Confidence Interval)	p-value	Difference (Confidence Interval)
ND	ZT	0.0001*	-0.24 (-0.31, -0.17)	0.02*	0.044 (0.0054, 0.082)
ND	E1	0.96	-0.013 (-0.085, 0.058)	0.067	0.036 (-0.0019, 0.074)
ND	E2	0.96	-0.012 (-0.084, 0.059)	0.027*	0.042 (-0.0038, 0.08)
ZT	E1	0.0001*	-0.22 (-0.29, -0.15)	0.95	-0.0073 (-0.045, 0.031)
ZT	E2	0.0001*	-0.22 (-0.30, -0.15)	0.99	-0.0016 (-0.04, 0.037)
E1	E2	1.00	-0.001 (-0.072, 0.07)	0.98	0.0057 (-0.032, 0.044)

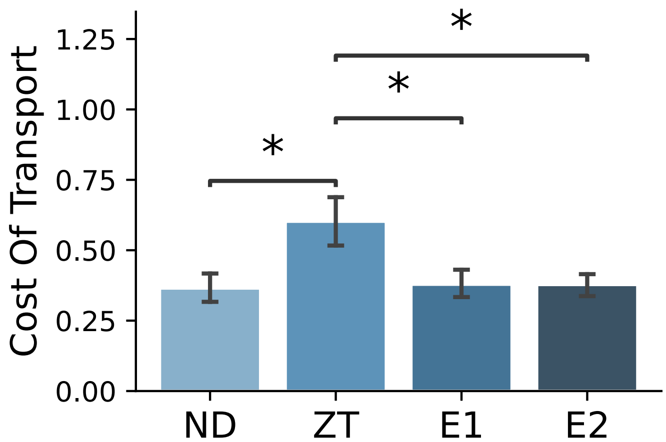

Fig. 3. Mean Cost of Transport (COT) for the four conditions (No Device (ND), zero-torque (ZT), exoskeleton powered trial 1 (E1), and exoskeleton powered trial 2 (E2)). The plot shows: mean ± std. 1) for ND the COT was 0.36±0.095, 2) for ZT 0.61±0.16, 3) for E1 0.38±0.09, and 4) for E2 0.38±0.072. The error bars indicate ±95% confidence intervals. The asterisks represent a significant difference ($p<0.05$). COT increased in ZT with respect to ND, and when the exoskeleton was powered returned to the levels of the ND condition.

(Table I). The cost of transport during the study was within previous measurements for treadmill walking (0.3-0.6) [46], [47]. There was no statistical difference between E1, E2, and ND with all $p>0.9$.

C. Step Regulation

On both step time (DF=33, F-Ratio=9.86, $p<0.0001$) and step width (DF=33, F-Ratio=6.63, p=0.0013) there was a significant fixed effect for the condition, and there was a significant random effect for the subject (step time: p=0.033; step width: p=0.028). For the three exoskeleton conditions (ZT, E1, E2), subjects took slower steps compared to the ND condition. The mean step time across subjects was 0.057 s greater for the ZT condition (p=0.0002), 0.049 s for E1 (p=0.0015), and 0.052 s for E2 (p=0.0006). The mean step width across subjects also increased compared to the ND condition by 2.59% for the ZT condition (p=0.0011), 2.02% for E1 (p=0.013), and 1.95% for E2 (p=0.017). For both step time and step width, there were no significant differences when the exoskeleton was powered (E1, E2) versus the ZT mode. Across all conditions, subjects exhibited similar mean step lengths with no statistical differences. The mean step length, step time, and step width are shown in (Fig. 4A, Table II).

Some of the differences observed in step width when wearing the exoskeleton could be explained by the increased width of the device at the ankle joint. The measured mean width of the exoskeleton at the ankle joint from the medial malleolus to the medial pin joint was 3.1±0.14% of leg length for each ankle-foot orthosis.

Previous studies have reported that step width may have an effect on walking speed [48], [49]. In order to verify that the changes observed in step width were not related to walking speed, we calculated the correlations (r) and R^2 values for speed vs. mean step width. There was not a significant correlation between mean step width and walking speed, for the ND condition (r: 0.24, p=0.45), for ZT (r: 0.17, p=0.59), for E1 (r: 0.034, p=0.92), and for E2 (r: -0.12, p=0.71). Similarly, the correlation of speed with std and alpha values of step width were also weak and not significant. This was in contrast to mean step length and step time, which did exhibit significant correlations to walking speed as expected.

In step width variability (DF=33, F-Ratio=11.16, $p<0.0001$), there was a significant fixed effect for the condition, and there was a significant random effect for the subject (p=0.033). For the three exoskeleton conditions, subjects significantly reduced their variability in step width in all exoskeleton conditions compared to ND. The variability in step width was reduced compared to the ND condition by -0.42% for the ZT condition (p=0.0007), -0.44% for E1 (p=0.0004), and -0.49% for E2 ($p<0.0001$). There were no significant differences when the exoskeleton was powered (E1, E2) versus the ZT mode with all $p>0.8$. Also, across conditions, there were no differences in variability for step length and step time. The variability in step length, step time, and step width is shown in (Fig. 4B).

In step width long-range correlations of variability (α) (DF=33, F-Ratio=7.33, p=0.0007), there was a significant fixed effect for the condition, but there was not a significant random effect for the subject (p=0.67). There was a significant statistical difference between α values of step width for ZT (p=0.0004) and E1 (p=0.045) compared to ND. There was also a significant difference between E2 and ZT (p=0.028). For step length and step time, we found no significant difference in α values (Fig. 4C).

Changes in walking speed were positively associated to changes in COT for all exoskeleton conditions when compared to shod walking (ZT: $r = 0.59$, p=0.043, $m = 0.78$, p=0.044, $b = -84.93$, p=0.001; E1: $r = 0.94$, p=0.001, $m = 0.71$, p=0.001, $b = -18.23$, p=0.001; E2: $r = 0.71$, p=0.01,

$m = 0.55$, p=0.009, $b = -16.98$, p=0.007) (Fig. 5). Changes in mean step width were negatively associated to changes in COT, but were not significant in any of the conditions (ZT: $r = -0.19$, p=0.52; E1: $r = -0.37$, p=0.21; E2: $r = -0.17$, p=0.57). For DFA, changes in α values were negatively associated to changes in COT, but only in the ZT condition was the correlation significant (ZT: $r = -0.78$, p=0.003, $m = -2.54$, p=0.003, $b = 36.42$, p=0.004; E1: $r = -0.19$, p=0.54, $m = -0.27$, p=0.54, $b = 3.34$, p=0.57; E2: $r = -0.41$, p=0.18, $m = -0.83$, p=0.18, $b = 3.00$, p=0.58). Changes in DFA were also negatively associated to walking speed, but similarly, only in the ZT condition they were significant (ZT: $r = -0.58$, p=0.05, $m = -1.43$, p=0.05, $b = -39.87$, p=0.002; E1: $r = -0.24$, p=0.45, $m = -0.45$, p=0.45, $b = -21.34$, p=0.017; E2: $r = -0.44$, p=0.15, $m = -1.13$, p=0.15, $b = -23.88$, p=0.005).

IV. Discussion

Humans, during gait, can benefit from exoskeleton assistance to reduce their energetic cost or increase their walking speed but often the individual outcomes are highly variable and even across exoskeleton devices the gait strategies adopted as a group differ substantially. People modulate their gait strategies to deal with multiple competing demands such as energetic cost, balance, or temporal cost. To do so, they exhibit different levels of control over task-relevant and task-irrelevant gait parameters [16], [17]. Here, we measured the effects of a powered ankle exoskeleton on step regulation and the impact these shifts in gait strategy could have over energetic cost and walking speed. Furthermore, we investigated the effects of a myoelectric controller on preferred walking speed and compared it to previous studies that used torque-profile controllers.

We observed a similar reduction in energetic cost to previous studies with tethered exoskeletons for both powered conditions [4], [25], [50], which suggests that the majority of the adaptation occurred in the first 15-minute period. The strong positive association between walking speed and cost of transport for all conditions (Fig. 5) implies that for this device, minimization of energy economy shifted to a slower walking speed than the user would select in their regular shoes. In general, participants needed to lower their walking speed when assisted by the exoskeleton by 18.23% in E1 and by 16.98% in E2 to lower their energetic cost to the levels exhibited during shod walking. The magnitudes of the slope and intercept were also larger in the first powered exoskeleton trial than in the second, which indicates that subjects may have been able to exercise a preference to walk faster while maintaining their energetic cost benefit, but as a group they chose not to.

Our hypothesis that when assisted by the exoskeleton, comfortable walking speed would be highly dependent on individual preference was partially supported. As a group, subjects significantly reduced their walking speed in the ZT and E2 conditions compared to ND, but there was not a persistent pattern across conditions and individuals (Fig 1). Between ND and E1, the difference was not statistically significant, but it approached significance with a similar magnitude as the second powered condition (E2) (Table I). These results show that while energetic cost plateaued in E2 compared to E1, participants continued to explore new walking speeds and gait strategies.

Our results differed from previous studies with unimpaired young adults, which showed that self-selected walking speed increased when assisted by an ankle exoskeleton [24], [29]. These conflicting results may be attributed to the differences in exoskeleton device and control. Previous studies have been conducted using controllers that applied a predefined torque-profile at the joint which did not vary in magnitude with walking speed. Exoskeleton controllers that create the same assistance profile at all walking speeds may provide implicit information to the subject and anchor their performance to a specific set of parameters. Young adults show a strong positive relationship between sagittal plane kinetics and walking speed [51], [52]. When changing walking speed, there are matching changes in peak ankle plantarflexion, knee extension, and hip flexion moments [53].

The observed reduction in walking speed when wearing the device seems to be primarily related to step time and not step length. One possible explanation for why step time increased when wearing the exoskeleton could be related to the increased weight. Browning et al. showed that as the weight at the foot increased, so did step time [54]. Adopting a strategy that increases step length and reduces step time can also have a negative effect on the mediolateral margin of stability of the user [55] and perhaps explain why we observed tighter control and a mean increase in step width when wearing the exoskeleton. Lastly, the changes in step width could also be related to the increased medial width at the ankle joint of the exoskeleton. The physical exoskeleton has only a minor increase in width around the ankle joint due to the ankle stirrup, and thus medial collisions are not likely. Furthermore, the measured increment in medial width caused by the exoskeleton's ankle joint was $6.2 \pm 0.28\%$ of leg length, which is well in the range of the step widths taken by the users ($16.22\% \pm 2.78\%$ of leg length) while walking in their regular shoes. In addition, the mean step width increase in the exoskeleton conditions with respect to ND was of 2.19% which is within one standard deviation of step width in ND. Therefore, we believe this is a voluntary strategy adopted by the user and not a reaction aimed at avoiding collisions due to the increase in physical width of the device. The adoption of this strategy could be related to increased caution [16], [56] to avoid collision at the ankle joint. Physical effects arising from the single rotation axis of the exoskeleton, as well as possible joint misalignment, can also be driving factors for the changes in step width. The inversion-eversion degree of freedom in the ankle joint plays an important role in the control of whole body angular momentum during gait [57], it is a redundant method available to humans alongside step width and plantarflexion force for stabilizing mediolateral balance. Experiments on unassisted gait have shown that using the inversion-eversion degree of freedom in the ankle joint can lead to reductions in step width [58], suggesting that the single rotation axis of the exoskeleton could have a confounding effect on step width.

TABLE II
STEP REGULATION

	Step Length (% leg length)	Step Time (s)	Step Width (% leg length)	Step Length Variability (% leg length)	Step Time Variability (s)	Step Width Variability (% leg length)	Step Length DFA (α)	Step Time DFA (α)	Step Width DFA (α)
ND	70.54±8.74	0.54±0.045	16.22±2.78	4.85±2.19	0.026±0.014	2.4±0.44	0.88±0.12	0.73±0.11	0.78±0.048
ZT	66.5±9.31	0.60±0.064	18.81±3.12	5.08±1.85	0.026±0.013	1.99±0.5	0.80±0.096	0.73±0.097	0.68±0.053
E1	68.79±8.6	0.59±0.06	18.24±3.75	4.37±1.61	0.026±0.0098	1.96±0.43	0.82±0.067	0.75±0.075	0.72±0.07
E2	66.5±7.96	0.59±0.052	18.17±3.74	3.85±1.16	0.023±0.0082	1.91±0.38	0.85±0.067	0.80±0.11	0.75±0.044

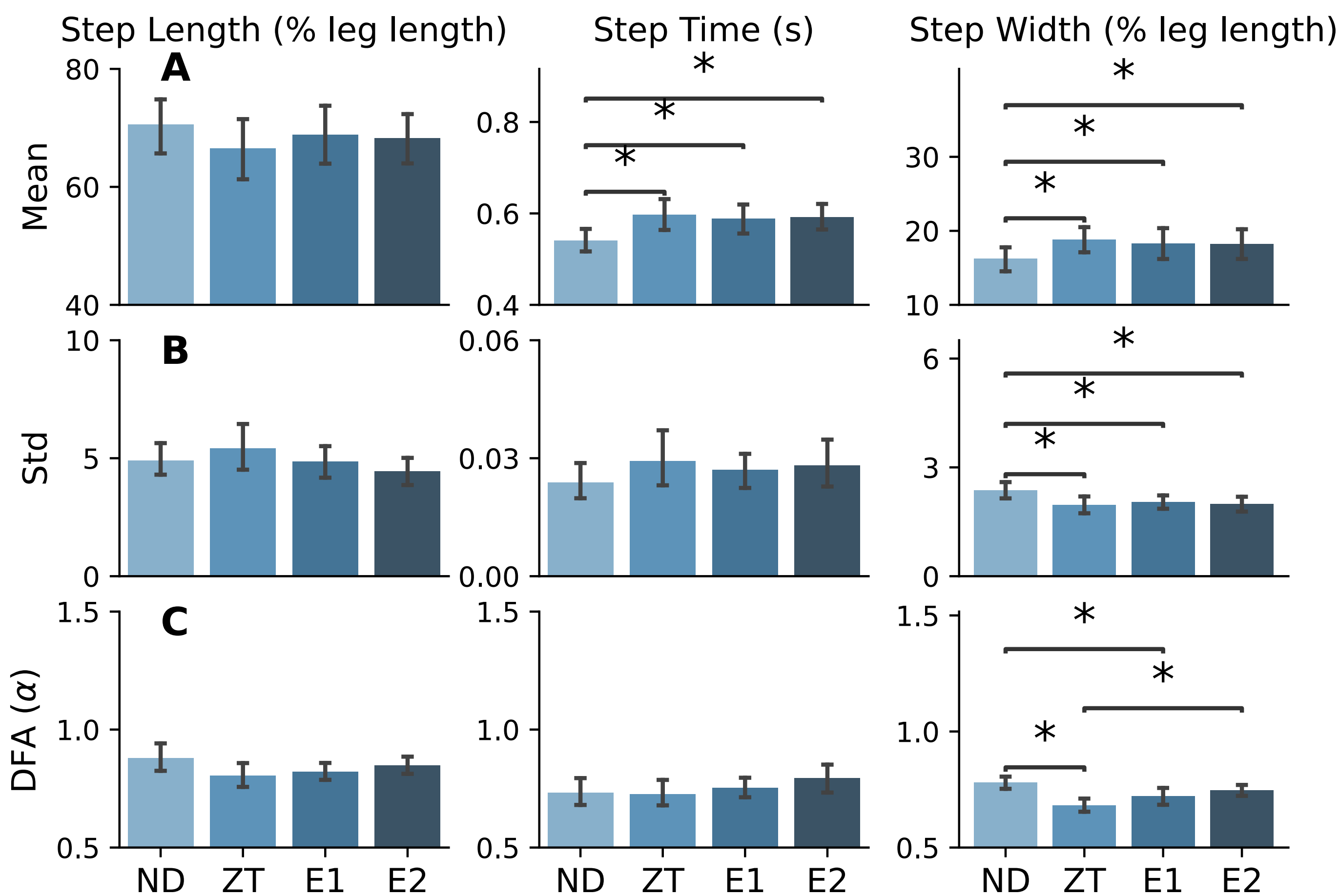

Fig. 4. (A) Mean, (B) Std, and (C) DFA values for step length, step width, and step time for 12 subjects walking at their self-selected walking speed at the four conditions. No Device (ND), exoskeleton in zero-torque mode (ZT), exoskeleton powered trial 1 (E1), and exoskeleton powered trial 2 (E2). Step length and step width are shown as a percentage of the subjects leg length. The error bars indicate ±95% confidence intervals. The asterisks represent a significant difference ($p<0.05$). There was a significant increase in step time and step width for all exoskeleton conditions compared to shoe walking. Variability in step width was higher in the ND condition than in the other conditions. For the DFA analysis, the α value was lower in E1 than in ND, but recovered in E2.

Our results partially confirmed our hypothesis that, when wearing the exoskeleton, participants could shift priority away from energy economy as the main objective. In this study, walking speed had a stronger relationship with energetic cost when in ZT, but subjects maintained higher variability and lower control over step length and step time than when the exoskeleton was powered or even when in ND (Tab. II). However, they significantly reduced variability and more tightly controlled their step widths. This strategy had important energetic consequences (Fig. 5, DFAvsCOT) but it was still preferred by participants. Once assistance was turned on, participants maintained the same step width as in the ZT condition, but in E2 they lowered the control over step width to the level exhibited during shod walking. Most importantly, the effects on energetic cost of tighter step width regulation appeared to be mitigated when the assistance was turned on (Fig. 5, DFAvsCOT). Because changes in step width away from the self-selected pattern drive increases in energetic cost [12], this suggests that part of the energetic benefit of the exoskeleton assistance was used to reduce the penalty associated with wider stepping.

Having the ability to explore the landscape of energetic cost and walking speed on a self-paced treadmill appears to be key to the adaptation within an exoskeleton. Previous

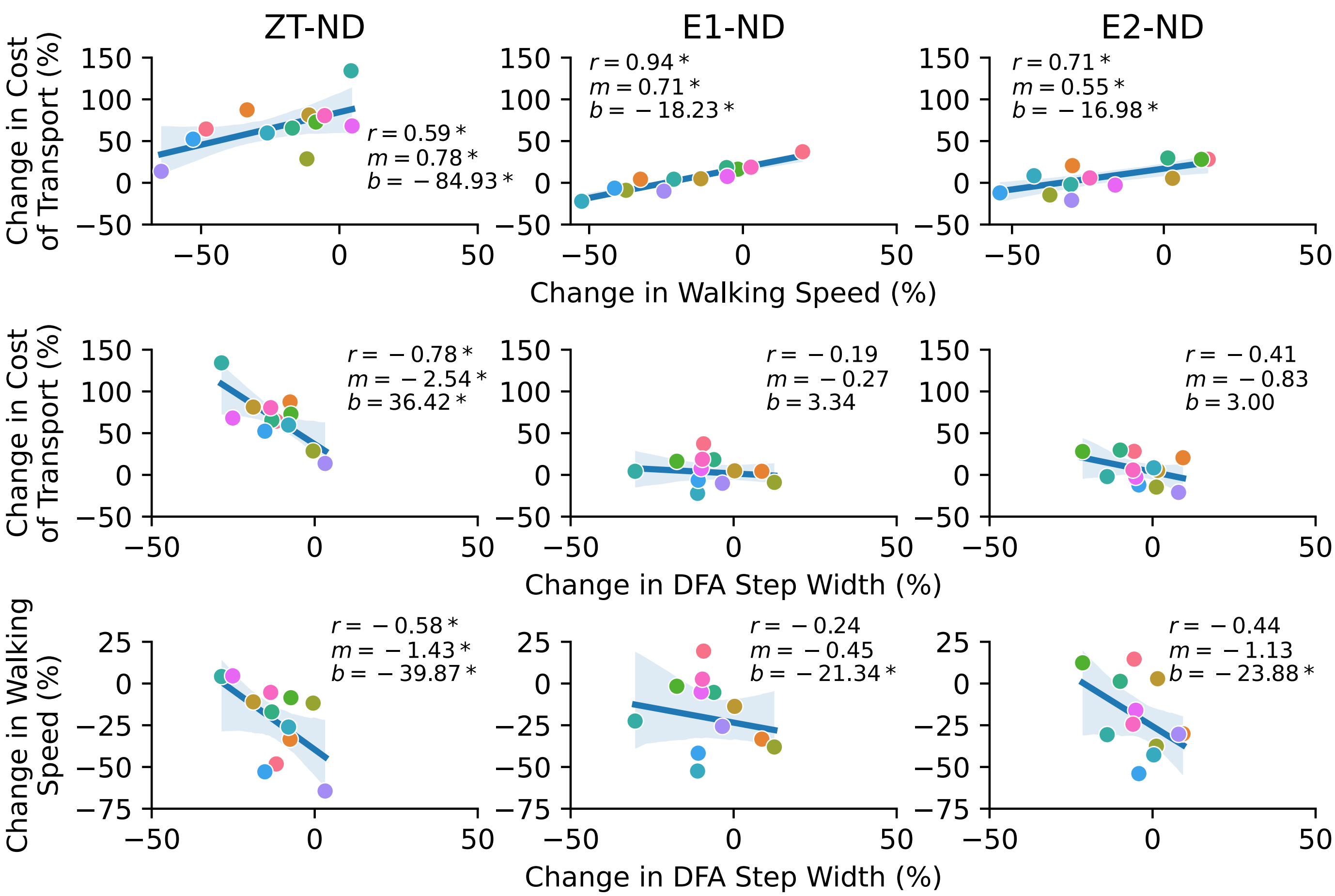

Fig. 5. Linear regression for the percent changes with respect to no device (ND) in cost of transport, walking speed, and step width DFA α values for zero-torque (ZT-ND), first powered exoskeleton trial (E1-ND), and second powered exoskeleton trial (E2-ND). The blue lines show the linear fits with their correlation coefficient (r), slope (m), and intercept (b) for the linear regression model, and each subject is represented by a different colored marker. The graphs show that cost of transport was influenced by walking speed, requiring subject to search for a new comfortable speed when wearing the exoskeleton. Increased step width control had a larger negative effect on cost of transport, specially when the exoskeleton was in zero-torque mode (ZT). Step width control was negatively correlated with walking speed, and cost of transport (i.e. as the control over step width increased the walking speed decreased and the cost of transport increased) but was only significant when the exoskeleton was in zero-torque mode (ZT).

investigations into the effect of manual and continuous tuning paradigms on assisted and unassisted gait [59], [60], [61] have shown that while the techniques have a strong correlation with each other, they do lead to differences in energetic cost. Because of the large differences in gait strategies and walking speed between regular shoes and powered conditions, we believe that the combination of proportional myoelectric control and a self-paced treadmill did not inhibit the necessary exploration required to identify the individual's preferred gait strategy.

One limitation with the present study is that we cannot conclude that the observed strategies are the optimal ones, neither for energy economy nor stability. Our results do demonstrate that the strategies adopted during exoskeleton walking are energetically detrimental, thus energy economy may not be the main priority during the task. Although the strategies adopted are consistent with those observed when balance is compromised [10], [11], [12], [13] (e.g. wider stepping, tighter control over step width, reduced variability) we cannot determine whether this is an intrinsic requirement or an individual choice that could be mitigated with training. Furthermore, even though we observed a trend in the linear models between step width regulation, walking speed, and COT, these were not significant for the powered condition, most likely due to sample size limitations.

People may adjust their stepping regulation strategy when assisted by the exoskeleton, specifically leaning towards tighter control over step width. This preference appears to be needed in order to walk at speeds equivalent to those exhibited during the regular shod condition. Exercising tighter control over step width also had a negative effect on energy economy, explaining why walking at slower speeds was energetically more beneficial when assisted by our exoskeleton. In summary, when evaluating the energetic consequences of walking with exoskeleton assistance, it is important to consider if all users are prioritizing energetic cost as their main objective. Furthermore, the important mitigation of these effects when assistance was turned on compared to the unpowered condition shows that assistance could be beneficial to those that see their capacity to control their stepping diminished. Further research with impaired populations would be necessary to determine if exoskeleton assistance can benefit their capacity to regulate gait parameters.

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