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Walking Speed Changes in Response to Novel User-Driven Treadmill Control

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ABSTRACT

Implementing user-driven treadmill control in gait training programs for rehabilitation may be an effective means of enhancing motor learning and improving functional performance. **This study aimed to determine the effect of a user-driven treadmill control scheme on walking speeds, anterior ground reaction forces (AGRF), and trailing limb angles (TLA) of healthy adults**. Twenty-three participants completed a 10-meter overground walking task to measure their overground self-selected (SS) walking speeds. Then, they walked at their SS and fastest comfortable walking speeds on an instrumented split-belt treadmill in its fixed speed and user-driven control modes. The user-driven treadmill controller combined inertial-force, gait parameter, and position based control to adjust the treadmill belt speed in real time. Walking speeds, peak AGRF, and TLA were compared among test conditions using paired t-tests ($\alpha=0.05$). Participants chose significantly faster SS and fast walking speeds in the user-driven mode than the fixed speed mode ($p>0.05$). There was no significant difference between the overground SS walking speed and the SS speed from the user-driven trials ($p<0.05$). Changes in AGRF and TLA were caused primarily by changes in walking speed, not the treadmill controller. Our findings show the user-driven treadmill controller allowed participants to select walking speeds faster than their chosen speeds on the fixed speed treadmill and similar to their overground speeds. Since user-driven treadmill walking increases cognitive activity and natural mobility, these results suggest user-driven treadmill control would be a beneficial addition to current gait training programs for rehabilitation.

INTRODUCTION

Treadmill-based gait training is one clinical intervention used to promote walking function for individuals with cerebral palsy, osteoarthritis, or after stroke that allows users to practice many repetitions of a cyclic motion in a controlled environment and is relatively cost effective (Damiano and DeJong, 2009; Dickstein, 2008; Segal et al., 2015). Since increased walking speed directly corresponds to improved quality of life for community-dwelling older adults, it is often a key outcome of rehabilitation (Abellan van Kan et al., 2009; Bohannon et al., 1991; Dobkin et al., 2010) especially for individuals after stroke who may walk as slow as 0.5 m/s, classifying them as limited community ambulators (Patterson et al., 2007; Perry et al., 1990). However stroke survivors able to walk before participating in rehabilitation have not achieved clinically meaningful increases in walking speed after treadmill-based gait training (Hsu et al., 2003; Mehrholz et al., 2017; Richards et al., 1999; Turnbull et al., 1995) and there is currently no consensus on the best combination of therapies to improve walking function for stroke survivors (Dickstein, 2008).

Since stroke can cause both motor and cognitive deficits (Thaut et al., 1997), studies on dual task walking (DTW) have examined the paired effects of cognitive and locomotor performance in rehabilitation environments. In DTW studies, individuals typically walk in a controlled environment while completing cognitive tasks such as counting backward from a given number in increments of 2 or listing as many words as possible that begin with the same letter (Cossette et al., 2014). Clinical tasks are recommended to be as realistic and complex as possible to maximize the cognitive load as well as carry-over effects and patient sensitivity to training (Alderman et al., 2003; Cossette et al., 2014; McFadyen et al., 2017; Vallée et al., 2006).

Overground DTW results in clinically meaningful improvements in walking speed and overall function for individuals with neurological injury (McFadyen et al., 2017). Preliminary studies show DTW combined with user-driven treadmill control has benefits comparable to overground DTW (Fung et al., 2006; Rábago and Wilken, 2011), which suggests including user-driven treadmill walking in gait training programs may promote improved walking function.

Implementing user-driven treadmill control may be an effective means of enhancing motor learning and soliciting improved functional performance. High-intensity and repetitive task-specific practice are leading strategies for stroke rehabilitation (Langhorne et al., 2009), but traditional gait training environments use fixed speed treadmill walking and passive training strategies, which operate independent of user input. Passive treadmill training promotes increased interlimb symmetry in spatio-temporal parameters compared to overground walking (Harris-Love et al., 2001) but limits stride-to-stride variability which is critical to motor learning after stroke (Stergiou and Decker, 2011). Therefore, implementing active training, which requires and responds to user input, through user-driven treadmill control may provide a more beneficial and realistic environment for poststroke gait training.

Active training in the form of user-driven treadmill control can increase cortical reorganization and motor learning for stroke survivors (Wagner et al., 2012) as well as increase cortical activity in healthy adults (Bulea et al., 2014). In addition, user-driven treadmill walking promotes interactive participation, enhances natural mobility, and allows users to respond instantaneously to small gait disturbances that require volitional control (Yogev-Seligmann et al., 2008; Yoon et al., 2014). User-driven treadmill walking also allows greater stride-to-stride variability than fixed speed treadmill walking which is crucial for motor learning (Sloot et al., 2014). These findings suggest user-driven treadmill control would be a beneficial addition to poststroke gait training. During user-driven treadmill walking with various control schemes, healthy adults have walked at speeds similar to their overground walking speeds without significant changes to their stride length and width, joint kinematics, moments and powers (Plotnik et al., 2007; Sloot et al., 2014). However, previous studies utilized user-driven treadmill control schemes based primarily on user position and spatiotemporal parameters without using inertial control which responds directly to measures of forward propulsion, such as anterior ground reaction forces.

As an alternative to fixed speed treadmill training, user-driven treadmill speed control adjusts the speed of the treadmill belts in unison to match the user's instantaneous walking speed. The objective of this study was to determine the effect of this user-driven treadmill controller on the users' ability to actively select their walking speeds, generate propulsive forces, and modulate step length. We hypothesized that users would select similar walking speeds during overground, fixed speed and user-driven treadmill walking. At consistent speeds, we expected that users would generate similar propulsive forces using similar mechanics. By quantifying the effect of the user-driven treadmill control on the healthy users' gait mechanics, we will determine the potential for user-driven treadmill control in clinical poststroke gait training programs.

METHODS

Data Collection

Twenty-five healthy adults with no history of lower limb musculoskeletal injury participated in this study, but two participants were excluded from the analysis since they were more than thirty years older than the other participants. Only the data from the twenty-three young, healthy adults were used for this analysis (22.9 ± 4.04 years, 1.72 ± 0.11 m, 69.6 ± 10.9 kg, 9 male 14 female,

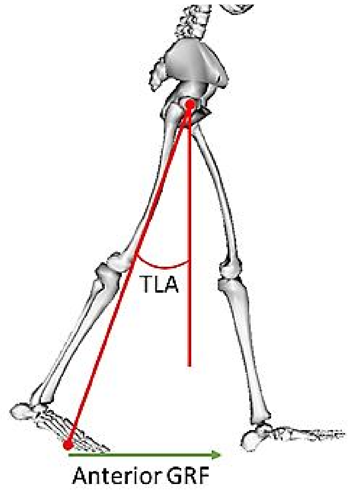

Figure 1: Definition of AGRF and TLA

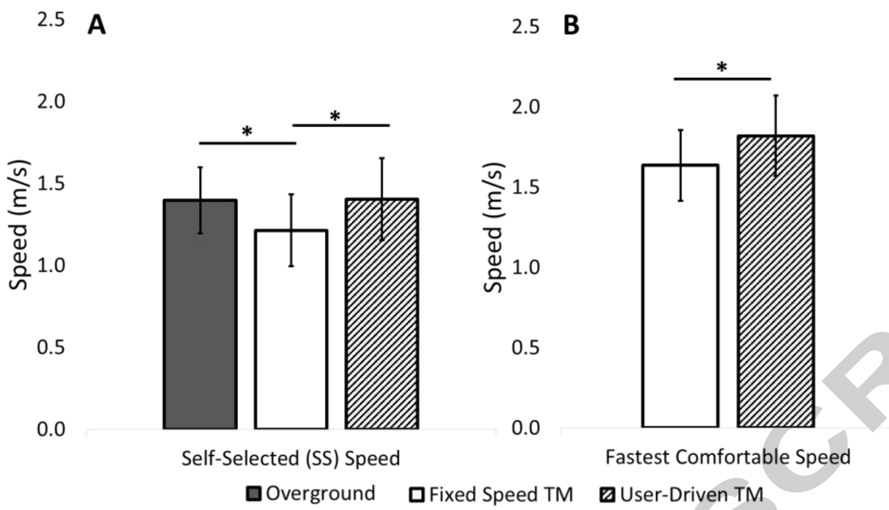

Figure 2: Comparison of walking speeds during the A) SS speed (n=23) and B) fastest comfortable speed (n=17) trials

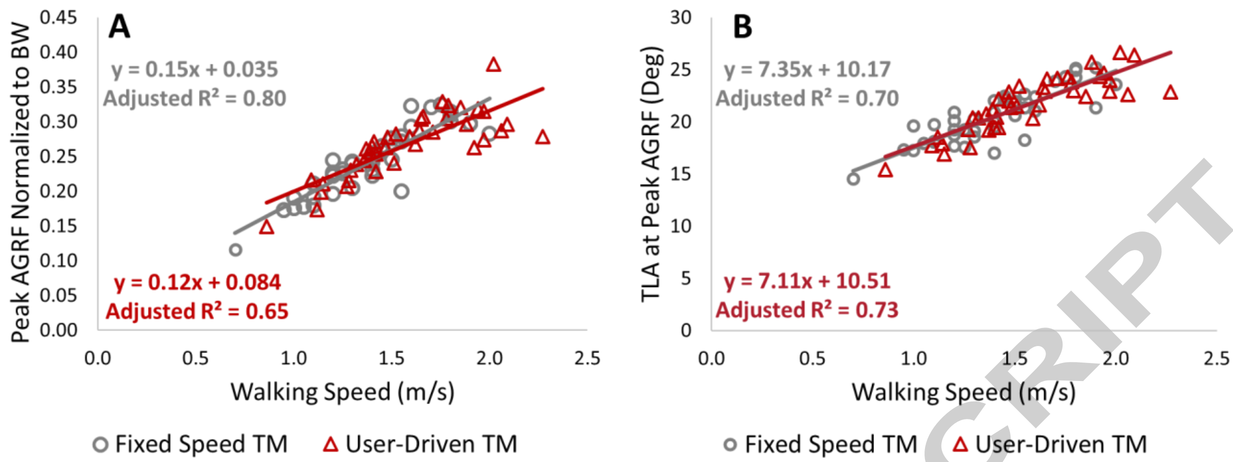

Figure 3: Relationship between walking speed and A) peak AGRF normalized to the subject's body weight (BW) and B) TLA at the moment of peak AGRF.

*Note: SS and Fast speed trials were combined to yield a continuum of walking speeds on the x-axis.).

The experimental protocol was approved by the University of Delaware Institutional Review Board, and each participant completed a written consent form and a modified physical activity readiness questionnaire (PAR-Q) before beginning the study.

This study was conducted in the Neuromuscular Biomechanics Laboratory at the University of Delaware. Participants were outfitted with 42 retroreflective markers: 26 single markers to define anatomical landmarks and 16 markers on rigid plastic shells to track the motion of their lower legs. Participants first completed a 10-meter overground walking task to measure their preferred overground self-selected (SS) walking speeds. Then, the treadmill based walking trials were performed in a random order on an instrumented, split-belt treadmill (Bertec Corp., Worthington, OH, USA) while the participants' motion was tracked with an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA, USA). All participants walked at their SS walking speeds on the treadmill in both its fixed speed and user-driven control modes. 17 (8 male, 9 female) of these participants also walked on the treadmill in both modes at their fastest comfortable walking speeds. Data were collected from the first group of subjects (n=8) to determine the study's feasibility and begin to examine the response of participants to the UDTM controller at SS speeds. Then, the additional 17 subjects were enrolled and the protocol was amended to compare the response of users at both SS and fast walking speeds.

To determine the users' preferred walking speeds on the fixed speed treadmill, researchers set the treadmill speed at the participants' SS speed from the 10-meter overground walking tasks. The speed was then increased or decreased in 0.05m/s increments according to the users' preferences. Once the preferred speed was set, data were collected for 1 minute. Kinematic data were sampled at 100Hz and kinetic data were sampled at 2000Hz. When using the user-driven treadmill control, participants were given up to 10 minutes to familiarize themselves with how the controller adjusted the belt speeds. For each trial, the participants took up to 1 minute to reach their chosen, steady state walking speed, and then data were collected for 1 minute at that steady state speed.

User-Driven Treadmill Controller

The user-driven treadmill controller used for this study combined inertial-force based control, gait parameter based control, and position based control. The speed of the treadmill belts were changed in response to the users' anterior ground reaction force (AGRF) impulse, step length, step duration, and position relative to the center of the treadmill. If individuals increased their push-off forces or moved near the front of the treadmill, the belt speeds increased. Conversely, if they decreased their push-off forces of moved near the back of the treadmill, the belt speeds decreased.

User-Driven Speed Calculation Algorithm

Each series of calculations was performed simultaneously for the two limbs. Sample calculations are shown for the right belt only.

1. Using analog data from the force plates, ground reaction forces and center of pressure (CoP) indicated whether a foot was in contact with the ground (Table 2).
    a. If the foot was in contact with the ground (i.e. stance phase), step duration (Eq. 1) and step length were calculated (Eq. 2). The step duration was calculated as the number of frames elapsed during the current step divided by the frame rate from the motion capture system (100 Hz). Then, the step length was calculated as the change in the anterior/posterior position of the foot CoP minus the distance the treadmill belt travelled during stance phase.

$$t_{sta} = \frac{n_F}{R_F} \qquad \text{Eq. 1}$$

$$L_{step} = \Delta CoP_Y - \left(v_{avg} * t_{sta}\right) \qquad \text{Eq. 2}$$

    b. During the stance phase, the AGRF impulse of the stance limb was summed over successive frames of data (Eq. 3). Then, the summed impulse was divided by the user's body weight and added to the previous belt speed (Eq. 4).

$$I_{AGRF,f} = I_{AGRF,i} + \frac{\sum_{k=1}^{end} F_{AGR}}{f_{analog}} \qquad \text{Eq. 3}$$

$$v_{new,f} = v_{new,i} + \frac{I_{AGRF,f}}{w_{body}} \qquad \text{Eq. 4}$$

c. If the foot was not in contact with the ground (i.e., swing phase), swing duration and the average stance limb velocity relative to the treadmill belt (Eq. 5 & 6) were determined.

$$dt = \Delta t_{swing} \quad \text{Eq. 5}$$

$$v_{LegR} = \frac{L_{step,R}}{t_{sta,R}} \quad \text{Eq. 6}$$

2. Regardless of the gait phase, the position of the body's center of mass was estimated using center of pressure measurements (Eq. 7 & 8). The anterior/posterior location of the body's center of mass was calculated as the midpoint between the CoP of the two limbs in terminal stance and expressed relative to the treadmill origin.

$$l_{CoM} = \frac{1}{2}\left(CoP_{y,L} + CoP_{y,R}\right) \quad \text{Eq. 7}$$

$$p_{CoM} = 1.5*(l_{CoM} - c_{TM}) \quad \text{Eq. 8}$$

3. The calculated speeds were averaged between belts, and the values of β and τ were tuned to the preferences of a pool of three healthy individuals and one individual post-stroke. The constants β=0.5, τ=1.5, and α (Eq. 9 & 10) were then used to average and smooth the belt speeds and ensure the belt accelerations felt natural for all participants (Eq. 10).

$$\alpha = 1 - e^{-\frac{dt}{\tau}} \quad \text{Eq. 9}$$

$$v_{smooth} = v_{Belt} + \alpha * \left(\frac{v_{new,L} + v_{new,R}}{2} - v_{Belt}\right) \pm \beta * p_{CoM}^2 \quad \text{Eq. 10}$$

The parameters were tuned to ensure subject safety. Throughout the testing, the two belt speeds were tied and the maximum belt acceleration was set to 0.2 m/s^2.

Data Analysis

Kinematic data were processed using Cortex 6 (Motion Analysis Corp., Santa Rosa, CA, USA), and kinetic and kinematic calculations were performed using Visual 3D software (C-Motion Inc., Germantown, MD, USA). Kinetic and marker data were filtered at 30Hz and 6Hz, respectively. Preferred walking speeds were calculated as follows:

- 10-meter overground walking task: The average time to travel 6 meters over 3 trials was used to calculate the average preferred walking speed.
- Fixed speed treadmill control: The walking speed was read directly from the treadmill interface.
- User-driven treadmill control: The steady state walking speed was calculated as the average walking speed over the steady state portion of the 1 minute trial.

Since individuals may increase their forward propulsion via increased AGRF and increase their step length by increasing their trailing limb angle (TLA) (Awad et al., 2014; Hsiao et al., 2016, 2015), the primary measures in this analysis were walking speed, AGRF, and TLA. TLA is defined as the angle between a straight line connecting the calculated hip joint center and the 5th metatarsal of the trailing limb and the vertical axis of the lab (Figure 1).

A paired t-test blocked by subject (α=0.05) was used to determine if participants selected significantly different SS and fast walking speeds when using fixed speed and user-driven treadmill control. Two subsequent paired t-tests blocked by subject (α=0.05) were used to determine if the participants' peak AGRF and TLA at the instant of peak AGRF varied between the fixed speed and user-driven conditions.

Since increased AGRF and TLA can lead to increased walking speeds, it is important to determine if differences in the participants' gait mechanics are due to changes in walking speed between the fixed speed and user-driven conditions or the treadmill controller being used. Therefore, an analysis of covariance (ANCOVA) was used to quantify the portion of any change in AGRF or TLA that is due to changes in walking speed versus changes in the treadmill controller.

RESULTS

Participants chose significantly slower SS walking speeds on the fixed speed treadmill compared to overground walking during the 10-meter task ($p<0.05$, Figure 2). Participants then selected significantly faster SS and fast walking speeds on the user-driven treadmill compared to the fixed speed treadmill ($p<0.05$). There was no significant difference between the participants' SS walking speeds in the overground and user-driven treadmill trials ($p>0.05$).

There were also significant differences in the participants' gait mechanics during fixed speed and user-driven treadmill walking. Participants had higher peak AGRF and greater TLA at the instant of peak AGRF when walking on the user-driven treadmill compared to the fixed speed treadmill ($p>0.05$, Figure 3). Linear models were fit to each data set and an analysis of covariance (ANCOVA, $\alpha=0.05$) was used to determine the relative contribution of changes in walking speed and treadmill control to changes in AGRF and TLA. Results of the ANOVA revealed that the changes in peak AGRF and TLA were primarily due to changes in walking speed and not changes in the treadmill control mode (**Table 3**). The results of the ANCOVA indicate that 100% of the changes in peak AGRF between the fixed speed and user-driven treadmill trials was explained by the difference in the participants' walking speeds. Likewise, approximately 97% of the change in TLA at the instant of peak AGRF between the two treadmill control conditions was due to the differences in walking speeds selected for the SS and fast walking trials.

DISCUSSION

In this study, participants selected faster walking speeds with the user-driven treadmill controller for both the SS and fast speed trials when compared to the fixed speed treadmill. This higher user-driven SS speed was similar to the average overground walking speed. The peak AGRF and TLA at the moment of peak AGRF also increased with the higher walking speeds on the user-driven treadmill. However, our results showed the increases in AGRF and TLA were primarily due to the increased walking speeds on the user-driven treadmill rather than different controller modes.

Since participants selected SS speeds similar to their overground SS speed while on the user-driven treadmill, user-driven treadmill walking may facilitate improved efficacy of locomotor training. In a study of typically developing children and those with cerebral palsy walking on a treadmill in its fixed speed and user-driven modes, participants chose walking speeds 7.3% faster in the user-driven mode (Sloot et al., 2015). In this study, participants walked 5.5% and 11.3% faster on the user-driven treadmill during the SS and fast walking trials respectively. This agreement suggests individuals with neurological injuries will see similar benefits using the proposed user-driven controller as found in literature. In addition, our results suggest treadmill users employ similar propulsive mechanics on the fixed speed and user-driven modes, which agrees with previous studies (Kim et al., 2012; Yoon et al., 2012) comparing fixed speed treadmill control with user-driven control incorporating inertial control elements and support the feasibility of using the user-driven control in a rehabilitation environment.

The strengths of this study lie in our isolation of the effect of the user-driven treadmill controller. Since both treadmill control schemes were implemented on the same treadmill, the methodology controls for differences in mechanical system. The participants were all young, healthy, and neurologically intact. Participants were given ample time to practice walking on the treadmill in the user-driven mode and data were only collected once the user had reached a steady state walking speed. To eliminate any effects of learning between trials, the order of walking trials was random and participants walked overground between any two trials.

This experimental setup includes several precautions to ensure participant safety. All participants wear a fall harness while walking on the treadmill and the treadmill control algorithm was designed so that crossover and missteps are ignored. In addition, the maximum acceleration and speed of the treadmill belts are limited. The tuning parameters (α, β, and τ) that weight the contribution of changes in treadmill speed due to push-off forces, step length and position relative to the center of the treadmill can be adjusted according to user preference.

This study is limited by its small sample size. However, the differences in walking speed maintained sufficient statistical power for 90% confidence. Due to the homogeneity of the participant pool and the high level of significance of speed differences and the strong positive correlations between walking speed and AGRF and TLA, we expect these trends to hold for larger samples as well. Although R^2 values decreased for faster walking speeds, the comparisons maintained strong correlations for all walking speeds. We expect this decrease to signal participants beginning to transition from walking to running and will further explore this transition region in future work.

In this study, an instrumented split-belt treadmill and motion capture system were used for the user-driven treadmill control. Other researchers have implemented user-driven control using a variety of ultrasonic (Minetti et al., 2003), depth (Kim et al., 2013), and force sensors (Christensen et al., 2000; Koenig et al., 2009) to lower costs and improve accessibility to user-driven treadmill control, and now, even standard exercise equipment offers user-driven treadmill control. However, the unique combination of inertial-force, position, and gait parameter control used in this study is suited for implementation in a poststroke gait training regimen because it allows users to increase their walking speeds smoothly and naturally. Since the controller responds directly to changes in forward propulsion and step length, it can respond to changes in walking speed before the changes can be measured at a whole-body level.

Our findings show the user-driven treadmill controller allowed the study participants to select walking speeds faster than their chosen speeds on the fixed speed treadmill and similar to their overground walking speeds. Previous studies found participants walked significantly slower on fixed speed treadmills compared to overground walking, but the overground exercise required less energy expenditure and resulted in lower perceptual and greater positive affective responses (Dasilva et al., 2011). Since user-driven treadmill walking increases cognitive activity and natural mobility (Bulea et al., 2014; Yogev-Seligmann et al., 2008; Yoon et al., 2014), it is reasonable to expect our participants would walk faster on the user-driven treadmill than the fixed speed treadmill, and these faster speeds would be similar to those from overground walking. In addition, the corresponding increases in peak AGRF and TLA at the time of peak AGRF were primarily due to the increased walking speeds on the user-driven treadmill, not the treadmill controller itself. Therefore, when this user-driven treadmill controller is implemented in a poststroke gait training program, we anticipate stroke survivors will select walking speeds similar to their overground walking speeds and expect similar benefits as well. This should increase the efficacy of the gait training program (Sullivan et al., 2002) and thereby, improve the patients' functional ability and subsequently their quality of life after rehabilitation.

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Conflict of Interest Statement

The authors have no conflicts of interest to report.

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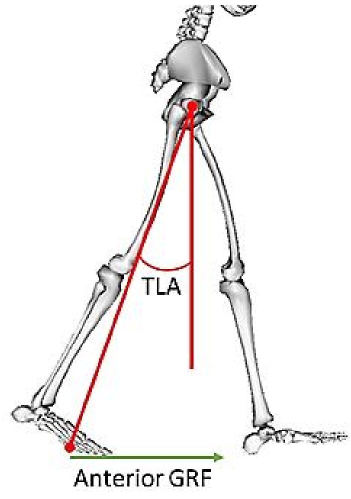

Figure 1: Definition of AGRF and TLA

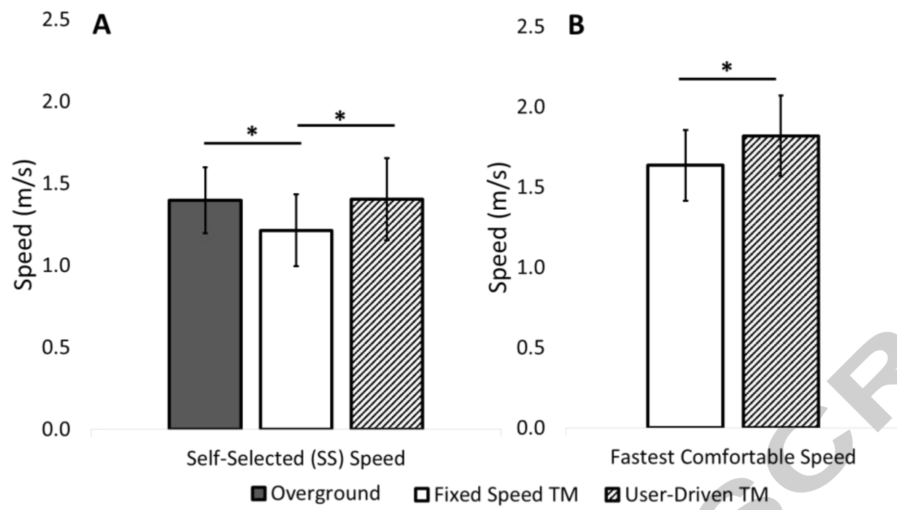

Figure 2: Comparison of walking speeds during the A) SS speed (n=23) and B) fastest comfortable speed (n=17) trials

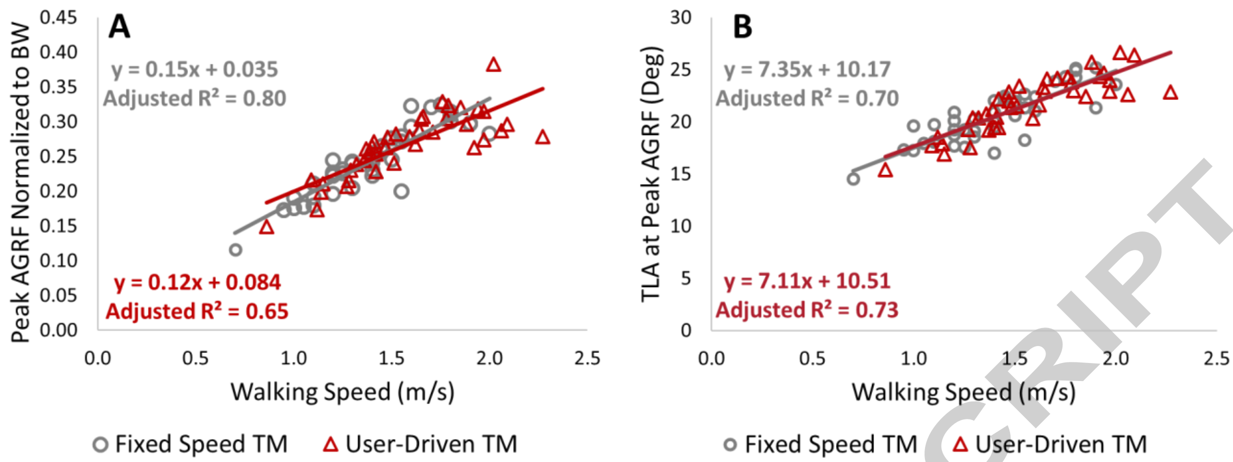

Figure 3: Relationship between walking speed and A) peak AGRF normalized to the subject's body weight (BW) and B) TLA at the moment of peak AGRF.

**Note: SS and Fast speed trials were combined to yield a continuum of walking speeds on the x-axis.*

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Table 1: Subject information and walking speeds for the 23 participants.

Note: WS = walking speed, TM = treadmill, SS = self-selected.

Subject No.	Height (m)	Mass (kg)	Age (years)	Gender	Overground SS Speed (m/s)	Fixed Speed TM Control		User-Driven TM Control	
						SS Speed (m/s)	Fast Speed (m/s)	SS Avg Speed (m/s)	Fast Avg Speed (m/s)
1	1.70	68.36	20	F	1.42	1.40	--	1.59	--
2	1.63	70.93	25	F	1.30	1.30	--	1.42	--
3	1.63	54.34	21	F	1.44	1.40	--	1.42	--
4	1.68	56.49	23	F	1.35	1.25	--	1.40	--
5	1.85	77.51	23	M	1.60	1.50	--	1.94	--
6	1.60	72.77	19	F	1.18	0.95	--	1.29	--
7	1.93	91.39	21	M	1.47	1.40	1.80	1.71	2.09
8	1.83	79.93	27	M	1.40	1.30	--	1.15	--
9	1.75	73.15	19	M	1.44	1.20	1.60	1.36	1.85
10	1.88	71.70	19	M	1.50	1.30	1.80	1.51	1.92
11	1.83	72.33	19	M	1.11	1.00	1.90	1.09	2.06
12	1.60	71.98	28	F	1.61	1.20	1.75	1.48	1.76
13	1.78	71.88	20	M	1.34	1.10	1.40	1.12	1.62
14	1.60	69.76	22	F	1.25	1.20	1.40	1.14	1.24
15	1.75	60.92	32	F	1.58	1.10	1.45	1.39	1.66
16	1.52	44.89	20	F	1.15	0.70	1.20	0.86	1.64
17	1.68	66.24	22	F	1.77	1.55	2.00	1.97	2.27
18	1.57	63.72	22	F	1.56	1.25	1.90	1.52	1.88
19	1.83	85.11	24	M	1.16	1.00	1.50	1.47	1.97
20	1.80	75.45	34	F	1.22	1.10	1.70	1.32	2.02
21	1.78	70.55	21	M	1.47	1.55	1.25	1.42	1.79
22	1.57	53.55	20	F	1.17	1.05	1.45	1.27	1.47
23	1.68	65.84	20	F	1.20	1.20	1.60	1.37	1.65
Average	**1.72**	**69.65**	**22.91**	**9 M**	**1.38**	**1.22**	**1.61**	**1.40**	**1.81**
Standard Deviation	**0.11**	**10.92**	**4.0**	**14 F**	**0.18**	**0.21**	**0.24**	**0.25**	**0.26**

Table 2: Definitions of variables used in speed calculations for the user-driven control

t_{sta}	Stance time for leg of interest
N_F	Number of frames elapsed during step
R_F	Camera frame rate
L_{step}	Step length for leg of interest
ΔCoP_y	Anterior/posterior distance between left and right CoP
I_{AGRF}	Impulse of the anterior ground reaction force
F_{AGR}	Anterior ground reaction force
f_{analog}	Analog data sampling frequency
v_{new}	Intermediate belt velocity that incorporates the AGRF impulse
w_{body}	Body weight in kilograms
v_{avg}	Average treadmill speed from previous iteration
V_{Leg}	Expected speed for the treadmill belt based on motion of the corresponding limb
P_{CoM}	Anterior/Posterior position of user CoM relative to center of treadmill
l_{CoM}	Anterior/Posterior location of user center of mass (CoM) relative to lab
c_{TM}	Center of treadmill belt
dt	Change in time
Δt_{swing}	Time elapsed during swing phase
v_{smooth}	Smoothed treadmill belt speed
α	Smoothing constant 1
β	Smoothing constant 2
τ	Time constant of the speed change

Table 3: Summary of ANCOVA results at SS and Fast walking speeds from Figure 3 data

Walking Speed	Dependent Variable	Independent Variable	Percent Change in Dependent Variable Explained	Model Adjusted R^2
SS Speed (n=23)	Peak AGRF Normalized to Body Weight	SS Speed (m/s)	100%	0.76
		TM Controller	0%	
	TLA (Deg)	SS Speed (m/s)	97%	0.65
		TM Controller	3%	
Fast Speed (n=17)	Peak AGRF Normalized to Body Weight	Fast Speed (m/s)	100%	0.31
		TM Controller	0%	
	TLA (Deg)	Fast Speed (m/s)	98%	0.46
		TM Controller	2%	