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Technical note

The accuracy of rapid treadmill-belt movements as a means to deliver standing postural perturbations

Jeremy R. Crenshaw[a,b], Kathie A. Bernhardt[a], Emma Fortune[a,c], Kenton R. Kaufman[a,*]

[a] Department of Orthopedic Surgery, Mayo Clinic, Rochester, MN, USA
[b] Department of Kinesiology and Applied Physiology, University of Delaware, Newark, DE, USA
[c] Department of Health Sciences Research, Mayo Clinic, Rochester, MN, USA

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ABSTRACT

Treadmill-induced postural perturbations are a promising tool in assessing and reducing the risk of falls. We evaluated the accuracy with which two treadmills (Simbex ActiveStep® and an AMTI instrumented treadmill) achieved commanded displacements, peak velocities, and average initial accelerations. To do so, we included a range of perturbation magnitudes (20, 30, and 40 cm displacements) applied in unweighted and weighted (body mass = 46–84 kg) conditions. Across treadmills and perturbation magnitudes, absolute errors in displacement (< 0.5 cm) and peak velocity (< 4 cm/s) were small (relative error < 5%). Between-treadmill differences in displacement and peak velocity were marginal (< 3%), regardless of the perturbation magnitude and participant body mass. Observed accelerations were more than 5% smaller than commanded values. The front, but not back, AMTI belt demonstrated less acceleration accuracy than the ActiveStep® (≈ 5% difference). In summary, both treadmills demonstrated a reasonable, consistent level of accuracy in delivering postural perturbations.

1. Introduction

In previous studies, rapid treadmill-belt translations have been used to induce falls in standing or walking participants. Such studies have shed light on the factors that influence fall-recovery success [1–3], as well as the effects of age [4], obesity [5], or neuromuscular impairment [6] on fall recovery. In turn, treadmill-delivered perturbations have led to objective assessments of fall recovery [7], as well as interventions to reduce falls and enable mobility [8–16]. In order to support the validity and reliability of these approaches, the accuracy and precision of such treadmill-belt perturbations must be quantified.

One device commonly used to induce falls is the ActiveStep® treadmill (Simbex, Lebanon, NH) [1,2,13,14,4–7,9–12], a commercially available product marketed as a “mobility simulator that prevents falls” (activestep.simbex.com). The ActiveStep® is capable of delivering rapid belt accelerations during walking or standing, triggering fall-recovery responses that simulate the response to a trip [3,17] or slip [18]. Such simulations are relevant, as the majority of falls in older adults are due to these common causes [19,20]. Despite the promising applications of the ActiveStep® device, its accuracy and precision in delivering rapid belt translations have not been reported.

Although it is equipped with a pressure mat, a limitation of the ActiveStep® is that it does not contain force plates, and therefore is limited to monitoring only vertical forces. In other studies, translatable force platforms have been used to induce falls. In turn, shear ground reaction forces [21] and lower-extremity joint moments [22] can be quantified to evaluate the force-generating aspects of the response. Such capabilities may enable researchers to identify underlying mechanisms of impaired balance reactions. Translating platforms, however, have limited displacement and acceleration capabilities compared to that of the ActiveStep®. Some force-sensing treadmills, although not marketed as tools to induce falls, have acceleration capabilities that rival that of the ActiveStep®. If such a treadmill accurately and precisely delivers rapid belt translations while supporting a human participant, then researchers can evaluate the kinetic aspects of the recovery response. To date, the accuracy and precision of perturbations delivered from an instrumented treadmill have not been evaluated.

The purpose of this study was to characterize and compare the accuracy of two treadmills in delivering rapid belt translations to induce falls in human participants. We compared the ActiveStep® to an instrumented treadmill (AMTI, Watertown, MA). We anticipated that, given the disparity in intended applications, the ActiveStep® would have better accuracy. We evaluated different

* Corresponding author. Present address: Motion Analysis Laboratory, Division of Orthopedic Research, Dan Abraham Healthy Living Center 4-214A, Mayo Clinic, 200 First Street SW, Rochester, MN 55905, USA.

E-mail addresses: kaufman.kenton@mayo.edu, borglum.barbara@mayo.edu (K.R. Kaufman).

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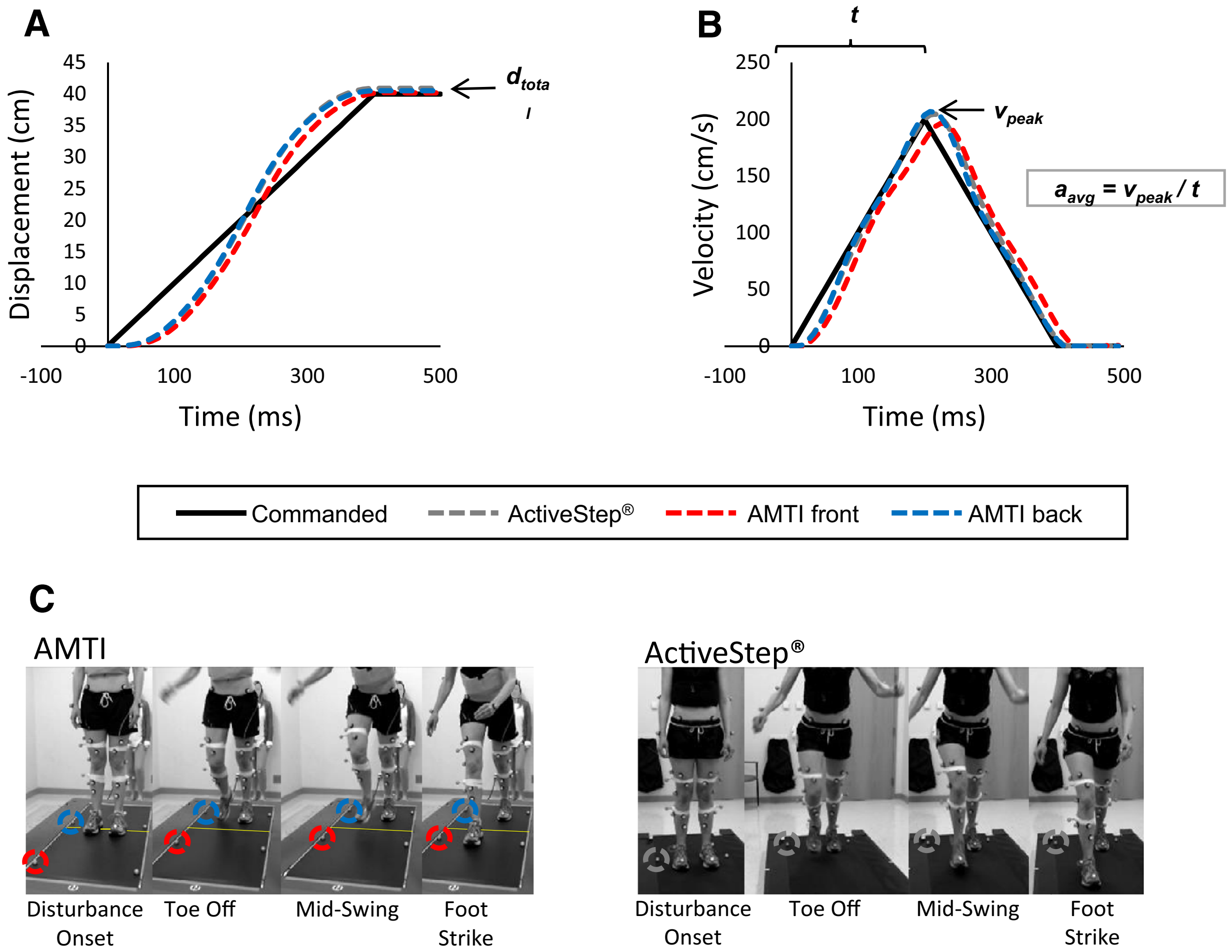

Fig. 1. Observed belt kinematics from a 40 cm translation applied to a 29 year old participant (height = 1.78 m, body mass = 65 kg). **A.** Commanded (black solid line) belt displacements over time compared to observed, filtered displacements from markers on the ActiveStep® (gray dashed line), AMTI front belt (red dashed line), and AMTI back belt (blue dashed line). The ActiveStep® belt trajectory (gray) is difficult to see due to its similarity with the back AMTI belt (blue). Total displacement (d_{total}), or the absolute change in position from disturbance onset to final position, is noted. **B.** Commanded (black solid line) belt velocities over time compared to observed velocities from the ActiveStep® and AMTI belts. Peak velocity (v_{peak}), as well as the time of peak velocity (t), are noted. The average initial acceleration (a_{avg}) was determined by dividing peak velocity by its timing. **C.** The participant responds to this disturbance on each treadmill. The marker of interest is noted by dashed circles in gray (ActiveStep®), red (AMTI front), or blue (AMTI back). The junction of the AMTI treadmill is highlighted using a yellow line.

perturbation magnitudes and body masses, as we anticipated that these factors would influence our outcomes.

2. Methods

2.1. Instrumentation

The ActiveStep® device was controlled by computer using its accompanying software. In this program, the user commanded the target velocities and times to reach such velocities within distinct perturbation phases. From these inputs, the commanded total displacement, peak velocity, and average acceleration were calculated.

The AMTI treadmill was controlled by computer using custom LabVIEW software (National Instruments, Austin, TX) that implemented commercially available virtual instruments to control the treadmill motors (Copley Controls, Canton, MA). In this custom software, the user commanded the targeted velocities and durations of each perturbation phase.

2.2. Participants

Five participants with no self-reported neuromuscular impairments provided informed consent for this study. Participants were 27–35 years of age (mean (SD) = 30.0 (3.2) years). Height (1.73 (0.12) m, range = 1.53–1.87 m) and body mass (63.8 (14.0 kg), range = 46.5–84.1 kg) were measured at both visits, with less than a 1.5% within-participant change in each measure observed between visits. This protocol was approved by the Mayo Clinic Institutional Review Board.

2.3. Protocol

Participants visited the laboratory for two sessions, 5 to 10 days apart. The two treadmills were assessed in the same motion analysis laboratory, but were located in separate, similarly sized motion capture volumes. On the first visit, the AMTI treadmill was assessed. On the second visit, the ActiveStep® was assessed. Each visit involved 30 perturbations. Each perturbation was comprised of 200 ms acceleration and deceleration phases (Fig. 1B), as delivered in our previous studies [7]. Anterior and posterior perturbations were alternated within blocks of 10 perturbations, first starting with 10 small, 20 cm perturbations, followed by blocks of 30 cm and 40 cm displacements (Table 1). Here, the direction of the perturbation refers to the translation of the treadmill belt, not the resulting fall direction of standing individuals. On the ActiveStep®, participants stood with feet placed halfway between the front and back of the treadmill belt. On the AMTI treadmill, participants

Table 1
Accuracy and precision of treadmill belt movements in unweighted conditions.

Outcome	Treadmill	Commanded Magnitude	Accuracy	Precision			
			Mean Difference	S.D.	Min	Max	C.V. (%)
Displacement (cm)	ActiveStep®	20	0.3	<0.1	0.2	0.3	<1
		30	0.4	0.1	0.3	0.5	<1
		40	0.5	0.1	0.4	0.7	<1
	AMTI (front)	20	0.2	<0.1	0.2	0.2	<1
		30	0.2	<0.1	0.2	0.2	<1
		40	0.3	<0.1	0.3	0.3	<1
	AMTI (back)	20	0.2	<0.1	0.2	0.2	<1
		30	0.2	<0.1	0.2	0.2	<1
		40	0.3	<0.1	0.3	0.3	<1
Peak Velocity (cm/s)	ActiveStep®	100	−0.6	0.1	−0.7	−0.4	<1
		150	−1.2	0.4	−1.6	−0.8	<1
		200	−1.5	0.2	−1.9	−1.3	<1
	AMTI (front)	100	−2.1	1.2	−3.4	−0.9	1
		150	−3.4	2.0	−5.3	−1.4	1
		200	−2.5	0.5	−3.0	−1.8	<1
	AMTI (back)	100	−2.0	1.2	−3.3	−0.8	1
		150	−3.5	1.7	−5.1	−1.9	1
		200	−2.7	0.2	−2.9	−2.5	<1
Average Initial Acceleration (cm/s^2)	ActiveStep®	500	−69.1	7.4	−74.1	−58.2	2
		750	−107.6	10.7	−114.1	−87.4	2
		1000	−149.4	0.9	−151.1	−148.3	<1
	AMTI (front)	500	−78.9	12.5	−100.2	−59.4	3
		750	−116.8	21.0	−151.0	−89.5	3
		1000	−141.0	17.0	−155.8	−119.6	2
	AMTI (back)	500	−78.4	12.2	−99.7	−59.2	3
		750	−122.1	15.2	−150.6	−92.3	2
		1000	−145.1	15.2	−155.4	−122.4	2

Note: 10 trials were evaluated for each commanded magnitude. Mean difference is observed – commanded values. S.D. is standard deviation across all trials. C.V. is the coefficient of variation.

stood on the junction of the front and back belts (Fig. 1C). Participants wore a safety harness attached to an overhead rail, and were instructed to "try not to fall". On each belt, two 25 mm reflective markers were placed near the lateral edges of the belt. The trajectories of these markers were recorded using 8 motion capture cameras (120 Hz; Motion Analysis Corporation, Santa Rosa, CA). The motion of one marker on each belt was analyzed for each trial.

2.4. Data analysis

Three-dimensional marker trajectories were low-pass filtered at 10 Hz (4th order recursive Butterworth filter). Marker velocity was calculated by differentiating the three-dimensional trajectories. Perturbation onset was defined as the time at which the anteroposterior speed of the belt exceeded 2.5 standard deviations from the mean of a 500 ms stationary period. This threshold was determined by trial and error as the minimum threshold that agreed with visually identified onsets. Lower thresholds were prone to gross misidentification of onsets due to small-scale, predisturbance fluctuations in the velocity signal. From this point, the *total absolute displacement* of the marker was calculated from the resultant change in position (Fig. 1A). The *peak velocity* was identified (Fig. 1B), as well as the *time of the peak velocity after perturbation onset*. The *average absolute initial acceleration* of the perturbation was calculated by dividing the peak velocity by this time (Fig. 1B). Data were analyzed using custom software (MATLAB®, Mathworks®, Natick, MA).

Dependent variables were the relative errors (δ = (observed – commanded) / commanded) in *total absolute displacement, peak velocity*, and *initial average acceleration*. We chose to analyze errors in these discrete variables, as opposed to the error throughout the entire time-series, because such discrete variables are commonly used to describe perturbation characteristics [1,2,3–14,].

For each dependent variable, separate general linear models (SPSS v24, IBM, Armonk, NY, "UNIANOVA" command) were created to evaluate differences between treadmill belts, including the ActiveStep®, AMTI front belt, and AMTI back belt. In these full-factorial models, the perturbation direction was also included as an independent measure. Body mass and the commanded perturbation parameter were included as continuous-measure covariates. For example, the model assessing peak velocity error included the independent variables of the treadmill belt (ActiveStep®, AMTI front belt, and AMTI back belt), perturbation direction (anterior or posterior), mass on the treadmill, commanded peak velocity, and interaction terms consisting of up to four of these variables. Both weighted and unweighted trials were included in these analyses. We assumed that our experimental manipulation of participant mass had substantially more influence on observed error than any characterized, active response from the individual. We also assumed that any between-belt relationships of error within the AMTI treadmill were small compared to the influence of body mass or perturbation magnitude. Under these assumptions, each trial was considered as an independent sample in our analyses. So, a total of 540 samples were considered (i.e. 5 trials × 6 wt conditions × 2 directions × 3 belts × 3 commanded perturbation magnitudes). In the case of significant interactions including mass, post-hoc comparisons of treadmills were evaluated using estimated marginal means at small (45 kg) and large (85 kg) masses. If no interactions including mass were significant, then estimated marginal means were evaluated at the mean body mass of 63.8 kg as to characterize relative error at mid-range mass values. In the case of significant interactions including commanded perturbation parameters, post-hoc comparisons of treadmills were evaluated at the smallest and largest commanded values (20 cm and 40 cm, or associated velocities and accelerations, Tables 1 and 2). If no interactions including commanded magnitudes were significant, then estimated marginal means were evaluated at perturbation parameters corresponding with a 30 cm translation. If significant interactions included direction, post-hoc comparisons between treadmill belts were made within each direction, and post-hoc comparisons

Table 2
Accuracy and precision of treadmill belt movements in weighted conditions.

Outcome	Treadmill	Commanded Magnitude	Accuracy	Precision			
			Mean Difference	S.D.	Min	Max	C.V. (%)
Displacement (cm)	ActiveStep®	20	0.3	0.1	0.2	0.4	<1
		30	0.5	0.2	0.1	0.9	<1
		40	0.5	0.3	−0.7	1.4	<1
	AMTI (front)	20	<0.1	0.1	−0.1	0.3	<1
		30	0.2	0.2	−0.1	0.7	<1
		40	0.1	0.2	−0.1	0.7	<1
	AMTI (back)	20	0.2	0.1	<0.1	0.5	<1
		30	0.2	0.2	<0.1	0.6	<1
		40	0.2	0.2	<0.1	0.6	<1
Peak Velocity (cm/s)	ActiveStep®	100	0.4	1.5	−3.0	3.4	<1
		150	2.9	1.9	−2.3	6.6	<1
		200	3.7	1.8	−0.9	7.2	<1
	AMTI (front)	100	−2.5	3.0	−10.0	2.9	3
		150	−2.5	4.7	−14.5	8.7	3
		200	0.2	4.0	−5.4	9.2	2
	AMTI (back)	100	0.9	1.6	−2.1	5.8	2
		150	0.6	2.6	−4.6	11.3	2
		200	1.2	4.2	−4.0	10.0	2
Average Initial Acceleration (cm/s^2)	ActiveStep®	500	−43.3	11.7	−69.0	−13.2	3
		750	−66.1	17.1	−102.1	−32.3	3
		1000	−98.9	16.9	−137.1	−45.4	2
	AMTI (front)	500	−62.1	19.9	−111.6	−25.2	5
		750	−107.1	33.8	−157.2	−27.9	5
		1000	−128.6	28.1	−164.3	−70.3	3
	AMTI (back)	500	−48.2	17.8	−87.3	−8.7	4
		750	−76.6	21.4	−126.6	−33.0	3
		1000	−91.9	33.2	−158.5	−30.8	4

Note: 50 trials were evaluated for each commanded magnitude (5 participant masses × 10 trials). Mean difference is observed – commanded values. S.D. is standard deviation across all trials. C.V. is the coefficient of variation.

within treadmill belts were made across directions. All post-hoc analyses were corrected using Bonferroni adjustments for each statistical model. Precision was characterized from the coefficient of variation (C.V.) of the relative error.

3. Results

3.1. Total displacement

For commanded 20, 30, and 40 cm displacements, average errors were less than 0.5 cm (< 2%, Tables 1 and 2). The general linear model explained just under half of the variance in displacement error ($R^2 = 0.47$, $p < 0.001$). A significant interaction between the treadmill belt and body mass was observed ($p = 0.03$). No other main effects or interactions were significant ($p = 0.24$ – 0.98). In order to explore the significant interaction, we compared treadmill belt estimated marginal means at masses of 45 kg and 85 kg, with a commanded displacement of 30 cm. In these tests, all treadmill belts demonstrated different error magnitudes ($p < 0.002$, Fig. 2). Across conditions, the ActiveStep® had the largest error (≈ 1.5%), and the front AMTI belt had the smallest error (≤ 0.5%).

3.2. Peak velocity

For commanded peak velocities of 100–200 cm/s, average errors were less than 4 cm/s (< 3%, Tables 1 and 2). The general linear model explained just under two-thirds of the variance in peak velocity error ($R^2 = 0.65$, $p < 0.001$). A significant interaction between the treadmill belt, mass, and the commanded velocity was observed ($p = 0.009$). A significant main effect of direction ($p = 0.006$) was also observed such that anterior perturbations were characterized by a positive shift in error (β (S.E.) = 4.0 (1.8)%). The in-

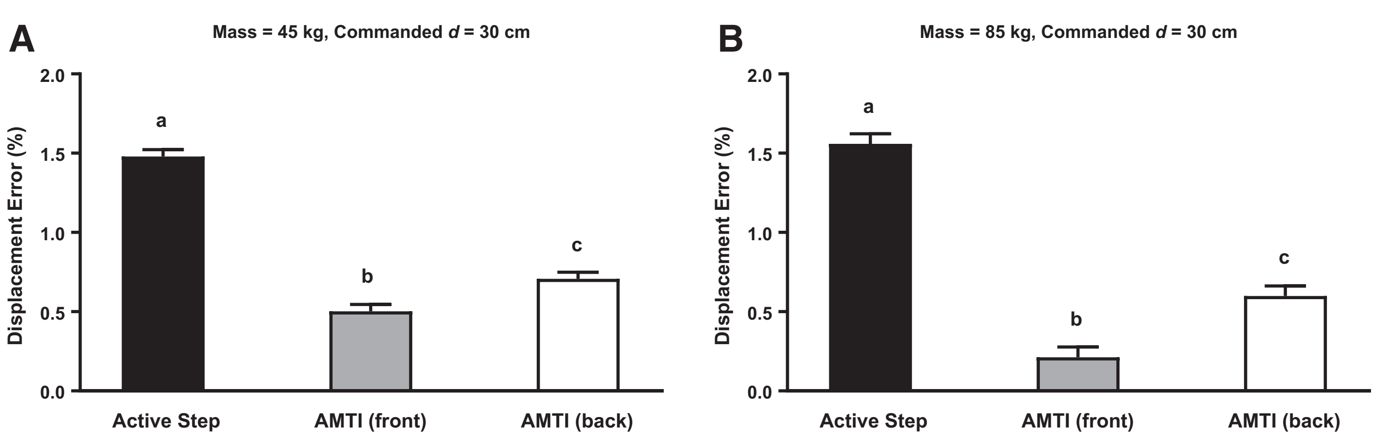

Fig. 2. Estimated marginal means (standard errors) of relative errors (observed – commanded / commanded) in total displacement, estimated at a commanded displacement of 30 cm and bearing masses of 45 kg (A) and 85 kg (B). Significant ($p < 0.01$) between-treadmill-belt differences are noted with different letter labels (a, b, c).

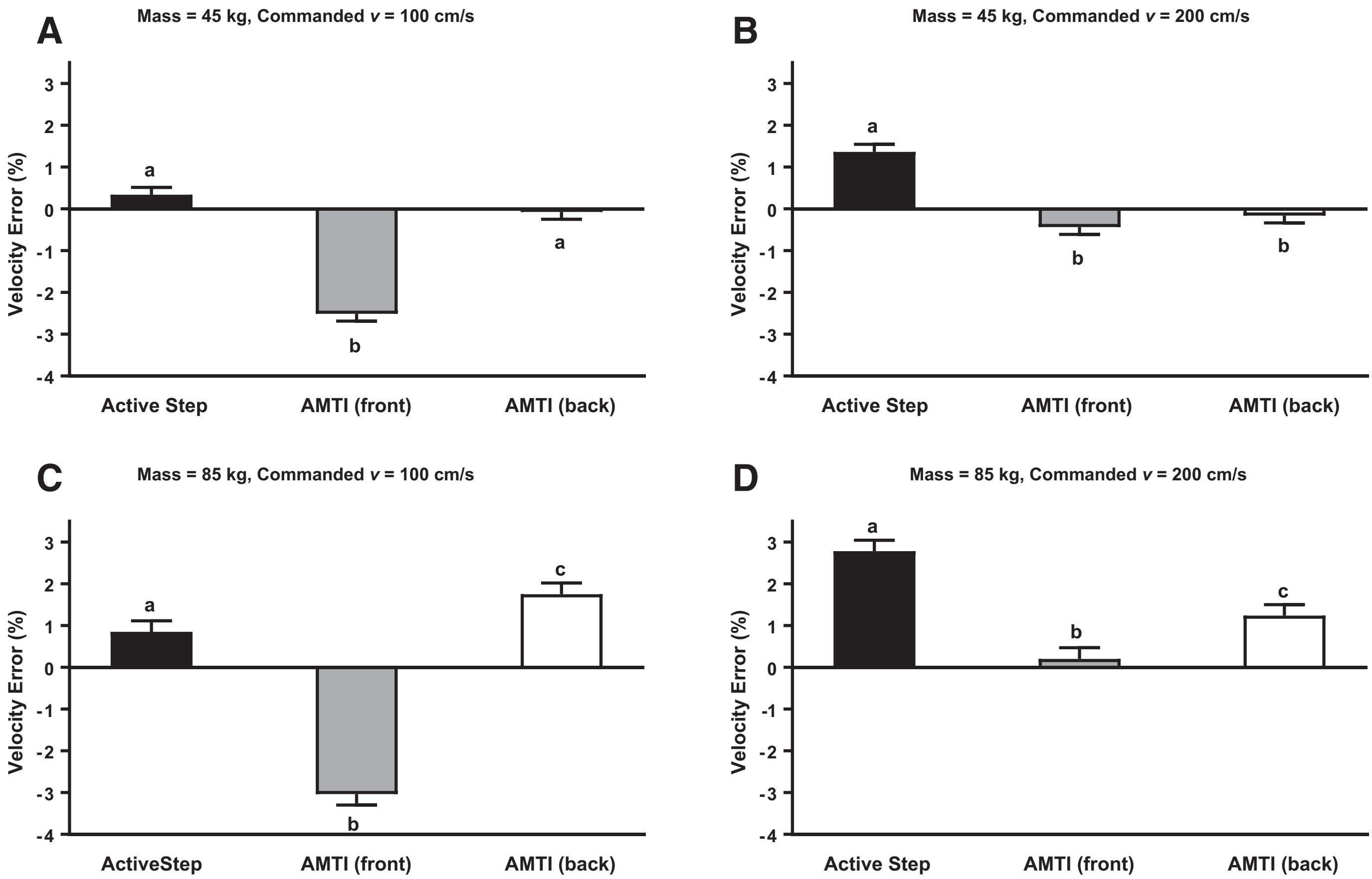

Fig. 3. Estimated marginal means (standard errors) of relative errors (observed – commanded / commanded) in peak velocity, estimated at commanded velocities of 100 cm/s (A &C) or 200 cm/s (B & C), as well as bearing masses of 45 kg (A & B) or 85 kg (C & D). Significant ($p < 0.05$) between-treadmill-belt differences are noted with different letter labels (a, b, c).

teraction of direction and commanded velocity approached significance ($p = 0.08$), slightly diminishing the direction main effect for larger perturbations. No other interactions containing direction were significant ($p > 0.37$). In order to explore the significant interaction of treadmill belt, mass, and commanded velocities, we compared treadmill estimated marginal means at combinations of mass (45 kg and 85 kg) and commanded velocities (100 cm/s and 200 cm/s) (Fig. 3). At the higher peak velocities, the AMTI belts had slightly better accuracy than that of the ActiveStep® ($p < 0.001$). At low commanded velocities, the ActiveStep® generally had slightly better accuracy than the AMTI belts ($p < 0.05$), a trend that was more pronounced when perturbing larger masses. At these slower velocities, the front AMTI belt had slightly worse accuracy than the back AMTI belt.

3.3. Average initial acceleration

For perturbations with commanded accelerations of 500–1000 cm/s^2, average errors were less than 150 cm/s^2 (< 17%, Tables 1 and 2, Fig. 4). The general linear model explained over half of the variance in initial acceleration error ($R^2 = 0.60$, $p < 0.001$). Significant main effects of mass ($p < 0.001$) and treadmill ($p = 0.03$) were observed. For every 10 kg of mass on the treadmill, the error shifted by about + 0.5% (β (S.E.) = 0.05 (0.04)%), generally resulting in less total error. The interaction of direction, mass, and commanded acceleration approached significance ($p = 0.09$), slightly shifting error in the negative direction for larger perturbations and more mass. The interaction of treadmill and commanded acceleration also approached significance ($p = 0.09$), with slight, positive shifts in error for the ActiveStep® and front AMTI belts at larger perturbation magnitudes. In order to explore the observed main effect of treadmill, we compared estimated marginal means at a mass of 63.8 kg and commanded acceleration of 750 cm/s^2. The front AMTI belt had more negative error than the back AMTI belt or ActiveStep® ($p < 0.001$, Fig. 4). The latter two belts did not demonstrate differences in error ($p = 0.18$).

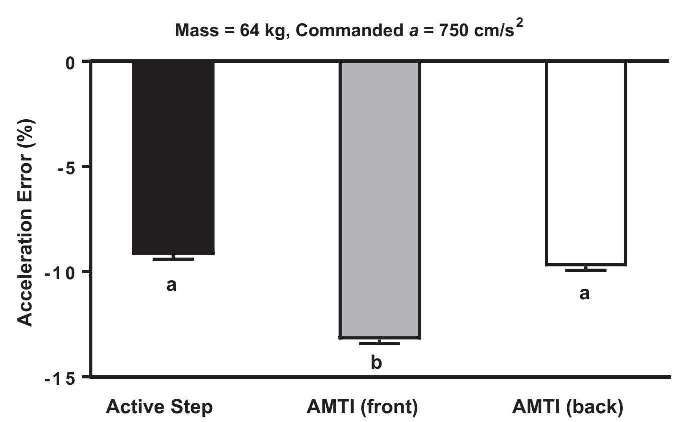

Fig. 4. Estimated marginal means (standard errors) of relative errors (observed – commanded / commanded) in initial average acceleration, estimated at a commanded acceleration of 750 cm/s^2, bearing a mass of 64 kg. Significant ($p < 0.001$) between-treadmill-belt differences are noted with different letter labels (a, b, c).

4. Discussion

The purpose of this study was to characterize and compare the accuracy of the ActiveStep® and AMTI treadmills in delivering rapid belt translations to standing human participants. We anticipated that the ActiveStep® would have better accuracy than the AMTI device. Although the ActiveStep® did, in some cases, have significantly less error, these differences were generally small and not consistent across perturbation magnitudes and masses on the treadmill.

For perturbations studied here, the ActiveStep® had slightly more displacement error (Fig. 2) and slightly less acceleration error (Fig. 4). Velocity accuracy was dependent upon the perturbation magnitude and mass on the treadmill (Fig. 3). These larger perturbations are within the range of previously reported multiple-stepping thresholds, or the anteroposterior perturbation magnitudes that consistently elicited more than one step in healthy adults [7]. In our proposed multiple-stepping threshold protocol, the size of the perturbation is progressively increased with each trial, with each “level” incremented by 4 cm (20 cm/s, 100 cm/s^2). In order to use different treadmills to measure the same outcome, between-treadmill differences in accuracy should be small enough so that adjacent levels do not overlap (i.e. < 50% of the increment). For the largest perturbation size studied here (i.e. 40 cm), this reasoning equates to a criteria of about 5% relative error. For all parameters, no between-treadmill differences in error exceeded these criteria (Figs. 3 and 4). So, our results support the use of these treadmills as a consistent means of evaluating stepping thresholds.

Given our new understanding of how acceleration error is dependent on subject mass, the aforementioned stepping thresholds, which are estimated from commanded acceleration values [7], could be overestimated for those with less mass. However, we believe the acceleration error is too small to alter the validity or reliability of this outcome. For example, stepping thresholds, as measured by the same ActiveStep® device, had no change in reliability (ICC(2,1) = 0.87–0.97) after adjusting previously published accelerations [7] using parameter estimates from our general linear model. In a study of 112 older women [23], thresholds calculated from commanded accelerations (expressed as N·m) were significantly correlated with age ($r = -0.38$– -0.64). Adjusting these accelerations resulted in less than a 0.005 change in correlation coefficients. Therefore, it does not appear that the errors evident from our study are clinically meaningful.

Between-treadmill differences in accuracy may be due to unique control strategies. The software to control the AMTI treadmill prioritized displacement over peak velocity (Fig. 1), intentionally cutting peak velocities short in order to reach the target displacement with a given deceleration. This control scheme may explain the smaller displacement error of the AMTI belts compared to the ActiveStep® (Fig. 2). Although we do not know the ActiveStep® control scheme, we can infer target control parameters from our data. The peak velocity precision of the ActiveStep® in weighted conditions, as quantified by the coefficient of variation (Table 2), was small (< 1%) compared to that of the AMTI belts (2–3%). Despite this inferred focus on velocity by the ActiveStep®, its accuracy was not superior to that of the AMTI belts in our largest, most-weighted conditions (Fig. 4B and D).

With participants initially standing on two belts of the AMTI (Fig. 1C), as opposed to one belt of the ActiveStep®, between-treadmill differences were not solely due to differences in equipment and control schemes. Instead, results may have been influenced by the smaller, partial body-weight load on the AMTI belts at the beginning of the perturbation, as well as the forceful impact of the stepping limb on unweighted AMTI belts. We chose to have participants stand at the belt junction in order to give equal recovery spaces for anterior and posterior perturbations, as well as to allow sufficient single-belt target spaces for recovery steps (i.e. limiting the influence of the junction on step placement). Changing this initial participant position may alter the outcomes of this study.

The observed treadmill perturbation characteristics were comparable to that of platform translations. In a validation of a moveable platform delivering shorter (5–10 cm) and slower (10–50 cm/s) translations, displacement and velocity errors were not markedly different (Error = 0.4%–1.6%, Precision < 4%) than those observed in this study [24]. This study, however, conducted evaluations in unweighted and weighted conditions from a single participant. By including a range of participant masses, we are better equipped to quantify the influence of mass on accuracy. Another study delivered longer (60 cm), but slower (peak velocity = 50 cm/s) platform translations with notable accuracy and precision (R^2 with commanded waveforms = 0.99 ± 0.001) [25]. It should be noted, however, that the coefficient of determination does not necessarily infer agreement between commanded and observed measures. Platform translations applied by the CAREN system, a setup that includes an instrumented treadmill embedded in a movable platform, were precise (< 1.2%) [26]. However, the observed velocity of these translations fell notably short of target values when velocities exceeded 0.4 m/s (accelerations = 4.5 m/s^2). Of note, the CAREN treadmill itself does have the acceleration capabilities necessary to induce falls [11]. The accuracy and precision of this system's belt movements, however, have not been reported.

This is the first study to evaluate the accuracy of treadmill-delivered postural perturbations. We have demonstrated that these two treadmill models have the accuracy and precision necessary for research applications. Our results may not be representative of all treadmills from these manufacturers. Accuracy and precision may also be dependent upon the belt tightness, a factor that can be different between treadmills and can change within a treadmill with use. Additional studies are needed to evaluate other treadmills, other perturbation magnitudes, and perturbations delivered during walking — a capability that treadmills uniquely provide [18].

Conflicts of interest

The authors declare that they have no competing interests.

Ethical approval

This protocol was approved by the Mayo Clinic Institutional Review Board (IRB# 14-000939).

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