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Onset timing of treadmill belt perturbations influences stability during walking

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ABSTRACT

Split-belt treadmills have become popular tools for investigating stability during walking by using belt accelerations to induce slip-like perturbations. While the onset timing of destabilizing perturbations is a critical determinant of an individual's stabilizing response, previous studies have predominantly delivered belt acceleration perturbations at heel strike or have not explicitly controlled onset as a percentage of the gait cycle. To address this gap, we 1) developed an algorithm to target transient increases in unilateral belt speed to begin at specific percentages of the walking gait cycle, 2) validated the algorithm's accuracy and precision, and 3) investigated the influence of different onset timings on spatial stability measures. We evaluated desired onset timings of 10, 15, 20, and 30% of the gait cycle during walking at 1.25 m/s and measured step lengths and widths, as well as anteroposterior and mediolateral margins of stability during the perturbed and four recovery steps in 10 able-bodied participants. From 800 perturbations, we found a mean (standard deviation) delay in onset timing of 5.2% (0.9%) of the gait cycle, or 56 (9) ms. We hypothesized later onset timings would elicit more stabilizing responses due to the less stable configuration of the body during late vs. early single stance. Our data generally supported this hypothesis – in comparison to earlier onset timings, later onset timings precipitated greater stabilizing responses, including larger step lengths, step widths, and anteroposterior/mediolateral margins of stability on the perturbed step, in addition to shorter step lengths and wider step widths on the first step post-perturbation.

1. Introduction

Rapid belt accelerations on split-belt treadmills can emulate slip-like perturbations to study dynamic stability recovery mechanisms with precise control. Although rapid belt accelerations are not identical to slips, the destabilizing effects of both forward pitching slips and belt accelerations are similar, causing a more anterior center of mass position relative to the base of support during the perturbed step (Debelle et al., 2020). Previous work has explored the influence of the direction (*i.e.*, acceleration vs. deceleration) and magnitude of single belt accelerations during walking on stability (Ilmane et al., 2015; Kagawa et al., 2011; Lee et al., 2019; van den Bogaart et al., 2020), but the relationship between gait phase progression at belt acceleration onset and stability measures has not been systematically explored. The majority of studies using rapid belt acceleration perturbations during walking have targeted heel strike as the perturbation onset (Figura et al., 1986; Kagawa et al., 2011; Liu et al., 2018; Roeles et al., 2018; van den Bogaart et al., 2020; Yang et al., 2013), or have not explicitly controlled the onset to begin at a certain percentage of the gait cycle (Berger et al., 1984; Gholizadeh et al., 2019; Lurie et al., 2013; Madehkhaksar et al., 2018). From experiments using other types of perturbations, such as lateral pushes, trips, and force plate translations, the onset timing of a perturbation has been shown to influence the stabilizing response (Eng et al., 1994; Hof et al., 2010; Tang and Woollacott, 1999). Our goals in this work were: 1) to describe and validate the performance of an algorithm that controls the onset of unilateral belt accelerations to begin at specified percentages of the walking gait cycle, and 2) to preliminarily investigate the influence of different onset timings on stability measures. To guide our secondary

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goal, we hypothesized onset timings later in single stance would be most destabilizing, and hence would elicit more stabilizing responses, since the center of mass is most anterior to the base of support in late single stance (Debelle et al., 2020).

2. Methods

2.1. Study design and procedures

Ten healthy participants (Table 1) walked at 1.25 m/s on an instrumented split-belt treadmill (CAREN; Motek, Netherlands; Fig. 1A). Following a 5-minute treadmill acclimation, participants continued to walk for approximately 30 min with unexpected belt accelerations (Zeni and Higginson, 2010). Individual belts were commanded to accelerate from 1.25 m/s to reach and hold at 2.5 m/s, then decelerate to 1.25 m/s (Fig. 2A). The commanded accelerations/decelerations were 15 m/s^2 and the duration of the entire perturbation was ~30% (~340 ms) of the gait cycle. A 10-camera motion capture system (Vicon, Oxford, UK) collected positions of reflective markers placed on bilateral heels, PSIS, and second metatarsals (MT2) at 100 Hz (Fig. 1A). Onsets of belt accelerations were targeted at 10, 15, 20, and 30% of the gait cycle (Fig. 1B). Each participant was perturbed 10 times on each leg at each onset timing (*i.e.*, 80 perturbations per participant). 40 and 50% onset timings were investigated but not analyzed since the perturbed feet did not reach maximum velocity (2.5 m/s) before toe-off. The timing and leg of each perturbation were randomized, and participants took an average of 15 steps (standard deviation: 4 steps) between perturbations. All participants provided informed consent to the protocol approved by the local Institutional Review Board.

2.2. Timing algorithm

Perturbation timing was implemented using D-Flow (Motek, Netherlands). Heel strikes and corresponding gait cycles were identified in real-time based on pelvis and heel marker positions (Zeni et al., 2008). After a perturbed leg and onset timing were selected, the algorithm would wait until the next heel strike on the selected leg. The algorithm then calculated the delay to deliver the perturbation at the specified gait cycle percentage. This delay was the product of the desired onset timing (*e.g.*, 10% of the gait cycle) and the duration of the previous gait cycle of the selected leg.

To validate the algorithm, perturbation onset time ($t_{Pert\ Onset}$) was measured as the elapsed time from heel strike to when velocity of the MT2 marker of the perturbed foot crossed above 110% of the unperturbed walking speed (*i.e.*, above 1.375 m/s). This threshold was similar to the minimal detectable change in self-selected walking speed for young adults (0.18 m/s; Washabaugh et al., 2017), so the measured onset timing could be interpreted as when the perturbation first became statistically meaningful. Absolute onset delays were calculated using Eq. (1):

$$Onset\ Delay_{Absolute} = t_{Pert\ Onset} - \frac{Desired\ Onset\ Timing}{100} \cdot t_{Prev\ Ips\ GC} \quad (1)$$

where $t_{Prev\ Ips\ GC}$ is duration of the ipsilateral gait cycle preceding the perturbation, and *Desired Onset Timing* is a percentage of the gait cycle. Absolute delays were converted to percentages of the gait cycle using Eq. (2):

$$Onset\ Delay_{Percent\ Gait\ Cycle} = \frac{Onset\ Delay_{Absolute}}{t_{Prev\ Ips\ GC}} \cdot 100 \quad (2)$$

Accuracy of the algorithm was validated using mean onset delays, while precision was validated using standard deviation of onset delays.

2.3. Data processing

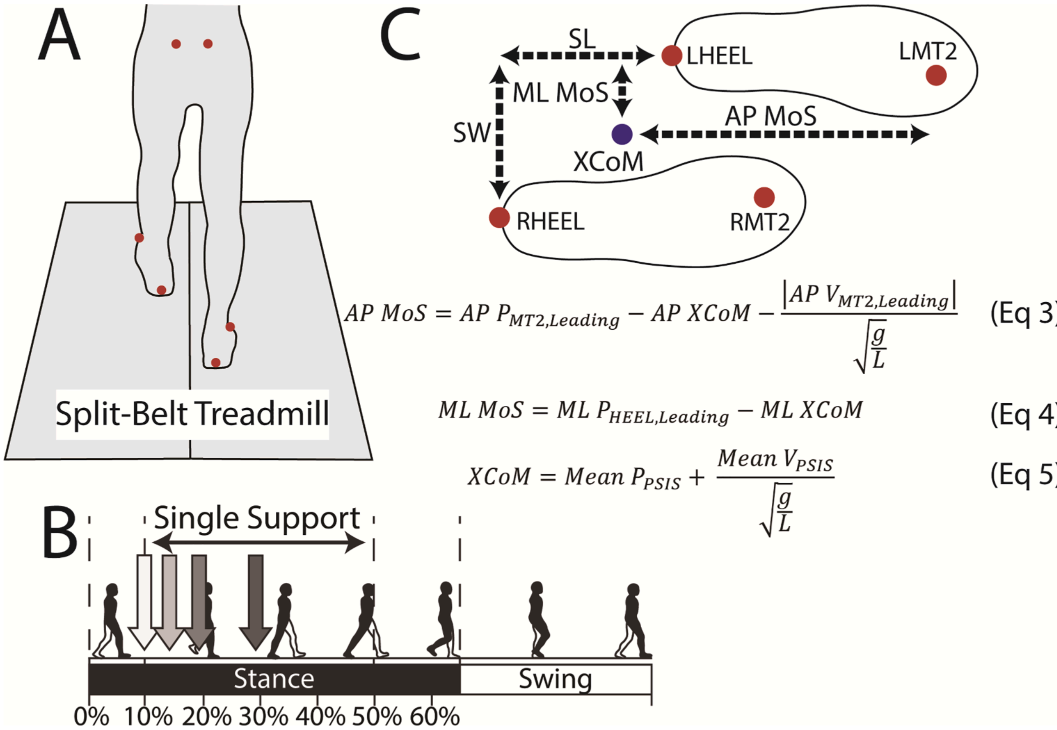
Data were analyzed in MATLAB (Mathworks, Natick, MA). Marker trajectories were low pass filtered at 6 Hz. Stability metrics are graphically presented in Fig. 1C. All metrics were calculated at heel strike. The perturbed step (S0 in Fig. 2) was defined by the first heel strike following, and on the side contralateral to, that of the perturbation. The differences in anteroposterior and mediolateral positions of the heel markers were used to calculate step length and step width, respectively. Margins of stability (Hof et al., 2005) were approximated using marker positions similar to McAndrew Young et al., 2012. Anteroposterior margin of stability (AP MoS) was approximated as the difference between the leading leg's anteroposterior MT2 position and the extrapolated center of mass (XCoM), corrected by the velocity of the leading MT2 marker scaled by leg length and gravity (Beltran et al., 2014; Süptitz et al., 2012; Fig. 1C Eq. (3)). Mediolateral margin of stability (ML MoS) was approximated as the mediolateral distance between the heel marker of the leading leg and XCoM (Fig. 1C Eq. (4)). Use of heel vs. lateral foot markers introduces an offset that underestimates ML MoS relative to literature, but should not affect relative differences in ML MoS across onset timings (Roden-Reynolds et al., 2015). XCoM was approximated using the mean PSIS position as the center of mass and the equations of Hof et al., 2005 (Fig. 1C Eq. (5)). A larger AP/ML MoS indicates increased stability in the anterior/lateral directions, respectively.

2.4. Statistics

Step length was the only outcome significantly affected by leg side following perturbations ($p = 0.046$, $p > 0.205$ for all other outcomes), but this difference was less than the minimal detectable change of this variable (0.47 cm vs. 1.88 cm; Rabago et al., 2015). Thus, left and right perturbations were combined. Shapiro-Wilks tests for normality were not significant for outcome measures ($p > 0.065$) except for step length ($p < 0.041$). Thus, Friedman's test with Dunn-Bonferroni post-hoc tests were used to assess the influence of perturbation timing on step length. For AP MoS, ML MoS, and step width, two-factor linear mixed models with a random effect of participant and a fixed effect of perturbation

Table 1
Participant demographics. SD = standard deviation.

Participant	Gender	Age (years)	Stature (cm)	Weight (kg)	Right Leg Length (cm)
1	M	29	189.0	73.2	100.0
2	M	18	176.5	73.5	89.4
3	M	23	173.7	89.0	88.9
4	M	27	183.5	87.5	95.2
5	M	23	188.0	73.0	100.3
6	F	22	156.5	47.8	83.6
7	M	21	181.5	76.6	93.3
8	M	22	170.0	74.5	92.0
9	M	25	183.8	82.4	96.5
10	F	28	161.0	63.1	86.3
Mean (SD)		24 (3)	176.4 (11.1)	74.1 (12.0)	92.6 (5.6)



$$AP\ MoS = AP\ P_{MT2,Leading} - AP\ XCoM - \frac{|AP\ V_{MT2,Leading}|}{\sqrt{\frac{g}{L}}} \quad \text{(Eq 3)}$$

$$ML\ MoS = ML\ P_{HEEL,Leading} - ML\ XCoM \quad \text{(Eq 4)}$$

$$XCoM = Mean\ P_{PSIS} + \frac{Mean\ V_{PSIS}}{\sqrt{\frac{g}{L}}} \quad \text{(Eq 5)}$$

Fig. 1. (A) Collection setup with reflective marker placements. (B) Targeted onset timings of belt acceleration perturbations as a percentage of the gait cycle. Onset timings were targeted to single support, with 30% of the gait cycle hypothesized to be most destabilizing since the center of mass is most anterior to the base of support in late single stance. (C) Spatial outcome measures measured at a hypothetical left heel strike with equations used to calculate AP and ML MoS. SL = step length, SW = step width, L/RMT2 = L/R second metatarsal heads, XCoM = extrapolated center of mass, AP/ML MoS = anteroposterior/mediolateral margin of stability, g = gravitational constant, L = leg length, P = positions, V = velocities.

timing were run with Bonferroni-corrected pairwise post-hoc tests. Statistical tests were run separately for each step relative to the perturbation since models including both step number and onset timing showed significant interactions between step number and onset timing ($p < 0.001$). All statistical analyses were performed in SPSS (IBM, Chicago, IL), with significance concluded when $p < 0.050$.

3. Results

3.1. Perturbation timing validation

The average accuracy across onset timings, as quantified by mean onset delay, was 5.2% of the gait cycle (56 ms), while the average precision, as quantified by the standard deviation of onset delay, was 0.9% of the gait cycle (9 ms; Table 2). The commanded and measured velocities are shown in Fig. 2A and B, respectively.

3.2. Stability outcome measures

On the perturbed step ($p < 0.001$) and the first step post-perturbation ($p = 0.016$), there was an effect of onset timing on AP MoS (Fig. 2C). On the perturbed step, later onset timings resulted in larger AP MoS – pairwise comparisons showed 30% onset timings had larger AP MoS than all other timings ($p < 0.001$), while 20% onset timings had larger AP MoS than 10% onset timings ($p = 0.018$). Conversely, for the first step post-perturbation, earlier onset timings resulted in larger AP MoS – 10% onset timings resulted in larger AP MoS than 30% onset timings ($p = 0.017$).

On all steps following the perturbation, there was an effect of onset timing on step length ($p = 0.002$ for perturbed step, $p < 0.001$ for all steps post-perturbation; Fig. 2D). On the perturbed step, later onset timings resulted in larger step lengths – 20 and 30% onset timings had larger step lengths than 10% onset timings ($p < 0.019$). For the first step post-perturbation, later onset timings resulted in smaller step lengths – 20 and 30% onset timings had smaller step lengths than 10% onset timings ($p < 0.019$) and 30% onset timings had smaller step lengths than 15% onset timings ($p = 0.003$). For the second through fourth steps post-perturbation, earlier onset timings resulted in smaller step lengths – 10% onset timings had smaller step lengths than 20 and 30% onset timings ($p < 0.034$), and for the second and third steps post-perturbation 15% timings had larger step lengths than 30% onset timings ($p = 0.034$).

On the perturbed step alone, there was a significant effect of onset timing on ML MoS ($p = 0.003$; Fig. 2E), with later onset timings tending to have larger ML MoS than earlier timings. 20% onset timings had significantly larger ML MoS than the 10 and 15% onset timings ($p < 0.017$).

On the perturbed step and first step post-perturbation, there was an effect of onset timing on step width ($p < 0.047$; Fig. 2F). For the perturbed step, 20% onset timings resulted in the largest step widths – 20% onset timings had significantly larger step widths than 10% onset timings ($p = 0.047$). For the first step post-perturbation, despite a significant effect of onset timing and a trend of 20% onset timings resulting in the largest step widths, no pairwise comparisons were significant ($p > 0.061$).

4. Discussion

4.1. Algorithm evaluation

To evaluate the algorithm's accuracy, we defined a successful single stance perturbation as a one that induced maximum belt speed of the stance foot before double support. Thus, based on our commanded belt acceleration, an unperturbed gait cycle duration, and the latest onset timing, the *a priori* maximum acceptable onset delay was 13% of the gait cycle. The measured accuracy fulfilled this requirement, with perturbations being delivered 5.2% of the gait cycle after the desired onset

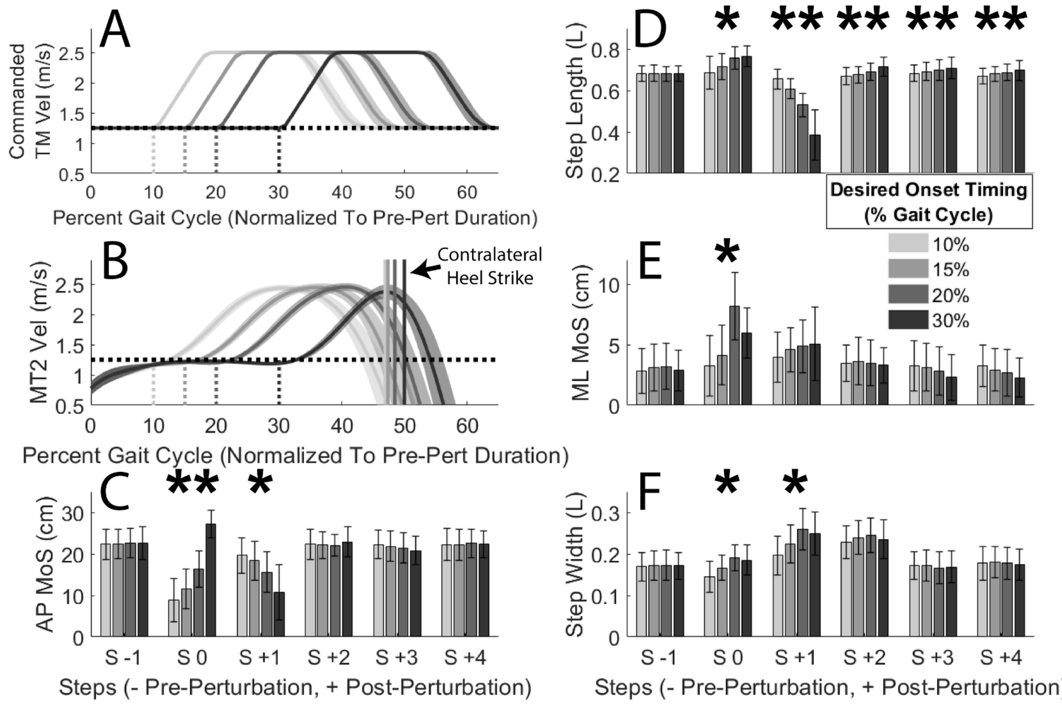


Fig. 2. (A) Commanded treadmill velocity profiles and (B) measured MT2 marker velocities of the perturbed foot for all analyzed perturbations. Gait cycles were normalized to the duration of the ipsilateral gait cycle preceding the perturbation. Solid vertical lines indicate the end of single support. Shaded areas represent ± 1 standard deviation. Dotted vertical lines represent desired start times. Horizontal dotted lines represent the unperturbed velocity (1.25 m/s). (C–F) Across-subject mean spatial stability measures. Error bars represent ± 1 standard deviation. L = metric normalized to participant leg length. AP MoS = anteroposterior margin of stability. ML MoS = mediolateral margin of stability. * = $p < 0.050$ and ** = $p < 0.001$ for effect of timing for that step. Pairwise comparisons are described in text.

Table 2
Accuracy and precision validation measures in percentage of gait cycle and milliseconds. Perturbation onset time was measured as the elapsed time from heel strike to when velocity of the MT2 marker of the perturbed foot crossed above 110% of 1.25 m/s (*i.e.*, above 1.375 m/s). Absolute onset delays in milliseconds were calculated by subtracting the desired onset percentage multiplied by the duration of the ipsilateral gait cycle preceding the perturbation from the measured onset time (Eq. (1)). Absolute onset delays were converted to percentages of the gait cycle by dividing by the duration of the ipsilateral gait cycle preceding the perturbation (Eq. (2)).

	Desired Onset Timing (% Gait Cycle)	10%	15%	20%	30%
Accuracy Metrics	Mean Onset Delay (% Gait Cycle)	5.1	5.1	5.2	5.4
	Mean Onset Delay (ms)	54	55	56	58
Precision Metrics	Standard Deviation of Onset Delay (% Gait Cycle)	0.9	0.9	0.9	1.0
	Standard Deviation of Onset Delay (ms)	9	9	9	10

time. Perturbations also had to be delivered such that they did not “spill over” into other onset timings. Thus, our *a priori* threshold for acceptable precision was half the minimum difference in targeted onset timings (*i.e.*, 2.5% of the gait cycle), which would ensure ~95% of the perturbations were delivered within ±5% of the gait cycle centered at each desired onset timing. Our measured accuracy (0.9% of the gait cycle) also fulfilled this criterion.

4.2. Effects of onset timing on stability

As the walking gait cycle progresses, the extrapolated center of mass first crosses in front of the anterior boundary of the base of support as early as 6% of the gait cycle and continues to move anteriorly until contralateral heel strike (Debelle et al., 2020). Thus, from the onset of single support to the beginning of double support, the negative AP MoS, and hence the risk of a forward fall, increases, while the time to make a corrective foot placement decreases (Hof et al., 2010; Vlutters et al., 2018). This spurred our hypothesis that belt accelerations with onset timings later in single stance would elicit more stabilizing responses, which was generally supported by our data.

Following a posterior perturbation of the base of support (mechanically analogous to an anterior center of mass push), *larger* step lengths on the perturbed step are stabilizing since the leading limb can apply a larger posterior braking force (Joshi and Srinivasan, 2019; Wang and Srinivasan, 2014), and more mechanical work is dissipated by collision (Donelan et al., 2002; Kuo, 2002). By decreasing center of mass velocity, this increased braking/dissipation precipitates a *smaller* post-perturbation step length (Kuo and Donelan, 2010), which is more stabilizing during recovery (Espy et al., 2010). Such stabilizing step length changes have been observed following both belt accelerations (Afschrift et al., 2019; Debelle et al., 2020; Roeles et al., 2018; Sloot et al., 2015) and anterior pelvic pulls (Vlutters et al., 2018). Consistent with these mechanisms and literature, we found later onset timings elicited more stabilizing step lengths on both the perturbed step and first step post-perturbation in comparison to earlier onset timings, with trends in AP MoS following these changes in step length.

In the frontal plane, later onset timings generally resulted in more stabilizing (Donelan et al., 2004), larger step widths and ML MoS for the perturbed and post-perturbation steps, except for onset timings of 30%. Coupling between AP and ML foot placement may explain larger step widths on the perturbed step (Bauby and Kuo, 2000; Kim and Collins, 2017), which together with a larger ML MoS elicits more lateral subsequent foot placement and larger step widths on the first step post-perturbation (Rankin et al., 2014). Less stabilizing step widths and ML MoS for 30 vs. 20% onset timings may stem from 1) perturbations with 30% onset timings being less destabilizing since double support begins midway through the perturbation (Fig. 2B) or 2) insufficient time for foot positioning exacerbated by the anterior perturbation shortening single-stance duration (Roeles et al., 2018; Sloot et al., 2015; Vlutters et al., 2018).

5. Limitations

Our participants were young able-bodied individuals, so it is unknown how these results would generalize to participants with balance impairments. Further, the belt accelerations we applied in early stance do not represent usual balance challenges faced in daily life – early slips are associated with backwards pitching and would be better represented by a belt deceleration beginning at heel strike (Heiden et al., 2006; Nagano et al., 2013). Despite these limitations, belt accelerations provide an accessible tool to investigate the stabilizing responses individuals use while walking.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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