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RESEARCH ARTICLE

Stepping behaviour contributes little to balance control against continuous mediolateral trunk perturbations

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ABSTRACT

Human bipedal gait is exceptionally stable, but the underlying strategies to maintain stability are unclear, especially in the frontal plane. Our study investigated balance strategies of healthy adults subjected to continuous mediolateral oscillations at the trunk during walking. We used a backpack with a passive inverted pendulum to create perturbations that were fixed, in-phase or out-of-phase with subjects' trunk. We evaluated subjects' corrective strategies and whether they yielded equivalent stability, measured by the margin of stability and the local divergence exponent. The margin of stability measure quantified adjustments in step behaviour relative to the centre of mass, and the local divergence exponent measure characterized the chaotic behaviour of the system throughout the entire trial. Among the conditions, there was no significant difference in the step width. We found a higher margin of stability for the out-of-phase condition and the lowest local divergence exponent for the in-phase condition and the highest for the fixed condition. These results indicate that the in-phase condition was more stable with respect to fluctuations throughout gait cycles, and the out-of-phase condition was more stable in terms of foot placement relative to centre of mass. To maintain equivalent or greater gait stability, subjects elected to reduce the motion of their centre of mass rather than alter step width. The reduction in centre of mass motion without a reduction in step width suggests direct control of the centre of mass to maintain stability was preferred over adjusting stepping behaviour.

KEY WORDS: Biomechanics, Locomotion, Gait stability, Balance strategies, Perturbations

INTRODUCTION

During walking, humans can maintain stability by overcoming various perturbations and environmental disturbances without falling. Gait stability is largely achieved through interactions between the body centre of mass (CoM) and the base of support (BoS) created by the feet contacting the ground (Bruijn and van Dieën, 2018; Hof et al., 2005). While the specific mechanisms and strategies to maintain stability are still unclear, there have been several proposed mechanisms. The interactions between the CoM and BoS can be modified by the stepping strategy (i.e. controlling foot placement), the ankle strategy (i.e. the modification of centre of pressure under the foot) (Hof et al., 2010), and upper body momentum strategy (Hof, 2007).

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Probing mediolateral balance during walking could reveal insights into the control strategies implemented by the central nervous system to maintain stability. Stabilization in the mediolateral plane requires active control with visual–vestibular feedback, whereas the sagittal plane is passively stable (Bauby and Kuo, 2000). A previous study investigating gait stabilization used external springs attached to the hips that resist motion in the lateral direction (Donelan et al., 2004). That study found that when subjects are given their choice of step width, they choose to lower their step width. Increasing step width can increase the BoS but is also energetically costly (Donelan et al., 2001). The stabilization supplied by the springs reduced the requirement for a wide BoS and also reduced the metabolic cost of walking by 5.7%.

Previous studies with external perturbations to the CoM found that the stepping strategy was crucial to maintaining balance in the frontal plane (Hof et al., 2010; Rankin et al., 2014; Vlutters et al., 2018, 2016). A study applying random lateral perturbations to the body CoM (Hof et al., 2010) suggested that balance was maintained using two different mechanisms. The first, yet slowest, mechanism is the stepping strategy, which can compensate for large perturbations by maintaining the distance from the CoM to the foot in the next step. The second mechanism is the ankle strategy that shifts the centre of pressure (CoP) in the frontal plane to maintain stability. Although the ankle strategy is faster, it can only move the CoP by 2 cm at most, approximately 10 times less than with the stepping strategy. A similar study (Vlutters et al., 2016) applied perturbations to the CoM in both the sagittal and frontal plane and found that perturbations in the frontal plane, but not the sagittal plane, were compensated with mediolateral foot placement adjustment that was proportional to the CoM mediolateral velocity.

In addition to stepping, the trunk can be used to maintain balance. The rotation of the trunk can indirectly move the body CoM through conservation of angular momentum (Hof, 2007). Some studies have explored the impact of trunk perturbations on gait stability, such as during load carriage (Walsh et al., 2018). Defining stability using the local divergence exponent (LDE) of trunk velocity, this study found that carrying a load of steel weights that was 15% of the subject's body mass decreased gait stability. The same study also used water as an added mass to create an unstable load. The LDE analysis showed that the use of an unstable load resulted in lower gait stability than with an equivalent stable mass. Although this does suggest that a moving carried load reduces gait stability, the use of water does not allow for a predictable continuous perturbation to be applied through the load carriage.

In the current study, we applied continuous, sinusoidal perturbations to the trunk with an oscillating mass to provoke changes in gait stability and overall stepping behaviour. The perturbations were created using a backpack with a mass on top of an inverted pendulum (Fig. 1) (Martin and Li, 2018). By changing the spring configuration, the natural frequency of the inverted pendulum can be set to create conditions where the pendulum

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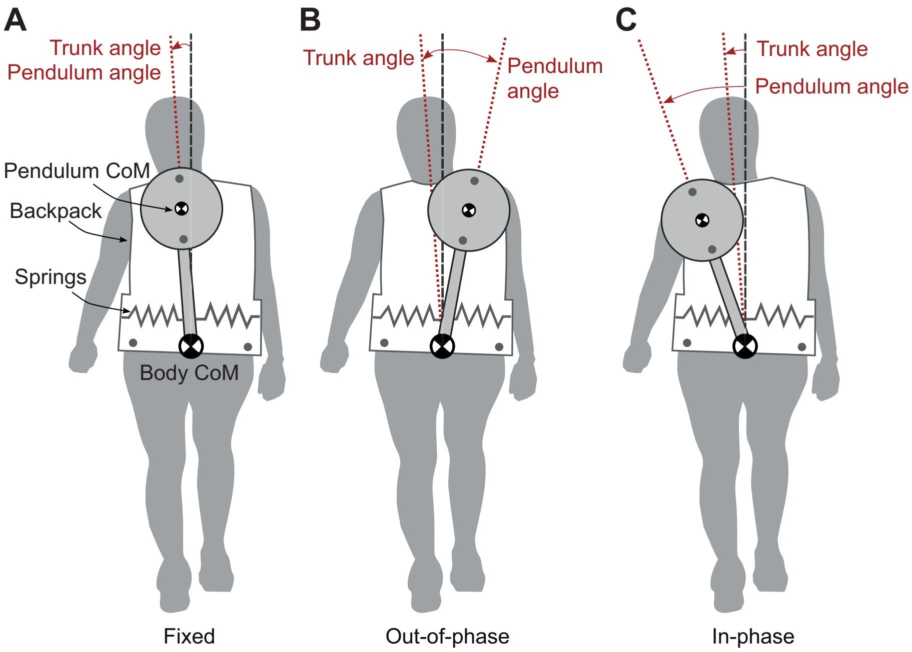

Fig. 1. Oscillating mass perturbation device. The device used to create the perturbation is a backpack with an inverted pendulum that uses linear springs to create the restoring force. During locomotion, the mediolateral motion of the trunk creates a base excitation that causes the pendulum to oscillate around a vertical equilibrium position. The pendulum can (A) be locked to the backpack, (B) oscillate out-of-phase with the subject's trunk or (C) oscillate in-phase with the subject. CoM, centre of mass.

moves in-phase and out-of-phase with the mediolateral motion of the wearer's CoM during walking.

It is unclear what stabilizing and destabilizing effects the pendulum will have on the subjects. Using an inverted double pendulum as a simple model of single support during gait, the device can create two seemingly opposing impacts on gait stability. This illustrative model has a main body mass on top of an inverted pendulum leg pinned to the ground that is displaced in the clockwise direction (Fig. 2). To maintain stability, the body CoM must be moved left to be aligned over the base of support. The pendular positional impact of the device (Fig. 2A,B) results in the pendular mass changing the total position of the CoM. The out-of-phase condition is potentially more stabilizing as the total CoM would be shifted to a more neutral position. The opposite would hold true for the in-phase case, with the pendulum mass creating a larger deviation from upright. The pendular torque impact of the device (Fig. 2C,D) is due to the reaction torque that arises from the gravitational pendulum torque. In the out-of-phase condition, this reaction torque moves the body CoM away from vertical, causing a destabilizing effect. In contrast, the direction of the reaction torque in the in-phase condition coincides with that of the goal motion, causing a stabilizing effect by moving the body CoM towards upright.

A previous study using the same oscillatory backpack device (Martin and Li, 2018) suggested that out-of-phase behaviour could be stabilizing. That study investigated the mechanical and metabolic

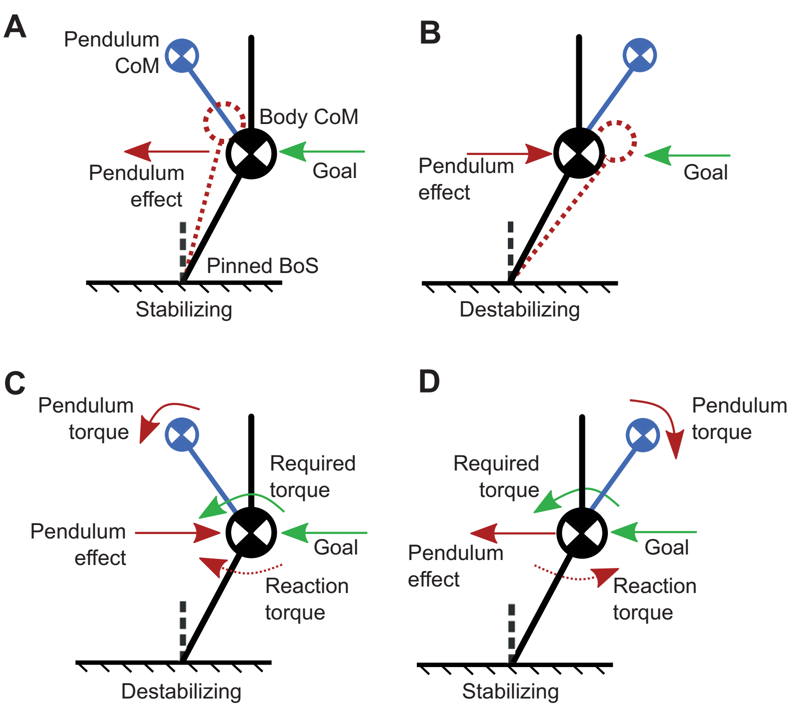

Fig. 2. Potential stabilizing and destabilizing impact of the device during single support represented by a double pendulum pinned to the ground. (A) The pendular position behaviour of the out-of-phase case provides stabilization through moving the total CoM (pendulum effect, red horizontal arrow) to a more neutral position (i.e. in the same direction as the goal, represented by the green horizontal arrow). (B) The pendular position behaviour of the in-phase condition creates a destabilization as it moves the total CoM further from a neutral position. (C) The reaction torque created by the out-of-phase device condition causes destabilization by moving the body CoM away from the base of support (BoS). (D) The reaction torque from the pendulum stabilizes the body CoM by moving it towards the neutral position. The dotted red pendulum in A and B indicates the position of the total CoM.

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changes created by the device with only the out-of-phase perturbations. When given free choice of step width, the subjects elected to lower their step width. This behaviour suggests that the out-of-phase condition increases gait stability as a wide BoS is no longer necessary and that the pendular mass effect of the device plays a more dominant role than the effect of the pendular torque. Although multiple measures of stability exist (Bruijn et al., 2013), we elected to quantify gait stability through two measures, the margin of stability (MoS) and the LDE. The MoS arises from the minimum distance between the extrapolated CoM (XCoM, i.e. the CoM position with an additional CoM velocity term) and the BoS (Hof et al., 2005) and is used to investigate adjustments in interactions between the BoS and CoM. The LDE proposed by Dingwell and Cusumano (2000) as a stability measure is used to measure the local dynamic stability of the CoM motion. Past studies have found that a reduction in step width coincides with a constant MoS (Arvin et al., 2016; Rosenblatt et al., 2012), as the motion of the XCoM is reduced proportionally with step width. Additionally, a reduction in step width leads to greater gait stability, assessed using the LDE (McAndrew Young and Dingwell, 2012). Therefore, we expected that the out-of-phase condition would lead to a constant MoS and a reduced LDE. If the in-phase condition has the opposite effect on stability, subjects should increase step width. This stepping behaviour may still result in a constant MoS if the XCoM motion increases proportionally with step width, as seen in reduced step width conditions. With the increased step width, we also expected that the LDE measure would indicate lower stability (McAndrew Young and Dingwell, 2012).

The goal of our study was to resolve how pendular oscillations at the trunk affect gait stability. We applied three conditions: fixed, out-of-phase and in-phase. To measure stability, we used the LDE to evaluate stability using fluctuations within the gait cycles and the MoS to investigate the alteration in BoS–CoM interactions to compensate for the perturbation in the different conditions. We hypothesized that the MoS measure of stability would remain constant for all three conditions. Furthermore, we hypothesized that the LDE measure of stability would result in the out-of-phase condition being the most stable and the in-phase condition being the least stable.

MATERIALS AND METHODS

To test our hypothesis, we asked healthy, adult subjects to walk with the perturbation device in the three different conditions (fixed, out-of-phase and in-phase). Motion capture and ground reaction force data were collected to measure spatiotemporal parameters and to compute stability in terms of the LDE and MoS. Stability measures of each condition were compared to determine the impact of the oscillation on stability, and spatiotemporal measures were evaluated to determine whether gait modifications could contribute to changes in stability.

Experiment

The perturbation device (Fig. 1) consisted of the inverted pendulum with an oscillating mass of 4.5 kg housed in a rectangular frame attached to a prefabricated backpack frame (the total device mass is 9.1 kg) (Martin and Li, 2018). The backpack frame included a chest strap and a hip belt that could be easily adjusted to ensure a comfortable fit. Up to 10 linear springs were attached to each side of the pendulum to provide a restoring force; the spring constant and base length of the springs were varied to modify the natural frequency of the device. The input lateral motion of the trunk directly results in the oscillation of the pendulum, creating a perturbation that is controlled by the subject's stepping pattern.

Twelve healthy young adults (8 male and 4 female, age 21.8±1.0 years, height 1.802±0.092 m, mass 72.3±11.2 kg) with no prior injuries or pathologies participated in this study. All subjects provided informed consent, and the experiments were approved and conducted in accordance with the General Research Ethics Board of Queen's University.

All participants wore the perturbation device and walked on an instrumented, split-belt treadmill (AMTI Force-Sensing Tandem Treadmill, AMTI Inc., Watertown, MA, USA) at a speed of 1.25 m s^{-1}. The first trial was normal walking for 3 min to allow the subject to become acclimatized to walking on the treadmill and to determine the subject's preferred stride frequency; the ground reaction forces were used to detect right heel strikes to compute the stride frequency. Based on the determined stride frequency, the perturbation device was then configured to one of three randomized conditions: locked, in-phase and out-of-phase, representing the oscillation of the pendular mass relative to the subject's body CoM. The in-phase and out-of-phase conditions were created by changing the spring configuration so that the natural frequency of the inverted pendulum was 130% and 70% of the stride frequency, respectively. These percentages were chosen so that the ratio of the amplitude of the pendular motion should ideally be equal for the in-phase and out-of-phase conditions (Martin and Li, 2019). The pendular mass was 6.4% of the subject's body mass on average. Subjects walked for 6 min on the treadmill with data collection starting after 2 min. Passive infrared markers, placed on the device and on the feet of the subjects, were tracked using eight motion-capture cameras (Qualisys AB, Gothenburg, Sweden) sampling at 100 Hz.

Analysis

As the device covered a large portion of the trunk and hips, we estimated the subject's mediolateral body CoM position with a virtual sacrum marker calculated from the average of the markers on the left and right side of the device. The CoM of the oscillating mass was calculated from markers on the mass (Fig. 1). The overall CoM position was calculated as the combination of the body's CoM and pendulum CoM. Strides were segmented based on ground reaction forces, with gait cycle defined from right heel strike to right heel strike. All reported values were made non-dimensionalized with a combination of standing leg length L and gravitational acceleration $\boldsymbol{g}$. Distance measures (e.g. step length, step width, CoM position) were divided by the subject's leg length L (0.935±0.052 m), step time by $\sqrt{(L/\boldsymbol{g})}$ (0.308±0.009 s) and velocity by $\sqrt{(\boldsymbol{g}L)}$ (3.02±0.08 m s^{-1}).

To determine the effect of a laterally oscillating mass on gait stability, the LDE and MoS were computed for each condition. The LDE is a direct measure of the sensitivity to initial conditions, providing an indication of the chaotic nature of any mechanical system as chaotic systems have a high sensitivity to initial conditions (Dingwell, 2006). The LDE quantifies the average logarithmic rate of divergence of the mechanical system represented in a state space. For our purposes, we interpret a larger LDE as an indication of lower gait stability. The LDE can theoretically be calculated using any kinematic data. The input data we selected for the calculation was the mediolateral velocity of the total CoM (backpack plus body). The velocity data used was for 150 strides that had been normalized to 100 data points per stride (Bruijn et al., 2009a). The mediolateral CoM velocity was selected as the perturbation device acts in the frontal plane and the CoM provides a good representation of the behaviour of the entire system. There was no filtering of the markers prior to calculating the LDE because of the difficulties involved with filtering non-linear signals

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(Bruijn et al., 2009b). The calculation of the LDE was completed using a version (Bruijn, 2017) of the algorithm proposed by Rosenstein et al. (1993). The state space was reconstructed using standard embedding dimension techniques (Mañé, 1981; Takens, 1981), where the state spaces (s) are reconstructed using the normalized velocity (v) and the time (t) delayed copies:

$$s(t) = [v(t), v(t+\tau), v(t+2\tau), \ldots, v(t+(d_E-1)\tau)]. \quad (1)$$

The embedding dimension (d_E) was selected using a Global False Nearest Neighbour test, which found a value of 5 to be sufficient to define the state of the system at all times (Dingwell, 2006). The time delay (τ) used was 10 as this value has been used in other gait analyses (England and Granata, 2007). The LDE was estimated as the slope of the divergence curve from 0 to 0.5 strides (Bruijn et al., 2013, 2009b).

A one-dimensional MoS in the frontal plane was calculated, requiring estimation of the body and oscillating mass CoM and the BoS. Unlike the LDE analysis, the marker positions were filtered using a 4th order Butterworth filter with a cut-off frequency of 6 Hz. The XCoM is then calculated as:

$$\mathrm{XCoM} = \mathrm{CoM} + \frac{v_{\mathrm{CoM}}}{\omega}, \quad (2)$$

where ω is the first eigenfrequency of the inverted pendulum model of walking; the pendulum length used in the eigenfrequency estimation is the leg length of the subject. The position and velocity (denoted CoM and v_{CoM}) are for the total CoM, the combination of the CoM of the body and the CoM of the pendulum. The BoS was estimated using the location of the overall CoP. The minimum MoS was found during single limb support of the right and left leg using ground reaction forces to locate toe off and heel strike events marking the beginning and end of single limb support. This minimum value was averaged between the left and right sides and across the same 150 strides as the LDE analysis. A negative MoS value indicates that the XCoM is lateral to the BoS.

We performed statistical comparisons to evaluate the effect of each condition (fixed, out-of-phase, in-phase) on the stability measures and spatiotemporal parameters. Repeated measures ANOVA followed by *post hoc* *t*-tests with Holm–Šidák corrections for multiple comparisons was used with a significance level of 0.05. The same statistical tests were performed for total CoM, body CoM and pendulum CoM amplitudes to test the effects from each condition. We performed additional tests to determine whether the pendular mass altered the total CoM. The amplitude of the body CoM and the amplitude of the total CoM were compared using paired *t*-tests using the same significance level.

RESULTS

During all trials, the device behaved in the intended in-phase or out-of-phase way. For the fixed case, the pendulum mass approximately matched the body CoM in both phase (6.6±3.0 deg) and amplitude (Fig. 3A). For the out-of-phase case, pendulum mass was approximately 180 deg (149.7±9.1 deg) offset and for the in-phase condition, the pendulum mass moved near synchronously (13.8 ±12.2 deg) with the body CoM but with a larger amplitude. The distance from the pendulum CoM to the body CoM behaved in a consistent manner for all the trials, with the largest distance occurring during the out-of-phase condition (0.0537±0.0165 m, mean±s.d.) and the smallest distance of 0.0069±0.0039 m during the fixed condition (Fig. 3B and Table 1). Both out-of-phase and in-phase

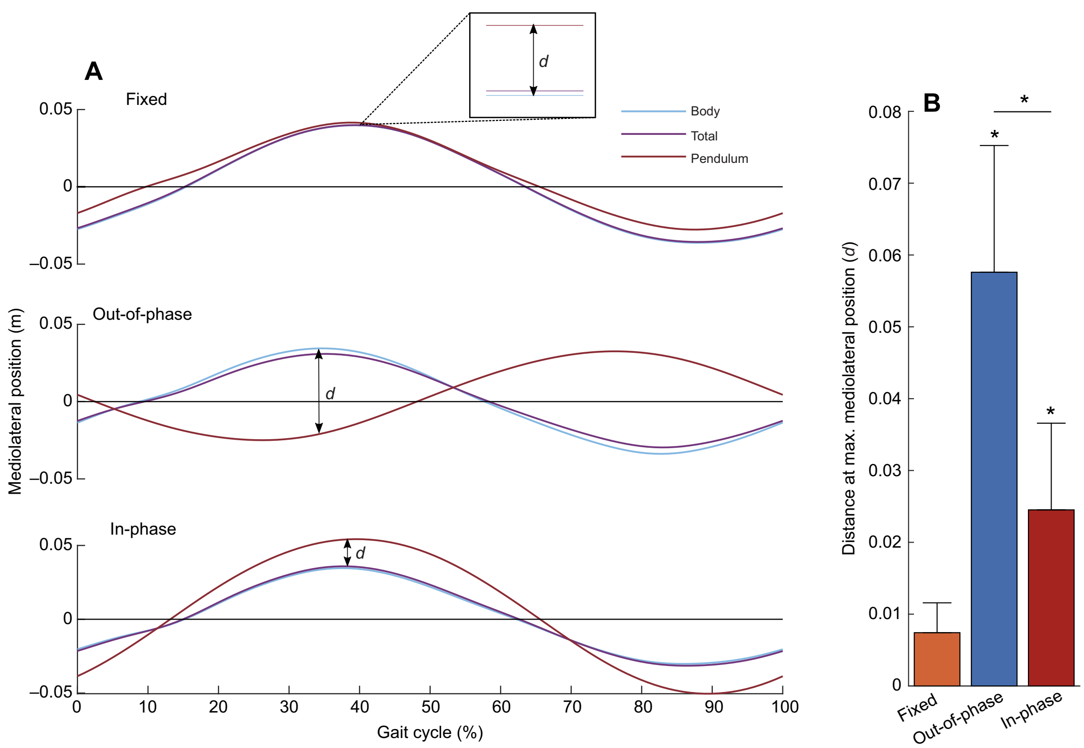

Fig. 3. Mediolateral behaviour of the perturbation device, body CoM and total CoM for the three different configurations. (A) Body, pendulum and total CoM mediolateral position of one representative subject averaged over 150 strides to illustrate fixed, out-of-phase or in-phase behaviour. (B) Mean distance (d) from the body CoM to the pendulum CoM at the maximum mediolateral position of the body CoM during a stride (see arrows in A). Bars are averages across all subjects (N=12), and error bars denote 1 s.d. The asterisk over the individual bars indicates that the condition is statistically different from the fixed condition, and the asterisk over the horizontal line indicates that the out-of-phase and in-phase condition are statistically different (*P<0.05).

Table 1. Summary of quantitative and statistical results

Metric	Fixed	Out-of-phase	In-phase	ANOVA	Fixed vs out-of-phase	Fixed vs in-phase	Out-of-phase vs in-phase
d	0.0074±0.0042	0.0576±0.0176	0.0245±0.012	2.07e−6*	2.1e−6*	0.0014*	6.2e−4*
LDE	2.212±0.194	2.076±0.180	1.957±0.176	2.63e−5*	0.0065*	2.7e−6*	0.0016*
MoS	−0.0189±0.015	−0.0018±0.015	−0.0171±0.0145	3.80e−9*	4.50e−6*	0.26	4.20e−6*
Step width	0.1089±0.0375	0.0991±0.0314	0.0906±0.0260	0.1911	–	–	–
Step length	0.7728±0.0371	0.7613±0.0326	0.7684±0.0308	0.2214	–	–	–
Step time	1.867±0.0586	1.840±0.0590	1.857±0.0593	0.2351	–	–	–
Step width variability	0.0230±0.0077	0.0263±0.0146	0.0242±0.008	0.4452	–	–	–
Step length variability	0.0124±0.0040	0.0143±0.0046	0.0112±0.0033	0.0685	–	–	–
Step time variability	0.0300±0.0097	0.0347±0.0114	0.02695±0.0078	0.0671	–	–	–
Total CoM amplitude	0.03693±0.00875	0.03117±0.0083	0.03255±0.00845	0.0047*	0.0037*	0.0015*	0.34
Body CoM amplitude	0.03709±0.0090	0.03489±0.0090	0.03143±0.00856	0.0066*	0.19	2.4e−4*	0.024*
Pendulum CoM amplitude	0.03501±0.0061	0.04564±0.0113	0.05707±0.00750	2.7e−5*	0.006*	1.50e−7*	0.0039*

The first three columns are mean (±s.d.) values reported in dimensionless units. The fourth column lists the *P*-values from repeated measures ANOVA, and the fifth to seventh columns are *P*-values from *post hoc t*-tests with Holm–Šidák corrections. The asterisks indicate statistically significant results. All statistical tests were conducted with a level of significance of 0.05. *d*, pendulum to body distance; LDE, local divergence exponent; MoS, margin of stability.

maximum distance were significantly different from fixed (*P*=2.1e−6 and *P*=0.0014, respectively) and from each other (*P*=6.2e−4). The amplitude of the pendulum from the average neutral position was 0.0327±0.0057 m, 0.0426±0.0106 m and 0.0534±0.0070 m for the fixed, out-of-phase and in-phase condition, respectively.

The LDE and MoS, as two different measures of stability, disagreed in terms of which conditions were more stable over fixed. The LDE measure of stability (Fig. 4A and Table 1) resulted in the in-phase condition displaying the highest gait stability (1.957, *P*=0.0065) while the out-of-phase condition resulted in gait stability that was higher than that of the fixed condition (2.076, *P*=2.7e−6) but lower than that of the in-phase condition (*P*=0.0016). In contrast, the MoS measure of stability (Fig. 4B) indicated that the out-of-phase condition resulted in higher gait stability with a 91.9% greater margin than fixed (*P*=4.5e−6). The in-phase condition was not significantly different from fixed (*P*=0.26). Therefore, the in-phase condition was more stable than fixed in terms of stride fluctuations, while the out-of-phase was more stable in terms of foot placement relative to CoM.

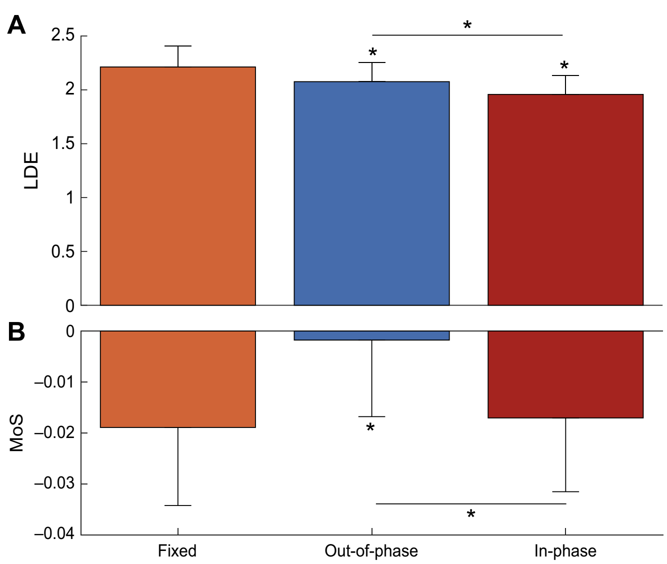

Fig. 4. Stability measures for each condition. (A) Mean and standard deviation of the local divergence exponent (LDE). Lower LDE values indicate greater local dynamic stability. (B) Mean and standard deviation of the average margin of stability (MoS) over the gait cycle. Higher MoS indicates greater stability through a larger margin between the extrapolated CoM (XCoM) and BoS boundaries. Bars are the averages across all subjects (*N*=12), and error bars denote 1 s.d. The asterisk over the individual bars indicates that the condition is statistically different from the fixed condition, and the asterisk over the horizontal line indicates that the out-of-phase and in-phase condition are statistically different (**P*<0.05).

Spatiotemporal parameters were unchanged for all conditions (Fig. 5 and Table 1). There were no significant differences found in step width (*P*=0.1911), step length (*P*=0.2214) and step time (*P*=0.2351). Likewise, there were no significant differences in step width variability (*P*=0.4452), step length variability (*P*=0.0685) and step time variability (*P*=0.0671). Across all conditions, mean step width was 0.0931 m, step length was 0.717 m and step time was 0.571 s.

While subjects maintained similar spatiotemporal measures, they changed the amplitude of their overall CoM in response to the pendular oscillations. For the fixed case, the pendulum CoM and body CoM had no significant differences in amplitude, as expected (Fig. 6 and Table 1). When pendulum motion was allowed, the motion of the total CoM amplitude decreased for both the out-of-phase and the in-phase conditions compared with the fixed condition (*P*=0.0037, *P*=0.0015, respectively). Although the pendulum mass was oscillating at a greater amplitude than the body CoM in the in-phase condition, subjects significantly reduced their body CoM amplitude by approximately 15.3% compared with the fixed condition (*P*=2.4e−4), creating a 11.8% reduction in total CoM (*P*=0.015) compared with the fixed condition. During the out-of-phase condition, the total CoM amplitude was decreased by 10.7% (*P*=4.3e−7) compared with the body CoM amplitude. In the in-phase condition, the total CoM amplitude was 3.6% higher (*P*=8.8e−7) compared with the body CoM.

DISCUSSION

We sought to determine the effects of out-of-phase and in-phase sinusoidal perturbations at the trunk on gait stability. We hypothesized that the out-of-phase condition would have a stabilizing effect through decreased LDE and a constant MoS maintained with a reduced CoM and BoS. We also expected that the in-phase condition would have a destabilizing effect on gait, measured through increased LDE and requiring a larger BoS for the larger CoM excursions to maintain a constant MoS. Instead, we found that the LDE was decreased for both the in-phase and out-of-phase conditions, indicating that the motion of the CoM was less chaotic for conditions where pendular motion was allowed. Although step width was consistent across all conditions, the

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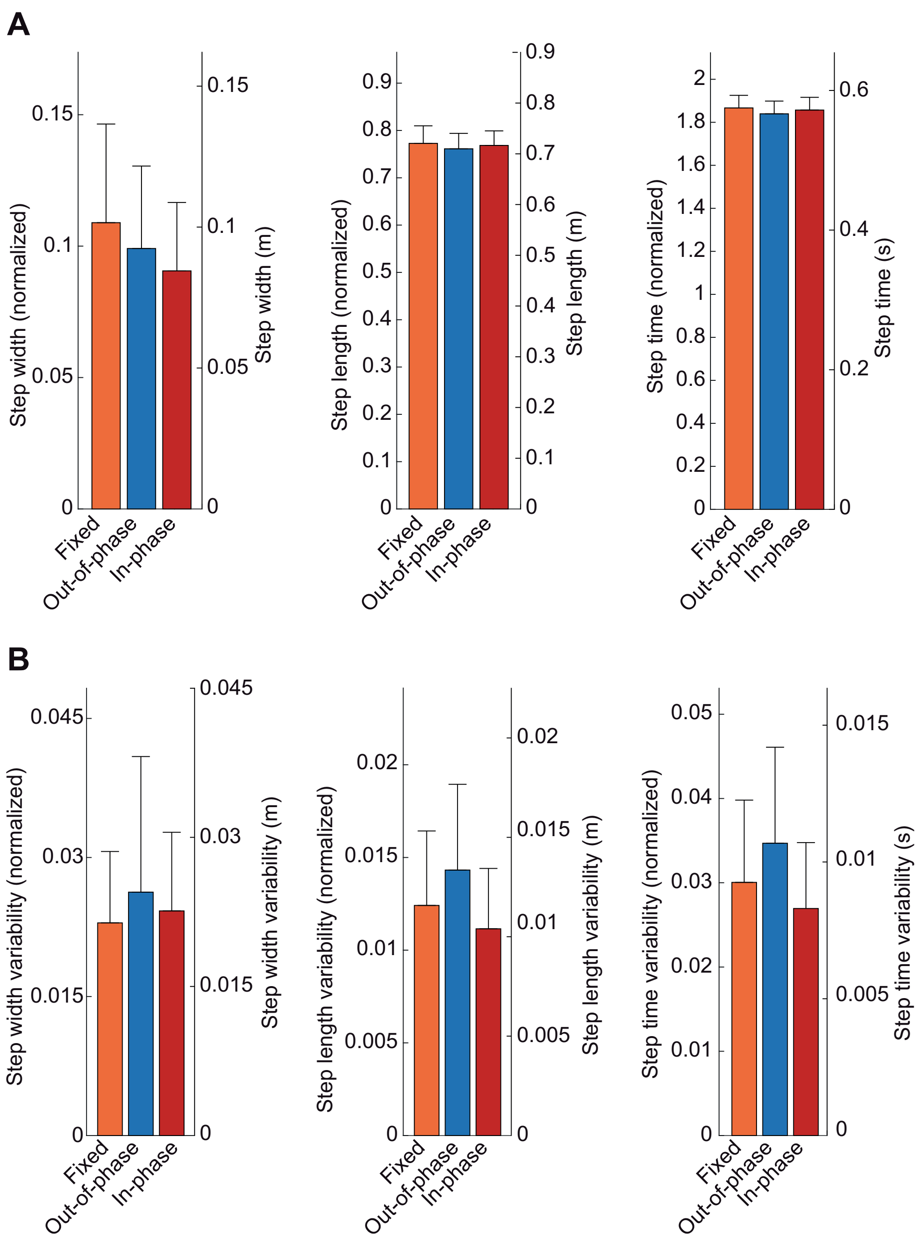

Fig. 5. Spatiotemporal measures for each condition. (A) Mean and standard deviation of the step width, length and time. (B) Mean variability of the same gait parameters (represented using the average standard deviation) and standard deviation of the variability. The left axes report the normalized values, and the right axes display the values with SI units. Bars are the averages across all subjects (N=12), and error bars denote 1 s.d.

MoS was larger for the out-of-phase condition and unchanged for the in-phase condition.

Based on the decreased step width found in our previous study (Martin and Li, 2018), we had predicted that the pendular position effect of the device would govern its effect on gait stability, leading to a destabilizing in-phase condition and stabilizing out-of-phase condition. Instead, we found a decreased LDE for the in-phase condition, which indicates that the impact of the pendular torque created a reduction in the fluctuations of the CoM, increasing dynamic stability throughout the trial through torques that restored the body CoM towards upright. Conversely, the increased MoS observed in the out-of-phase condition suggests that the pendular mass shifted the total CoM position to a more neutral position, creating a larger margin. The more neutral position of the CoM may have led to the lower LDE observed for the out-of-phase condition. Therefore, the LDE measure seemed to capture the pendular torque impact of the oscillating mass, and the MoS captured the positional impact. The double inverted pendulum model (Fig. 2) proposed did not include the effect of the restoring torques applied by the device due to the attachment of multiple physical springs. If the restoring torque is greater than the pendulum torque, then stabilizing and destabilizing effects would be similar to the positional torque effects.

In this study, gait stability was defined using two different measures, which provided differing results regarding the effects of each condition. The LDE quantified the chaotic behaviour of the system by examining fluctuations in the state of the system throughout the entire trial. The MoS measure of stability investigated BoS–CoM modifications to compensate for the perturbation. The increased MoS observed in the out-of-phase condition suggests that the more lateral foot placement relative to the CoM provided greater stability. Although the LDE was lower than for the fixed condition, it was still higher than that of the in-phase condition, suggesting that the out-of-phase condition was less stable than the in-phase condition. The behaviour of the device, a passive system, could provide a possible explanation for these conflicting

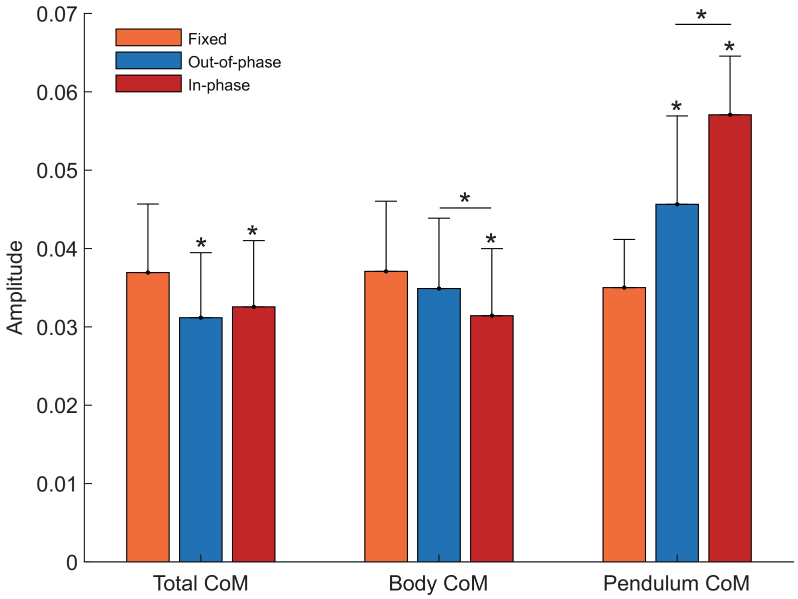

Fig. 6. Mean and standard deviation of the average amplitude of the motion of the total CoM, body CoM and pendulum CoM. The amplitude of the total CoM is significantly lower than in the fixed case for both the in-phase and out-of-phase conditions. The amplitude of the body CoM is significantly different when compared between the out-of-phase and in-phase condition. Bars are the averages across all subjects (N=12), and error bars denote 1 s.d. The asterisk over the individual bars indicates that the condition is statistically different from the fixed condition, and the asterisk over the horizontal line indicates that the out-of-phase and in-phase condition are statistically different (*P<0.05).

results. The distance from the body CoM to the pendulum CoM at the peak deviation of the body CoM from the centre line was more variable for the out-of-phase condition than for the other two conditions (denoted as the distance d in Fig. 3A). The out-of-phase variability (measured using the standard deviation of the distance from the CoM to the pendular mass) was approximately 6 times larger than that for the fixed condition (P=2.5e−9) and twice as large as that of the in-phase condition (P=8e−8). As a result of the increased variability, it is possible that the subjects were not able to predict the behaviour of the device, limiting their ability to compensate for the perturbation. The increased difficulty could lead to a more variable motion of the body and total CoM during the trials, increasing the LDE as compared with the in-phase condition. The increase in metabolic cost observed in the previous study with the device (Martin and Li, 2018), despite the decrease in step width, further suggests the need for increased active control (Donelan et al., 2004) to compensate for the perturbation from strategies other than stepping.

Unlike a previous study with the same device (Martin and Li, 2018), we did not observe a decrease in step width. This could be due to changes in experimental protocol for determining preferred step width. In the previous study, the preferred step width was recorded during the last 2 min of the 10 min treadmill acclimatization period whereas in the current study, step width was recorded during the fixed condition, which appeared in randomized order. The reduction in step

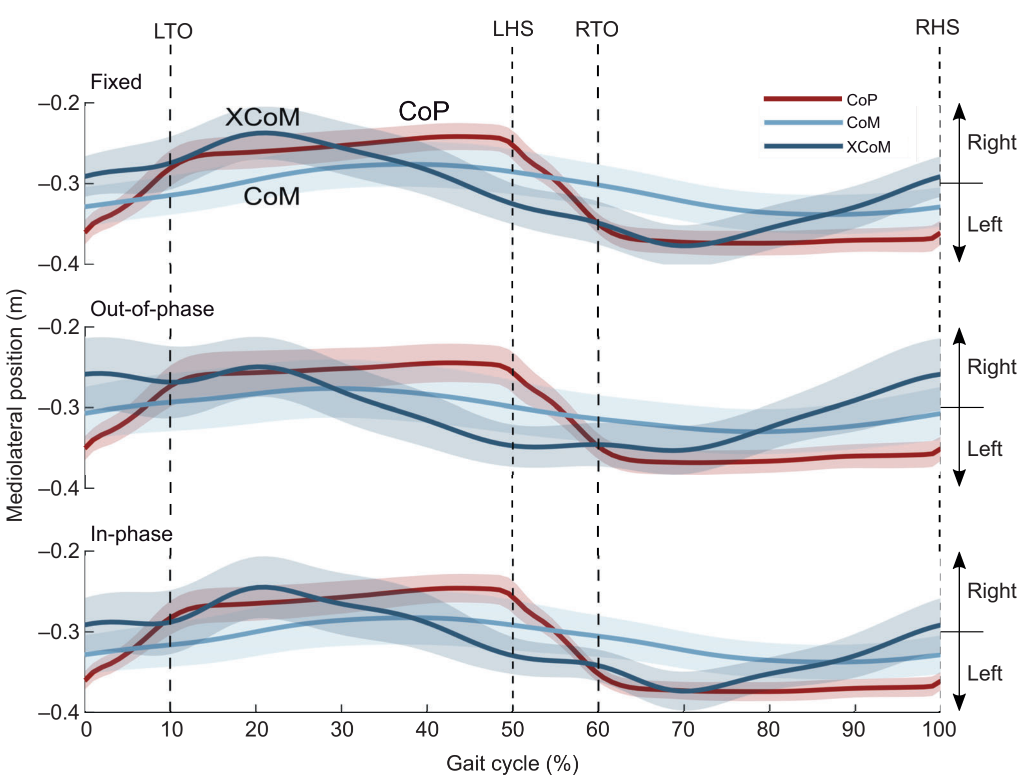

Fig. 7. Average mediolateral trajectories of the centre of pressure (CoP), CoM and XCoM. The XCoM during the out-of-phase condition is more medial to the CoP during single limb support [between left toe off (LTO) and left heel strike (LHS) and between right toe off (RTO) and right heel strike (RHS)] than for the other two conditions. The XCoM moves lateral to the BoS at approximately 20% of the gait cycle (for all conditions) and 70% of the gait cycle (for fixed and in-phase) for right and left stance, respectively. The shaded area represents 1 s.d. of the trajectories, and the opaque line represents the average trajectory (N=12).

width previously observed may have been the result of treadmill acclimatization rather than an effect of the device. For the in-phase condition, we had hypothesized that the step width would be increased as the device perturbs the total CoM in the lateral direction. However, with this device, an increase in step width could result in a larger amplitude of motion of the device (Martin and Li, 2018), which was not preferred.

To avoid larger perturbations, subjects elected to directly control their CoM rather than increase step width. Although step width was relatively constant between all conditions, the MoS was not constant for the out-of-phase condition. The altered MoS instead caused variation in CoM lateral position among the different conditions (Fig. 6) and was reduced for both the out-of-phase and in-phase conditions. The finding of an unaltered step width but reduced CoM motion suggests that subjects are actively controlling their CoM state to maintain stability. CoM control may be achieved by a reduction in the mediolateral motion of the pelvis through increased hip adductor moments (Maki and McIlroy, 1997) or of the trunk through increased co-contraction of back muscles. These types of behaviours could support our negative minimum MoS results as these control strategies provide stabilizing torques that are not reflected in the simple inverted pendulum model of walking that is the basis of MoS (Hof et al., 2005). If the total CoM has moved beyond the BoS provided by the stance foot, then strategies other than CoP modulation, such as upper body momentum, are required to stabilize the body (Herr and Popovic, 2008; Hof et al., 2005; Otten, 1999).

Contrary to previous studies that found mostly positive MoS values (Buurke et al., 2018; Hof et al., 2010, 2005), we found negative values for minimum MoS. Although the other control strategies previously suggested allow the possibility of a negative MoS, our simplified CoM calculation could have also contributed towards negative values. The estimation of the CoM position based on the markers on the frame of the backpack could deviate from the location of the actual CoM. The average trajectories of the XCoM and CoP (Fig. 7) show that during single support, the XCoM does move to a position that is lateral to the CoP. It is possible that relative motion between the backpack and trunk caused a lateral shift in the reported CoM position and velocity. However, the general MoS trend (reported in Fig. 4) is still present in the trajectories as the position of the XCoM relative to the BoS is more medial compared with that for the fixed and in-phase conditions.

There are some limitations of the current study that must be considered. Because of the passive nature of the device, the in-phase and out-of-phase conditions do not perfectly oscillate in-phase and out-of-phase. However, as seen in the sinusoidal patterns (Fig. 3A), there is a clear trend of in-phase and out-of-phase behaviour. Although we intended for the pendulum amplitudes of the in-phase and out-of-phase perturbations to be equal, they were not perfectly equal, which could be an additional source of variation between the two perturbations. Additionally, the pendular mass was not scaled to the subject's mass, potentially leading to reduced consequences for subjects with larger body masses. The backpack itself also covered a large portion of the subject's body, which led to a simplified estimate of the CoM and did not allow for extensive kinematic analysis to further explore the compensations made to maintain stability. As a result of the simplified estimate of the CoM, we could not obtain accurate estimates of the sagittal or vertical position of the CoM to calculate effective pendulum length. Instead, we used leg length as a simplified estimation of the pendulum length, similar to XCoM calculations from other studies (Buurke et al., 2018; Hof et al., 2005; Vlutters et al., 2016). During the trials, there was a 2 min acclimatization period to allow subjects to adapt to the device, in line with previous studies with similarly brief (Bruijn et al., 2009b; Donelan et al., 2004) to no (Liu and Lockhart, 2013; Walsh et al., 2018) acclimatization periods. Subjects were able to adapt to the device during this period, but it is unknown whether subjects would maintain the same gait behaviour with much longer acclimatization periods.

Regardless of the limitations of the study, the overall decreased CoM motion and unchanged step width suggest direct active control of the mediolateral motion of the CoM to maintain stability. Modification of upper body momentum, one of the main balance strategies, with an oscillating pendulum did not affect step width. Although the results presented are produced by a device that would not typically be worn outside the lab, they do have interesting implications for the control of mediolateral gait stability. The upper body momentum strategy suggested by the results is advantageous when foot placement is restricted, such as in narrow pathways or crowded areas, or less beneficial (as it was with our device). Insights into control strategies to counter trunk perturbations can be applied to improve the stability of elderly adults and individuals with chronic disorders (Bruijn et al., 2013), through, for example, exercise regimens (Bellew et al., 2005; Persch et al., 2009; Sato et al., 2015) and lower-limb assistive exoskeletons (Ugurlu et al., 2014). Balance strategies that actively control the CoM can be applied to exoskeleton controllers with upper-body stabilization and dynamic control algorithms to create dynamically stable gait in bipedal walking robots.

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