The effect of external lateral stabilization on the control of mediolateral stability in walking and running

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It is still unclear how humans control mediolateral (ML) stability in walking and even more so for running. Here, foot placement adjustment as a main mechanism of active control of mediolateral stability was compared between walking and running. Moreover, to verify the role of foot placement as a means of active control of ML stability and associated metabolic costs in both modes of locomotion, this study investigated the effect of external lateral stabilization on foot placement control. Ten young adults participated in this study. Kinematic data of the trunk (T_6) and feet (heels) as well as breath-by-breath oxygen consumption data were recorded during walking and running on a treadmill in normal and stabilized conditions. Coordination between ML trunk Center of Mass (CoM) state and subsequent ML foot placement, step width, and step width variability were assessed. Two-way repeated measures ANOVAs (either normal or SPM1d) were used to test for effects of walking vs. running and of normal vs. stabilized locomotion. We found a stronger association between ML trunk CoM state and foot placement in walking than in running from 90-100% of the gait cycle and also a higher step width variability in walking, but no significant differences in step width. The association between trunk CoM state and foot placement was significantly decreased by external lateral stabilization in walking and running, and this reduction was stronger in walking than in running from 75-100% of gait cycle. Surprisingly, energy cost significantly increased by external lateral stabilization, which was more pronounced in running than walking. We conclude that ML foot placement is coordinated to the CoM kinematic state to stabilize both walking and running. This coordination is more tight in walking than in running and appears not to contribute substantially to the energy costs of either mode of locomotion.

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Abstract

It is still unclear how humans control mediolateral (ML) stability in walking and even more so for running. Here, foot placement adjustment as a main mechanism of active control of mediolateral stability was compared between walking and running. Moreover, to verify the role of foot placement as a means of active control of ML stability and associated metabolic costs in both modes of locomotion, this study investigated the effect of external lateral stabilization on foot placement control. Ten young adults participated in this study. Kinematic data of the trunk (T_6) and feet (heels) as well as breath-by-breath oxygen consumption data were recorded during walking and running on a treadmill in normal and stabilized conditions. Coordination between ML trunk Center of Mass (CoM) state and subsequent ML foot placement, step width, and step width variability were assessed. Two-way repeated measures ANOVAs (either normal or SPM1d) were used to test for effects of walking vs. running and of normal vs. stabilized locomotion. We found a stronger association between ML trunk CoM state and foot placement in walking than in running from 90-100% of the gait cycle and also a higher step width variability in walking, but no significant differences in step width. The association between trunk CoM state and foot placement was significantly decreased by external lateral stabilization in walking and running, and this reduction was stronger in walking than in running from 75-100% of gait cycle. Surprisingly, energy cost significantly increased by external lateral stabilization, which was more pronounced in running than walking. We conclude that ML foot placement is coordinated to the CoM kinematic state to stabilize both walking and running. This coordination is more tight in walking than in running and appears not to contribute substantially to the energy costs of either mode of locomotion.

Keywords: foot placement strategy**,** balance, gait stability, walking, running.

1. Introduction

It is still unclear how humans walk and run with such ease, that is, stable and with low energy costs. Gait stability requires control of the Center of Mass (CoM) relative to the Base of Support (BoS) [1-3]. During walking and running, motions of the CoM relative to the BoS are thought to be controlled by passive dynamics as well as active processes [1-3]. Small perturbations may be controlled by passive dynamics without Central Nervous System (CNS) involvement, and larger instabilities in the system are countered by active control, which requires sensing of perturbations, generating appropriate motor commands, and producing compensatory motions [1].

Foot placement adjustment is the main mechanism of active control of Medio-lateral (ML) stability in walking and running [4-7]. External lateral stabilization by means of a spring-like construction reduced ML CoM movement [8] and this coincided with a 24–60% reduction in step width in walking [8-10] and 30-45% and 12.3% reductions in step width variability in walking [9, 10] and running [3], respectively. The coupling between CoM movements and step width is reciprocal, i.e. constraining CoM kinematics leads to adjustments of foot placement, but constraining foot placement also leads to adjustments of CoM kinematics [11, 12]. This coupling between CoM displacement and foot placement is reflected in correlations of the CoM position and velocity during the swing phase with the subsequent foot placement during walking [13-15].

Modulation of step width in response to variations in CoM movement is an active process [16, 17]. In line with this, it has been reported that external lateral stabilization may decrease energy costs [3, 9, 10, 18], by 3-6% in walking [9, 18] and 2% in running [3]. However, not all studies found a (significant) decrease in energy cost due to external lateral stabilization [8, 19, 20].

Although modulation of foot placement is important for control of gait stability, to date, we do not fully understand the mechanisms underlying the control of stability of walking and even less of running. It has been shown that humans run with step widths close to zero [4]. A step width near zero may imply that there is a lower need for active control of ML stability in running. In line with this, McClay and Cavanagh [21] demonstrated that humans run by placing the foot along the middle of the body, which aligns the vertical ground reaction forces close to the CoM, minimizes the ML ground reaction forces on the body from step-to-step, and minimizes the moment generated about the AP axis [22]. Thus, most of the CoM displacement is directed forward and ML motion is relatively small [22]. Decreasing ML CoM motion may be a strategy for control of stability during running, and if this is the case, the effect of external lateral stabilization on ML displacement of CoM, step width adjustment, correlation of preceding CoM state with the subsequent foot placement [13], and energy cost of active control of ML stability will be lower in running than in walking. In the current study, we set out to test the idea that running poses less challenge to ML stability than walking.

The effect of speed on the stability of walking has been investigated in several studies [23-26], however there is a lack of information on running. Most relevant for our present focus, the coordination between ML CoM state and foot placement was not influenced by walking speeds between 3.6 – 5.04 km/h [13, 15], but it was affected by speeds between 0.72 – 3.6 km/h, with less tight coupling at lower speeds [15]. In this study, we intend to test the idea that the speed influences coordination between ML trunk CoM state and subsequent ML foot placement, step width, step width variability, and energy costs required for control of ML stability in running.

We hypothesized that (1) walking and running are stabilized by active control in the frontal plane as reflected in correlation of the ML CoM position and velocity during the swing phase with the subsequent ML foot placement in these two modes of locomotion. (2) Foot placement strategy is more critical in walking than in running as reflected in a higher aforementioned correlation in walking. We further hypothesized that (3) external lateral stabilization decreases active control of ML stability in both modes of locomotion, as reflected by a reduction in the correlation between CoM state and subsequent foot placement, alongside a decrease in step width, step width variability and energy costs. Since we expect more active control of lateral stability in walking than in running, we hypothesized that (4) the reduction in aforementioned parameters is stronger in walking than in running. Finally, we hypothesized that (5) running speed influences these parameters.

2. Method

2.1. Participants

After signing the informed consent, a convenience sample of 10 young (6 men, 4 women) participants (age: 27.70±4.78 years, mass: 73.80±8.57 kg, and height: 181.30±6.57 cm) participated in this study, which had been approved by the local ethics committee of the Faculty of Behavioral and Movement Sciences of the Vrije Universiteit, Amsterdam. Exclusion criteria were: lower extremity injuries, history of surgery in the lower extremity, as well as any kind of impairments, medications, and infectious diseases which might affect walking mechanics or energy consumption. All of these exclusion criteria were self-reported by participants. Participants were asked to refrain from strenuous activity the day before experiments and to refrain from using coffee and alcohol on the day of the experiment.

2.2. Experimental protocol

Participants visited the laboratory during one session and they were measured during walking and running on a motorized treadmill in two (normal, stabilized) conditions. The participants were familiarized with walking and running on the treadmill in each condition, and they were instructed not to resist the spring forces of the stabilization frame [10]. Familiarization for each mode and each condition took about 2 minutes. Data collection started 10 minutes after the end of the familiarization protocol.

For each participant, first the conditions (normal and stabilized) were randomized and then speeds (walking at 4.5 km/h and running at 7.5, 9.0 and 10.5 km/h) were randomized within each condition. Participants completed 8 trials, each trail with a duration of 5 min. Trials were separated by a resting period of approximately 5 min.

2.3. Experimental set-up

A light-weight frame (mass = 1.5 Kg) was used for the external lateral stabilization condition, it was attached through a belt around the waist. Two sliders on both sides allowed participants to rotate their pelvis relative to the frame in the transverse plane, with minimal friction. Two stiff ropes attached to the frame on either side, joined each other at 0.5 m from the frame, providing space for free arm swing. From this junction, springs with spring stiffness of approximately 1260 N m^{-1} were attached to a slider on a vertical rail, which in turn was connected to two horizontal rails placed at the height of the pelvis of the participant. Thus, the set-up did not restrict movement in vertical and AP directions, nor rotations about the vertical axis, and transverse spring forces acted approximately at the level of the CoM during walking and running trials (Fig. 1).

Fig 1. goes here

2.4. Instruments

Kinematic, kinetic, and breath-by-breath oxygen consumption data during walking and running trials were obtained from an Optotrak motion analysis system (Northern Digital Inc, Ontario, Canada), sampled at 100 samples/s, from force plates embedded in the treadmill (ForceLink b.v., Culemborg, the Netherlands), sampled at 1000 samples/s, and from a pulmonary gas exchange system (Cosmed K4b^2, Cosmed, Italy), respectively. Clusters of three infrared markers were attached to the thorax (over the T_6 spinous process) and the heels.

2.5. Data processing

Ground reaction force data were filtered with a 10 Hz cut-off frequency (2nd order, bidirectional Butterworth digital filter). Heel strikes were calculated from center of pressure data [ref to Roerdink paper]. Kinematic data from the Optotrak system were not filtered.

The trunk accounts for almost two-thirds of a person's body mass and the effect of its motion on active control of gait stability has been shown by a strong relationship between step-by- step variation in ML trunk CoM kinematics and step width during walking [14]. The mean of the three infrared markers was used to approximate the ML trunk CoM position. The ML trunk CoM velocity was calculated as the first derivative of the ML marker cluster position time series. The heel markers were used to determine ML foot placement position. Next, these data were separated into gait cycles, which were time normalized to 0-100%. For each step, the ML position of the current stance foot was defined as the origin, and the position of the next stance foot and trunk were expressed relative to this origin. To further simplify the modeling (i.e. making sure that no offset was needed), all relevant variables (the position of the next stance foot and trunk, and trunk velocity), were zero-centered by subtracting the mean for each percentage of the gait cycle.

Next, a model predicting foot placement was developed. This model was created for each trial and it links the ML foot placement at heel strike to trunk CoM position and velocity during the preceding stride [13]:

$$\mathbf{FP} = \boldsymbol{\beta}\mathbf{1}(\mathbf{i}) \cdot \mathbf{CoM}(\mathbf{i}) + \boldsymbol{\beta}\mathbf{2}(\mathbf{i}) \cdot \mathbf{VCoM}(\mathbf{i}) + \boldsymbol{\varepsilon}(\mathbf{i})$$

with β1 and β2 being the regression coefficients, ε the error, and i the indicator of the % of gait cycle that was used for the prediction. The R^2 (i.e the strength of the relationship) between model prediction and actual ML foot placement was calculated as the primary outcome.

Mean and variability of step width were calculated for each trial. Step width was defined as the mean of the ML distances between heel markers during successive instances of initial contact, and step width variability was defined as the standard deviation thereof.

To evaluate the effect of external lateral stabilization on energy cost, oxygen uptake ($\dot{V}O_2$; ml min^{-1}) and respiratory exchange ratio (RER) were determined with the pulmonary gas exchange system during the last minute of each trial. We calculated gross metabolic energy expenditure (E_{gross}; J min^{-1}) as [27]:

$$\mathbf{E_{gross} = (4.940 \cdot RER + 16.40) \cdot \dot{V}O_2}$$

Resting metabolic rate, determined with the same method as we did for gross metabolic rate during seated position for 5 min prior to the trials, was subtracted from gross metabolic rate to calculate net energy expenditure during walking and running. To calculate net energy cost (EC; J kg^{-1} m^{-1}), net energy expenditure was divided by body mass (kg) and speed (m min^{-1}).

2.6. Statistical analysis

First, because our results indicated significant, but only very small differences between legs, and between running speeds (see supplementary figures, hypothesis 5), we calculated the average R^2 over legs and over running speeds. Next a vector analysis of the R^2 time series data was conducted using the SPM method [28]. The F-test was used to test our hypotheses; specifically, in line with our second hypothesis, we first tested for the difference in R^2 between normal walking and running. Next, we used a full factorial model to test the effects of lateral stabilization (hypothesis 3), and the difference in this effect between walking and running (i.e. mode of locomotion X condition interaction, hypothesis 4). The output of SPM provides an F-value for each sample of the R^2 time series, and the threshold corresponding to α set at 0.05. The values of F above the threshold (shaded areas in Fig. 3. and Fig 5. A, B, and C) indicate significant effects in the corresponding portion of the time series. Because our results indicated no significant effect of speed on step width, step width variability, and energy cost in running, we calculated the average of them over running speeds (see supplementary figures, hypothesis 5). Next, again in line with our second hypothesis, first, we tested the difference between normal walking and running. To test our third and fourth hypothesis, a two-way repeated analysis of variance with conditions (normal vs stabilized) and model of locomotion (walking vs running) as within-subject factors was conducted to evaluate the effect of external lateral stabilization and mode of locomotion on step width, step width variability, and energy cost. Level of significance for all statistical analyses was set at $p<0.05$.

3. Results

In line with our first hypothesis, the ML trunk CoM state and subsequent ML foot placement were highly correlated with R^2 ranging between ~0.6-0.8 from 75-100% of the gait cycle in walking and between ~0.6-0.7 from 60-100% of the gait cycle in running (Fig. 2). In line with our second hypothesis, we found a stronger relationship in walking than in running from 90-100% of the gait cycle (Fig. 3). No significant differences were found in step width between walking and running ($p= 0.101$ *and* $F (1, 9) = 3.33$) (Fig. 4. A). However, step width variability was significantly higher in walking than in running ($p= 0.042$ *and* $F (1, 9) = 5.59$) (Fig. 4. B).

Fig 2., 3., and 4. go here

In line with our third hypothesis, external lateral stabilization significantly decreased R^2 to ~0.2-0.5 during 50-100% of the gait cycle in walking and running (Fig. 2 and Fig. 5. A). In addition, step width and step width variability were significantly reduced by external lateral stabilization condition in walking and running (both $p \leq 0.001$, $F (1, 9) = 26.96$ for step width and $F (1, 9) = 106.06$ for step width variability) (Fig. 4. A and B).

Fig 5. goes here

In line with our fourth hypothesis, the effect of external lateral stabilization on R^2 was larger in walking than in running from 75-100% of the gait cycle (interaction effect, Fig. 5. C). In addition, the effect of external lateral stabilization on step width and step width variability was larger in walking than in running ($p = 0.003$ and $F (1, 9) = 16.02$ for step width as well as $p = 0.001$ and $F (1, 9) = 22.04$ for step width variability) (interaction effects, Fig. 4. A and B).

As expected, the energy cost was significantly higher in running than walking ($p \leq 0.001$ and $F (1, 9) =225.15$). However, in contrast with expectations, energy costs were significantly higher in the stabilized conditions ($p=0.039$ and $F (1, 9) =5.80$) and the increase in energy costs with external lateral stabilization was more pronounced in running than in walking ($p=0.028$ and $F (1, 9) =6.84$) (interaction effect, Fig. 4. C).

4. Discussion

Our results demonstrated a strong coupling between ML trunk CoM state in the swing phase of gait and the subsequent ML foot placement during both walking and running. ML trunk CoM position and velocity explained over 60% of the variance in ML foot placement in both modes of locomotion. Our hypothesis that foot placement adjustment as an active control mechanism of ML stability is more critical in walking than in running, was supported by a stronger correlation between the trunk CoM state at the end of the gait cycle, and a higher step width variability in walking than in running. Furthermore, our hypothesis that external lateral stabilization significantly decreases coordination of foot placement to the trunk CoM state, was also supported for both modes of locomotion. This hypothesis was also supported by significant reduction in step width and step width variability in the stabilized condition compared to the normal condition, but not by energy cost. The hypothesis that foot placement adjustment as an active control mechanism of ML stability is more critical in walking than in running was supported by stronger reductions in the coordination between ML trunk CoM state and subsequent ML foot placement, and in step width, and step width variability in stabilized walking than in stabilized running.

Our results confirmed that ML foot placement is coordinated to ML trunk CoM dynamics to control ML stability in walking. Similar to previous studies, which reported that 50-84% of ML foot placement variance can be explained by ML trunk, ML pelvis, and whole body CoM state during walking [13-15], our results indicated high predictive ability of the trunk CoM state on subsequent ML foot placement, with R^2 ranging between 60-80% during the last 25% of the gait cycle in walking. As our results also indicated high correlation between CoM state and subsequent ML foot placement (R^2 = 60-70%) during 60-100% of the gait cycle in running, we extended this to this mode of locomotion. The high predictive ability of trunk kinematic state in walking and running is likely to be due to active control of ML stability through foot placement, but could also be due to passive dynamic coupling of lower extremity movements to movements of the upper body. Active control of ML stability through foot placement is supported by studies on external lateral stabilization [8-10]. More evidence for the idea of active control of ML foot placement during walking comes from studies using mechanical perturbations [16, 29] and vibration [17] on this coupling.

In comparison to walking, our results indicated that control of ML foot placement is less tight in running. It has been suggested that the active control of subsequent foot placement begins earlier in the control of ML stability of *walking* when less time is available to complete the step (i.e. during walking at higher speeds) [15, 29]. However, the more pronounced reduction in step duration in running could limit the possibility of using foot placement to control gait stability. If this is the case, one step after a deviation of the CoM state might not be enough to restore stability and more consecutive steps might be required to stabilize the CoM in running. Additionally, during running an absorption strategy, allowed by flexion in the lower limb, during the stance phase may be used to control the CoM trajectory, which may limit the need for accurate foot placement.

It has been reported that external lateral stabilization decreases ML displacement of the CoM [8], accompanied by a 24–60% reduction in step width in walking [8-10] and 30-45% in step width variability in walking [9, 10]. Consistent with these studies, our results indicate that external lateral stabilization decreased the active control of ML stability in walking, as reflected by a reduction in coordination of ML foot placement to CoM state, alongside a reduction in step width and step width variability during stabilized walking. The results of the current study also indicate that external lateral stabilization decreases the control of ML stability in running, although less so than for walking, in line with a smaller decrease in step width variability of about 12% with external stabilization reported previously [3]. This smaller decrease may suggest that subjects need more foot placement control during stabilized running than during stabilized walking. This would appear to contradict the notion that the foot placement strategy is less important during normal running than normal walking. However, there may be several alternative explanations. First of all, the external lateral stabilization may have different effects on the ML stability in running and walking; it may be less effective during running, as the ML forces may affect body movements differently during the flight phase in running compared to the single leg stance phase in walking. In single leg stance, the spring forces and ground reaction forces on the stance leg may produce a rotational couple, which does not occur during the flight phase in running. It could be that this rotational component is key to stabilizing subjects. Thus, the stabilizing effect may be different between walking and running. A second explanation, may be that subjects do not experience the frame as sufficiently stabilizing in running and thus do not “offload” control to the frame as much as they do in walking. However, participants were familiarized with all conditions, and did not express feelings of discomfort during any of the conditions.

It has been reported that the active control of ML foot placement in walking, which is reflected by a correlation between the ML CoM state during the swing phase and the subsequent ML foot placement, is not affected by walking speed between 3.6-5.4 km/h [13, 15]. We extended this to running and our results showed that the active control of ML foot placement in running is affected in a minor way by speeds ranging between 7.5-10.5 km/h.

We measured energy to assess costs of active stability control. Reduced energy costs in stabilized walking would support that ML stabilization is an active process and differential effects between walking and running might indicate differences in these costs between these modes of locomotion. However, previous studies reported conflicting results of the effects of external lateral stabilization on energy costs. Some studies reported significant reductions [9, 18] and reported no effects [8, 10, 19, 20]. The current study even showed an increased energy cost in the stabilized condition. Our results showed that foot placement is used to control ML stability in walking and running. The energy cost of this strategy appears to be low, as the decrease in the use of the foot placement strategy during stabilized walking and running did not lead to decreases in energy costs. Instead, energy costs slightly increased. Unintended effects of the external stabilization, e.g. on propulsion may have outweighed the benefits. Low energy costs may explain why foot placement is likely to be preferred over other stability control strategies, such as control through ankle moments [6, 7, 30].

5. Conclusion

ML trunk CoM state explained over 60% of the variance in ML foot placement in walking and running. This suggests that ML foot placement is adjusted to ML trunk CoM dynamics to control ML stability in walking and running. The control of ML foot placement was stronger in walking than in running as reflected by a higher correlation between CoM state and subsequent foot placement in the former mode of locomotion. External lateral stabilization decreased this correlation, step width, and step width variability in both walking and running, with stronger reductions during the former. This may imply that there is a higher need for active control of ML stability via foot placement in walking. The correlation between ML trunk CoM state and subsequent foot placement was influenced to a negligible extent by speed in running.

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Figure captions

Fig 1. (A) Schematic representation of the experimental set up. Inset **(B)** shows the stabilization in more detail. (1) frame; (2) springs; (3) height-adjustable horizontal rail; (4) ball-bearing trolley freely moving in anterior-posterior direction; (5) slider freely moving in vertical direction; (6) vertical rail; and (7) rope attached to frame.

Fig. 2. The ability of ML trunk CoM state to predict subsequent foot placement (R^2) during normal and stabilized conditions in walking and running.

Fig. 3. The effect of mode of locomotion (walking & running) in normal condition on R^2.

Fig. 4. Effect of condition and movement patterns on step width (**A**), step width variability (**B**), and energy cost (**C**).

Fig. 5. A. The effect of condition (normal & stabilized) on R^2. **B.** The effect of mode of locomotion (walking & running) in both conditions (normal & stabilized) on R^2 **C.** The interaction effect (condition x mode of locomotion) on R^2.

Figure 1

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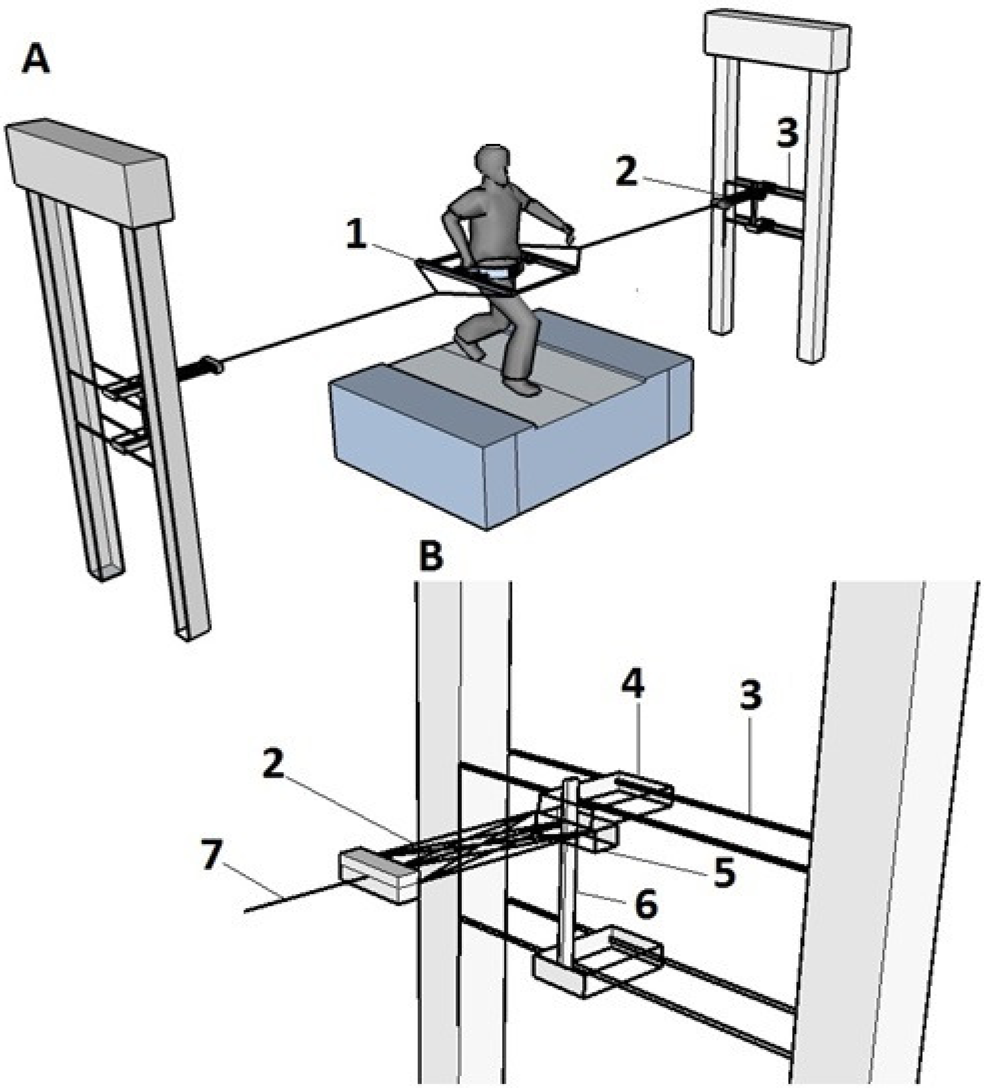

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Figure 2

Fig. 2. The ability of ML trunk CoM state to predict subsequent foot placement (R^2) during normal and stabilized conditions in walking and running.

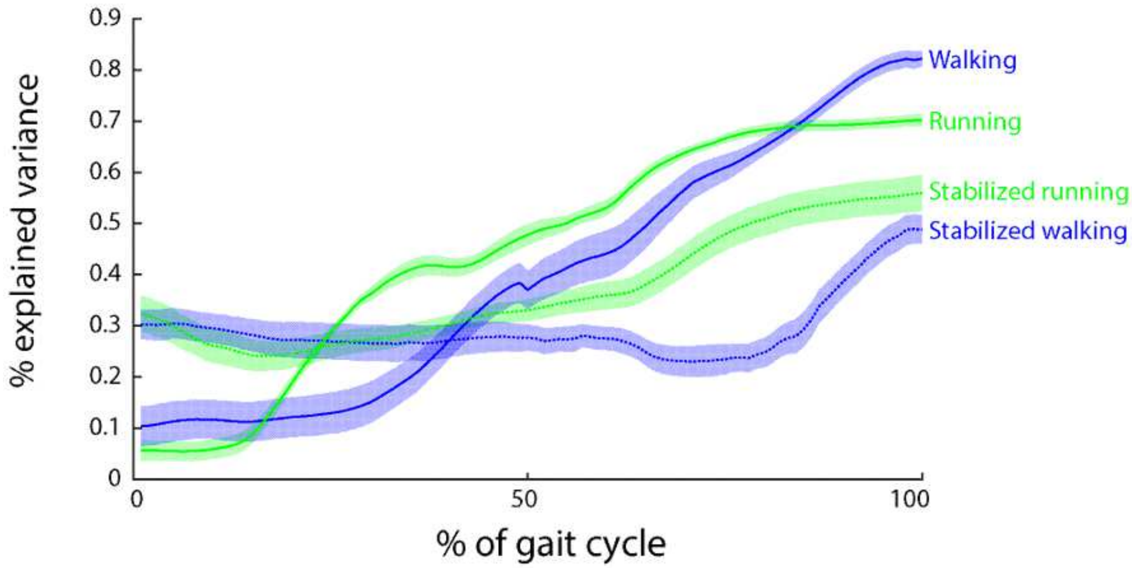

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Figure 3

Fig. 3. The effect of mode of locomotion (walking & running) in normal condition on R^2.

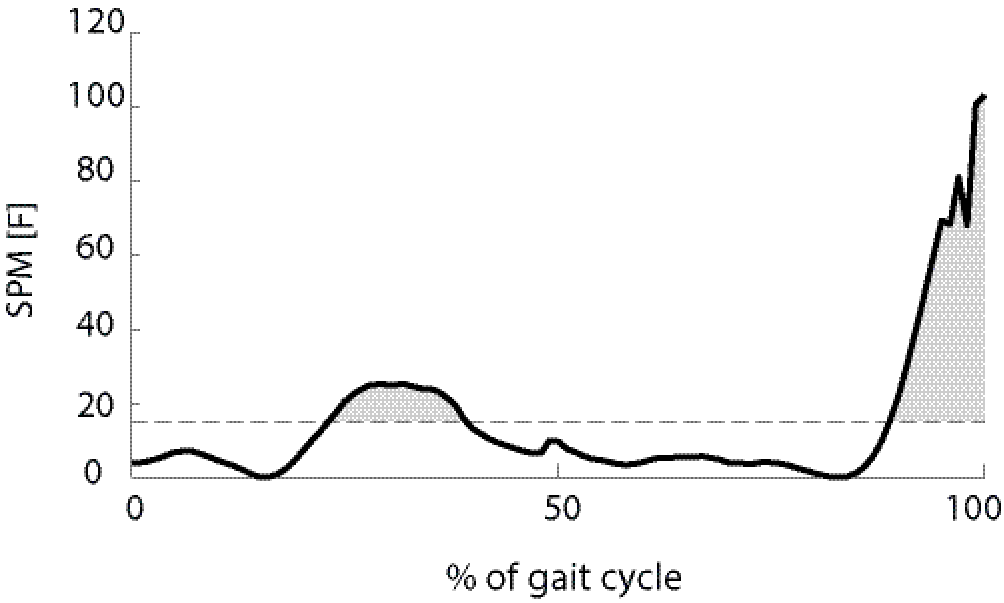

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Figure 4

Fig. 4. Effect of condition and movement patterns on step width (A), step width variability (B), and energy cost (C).

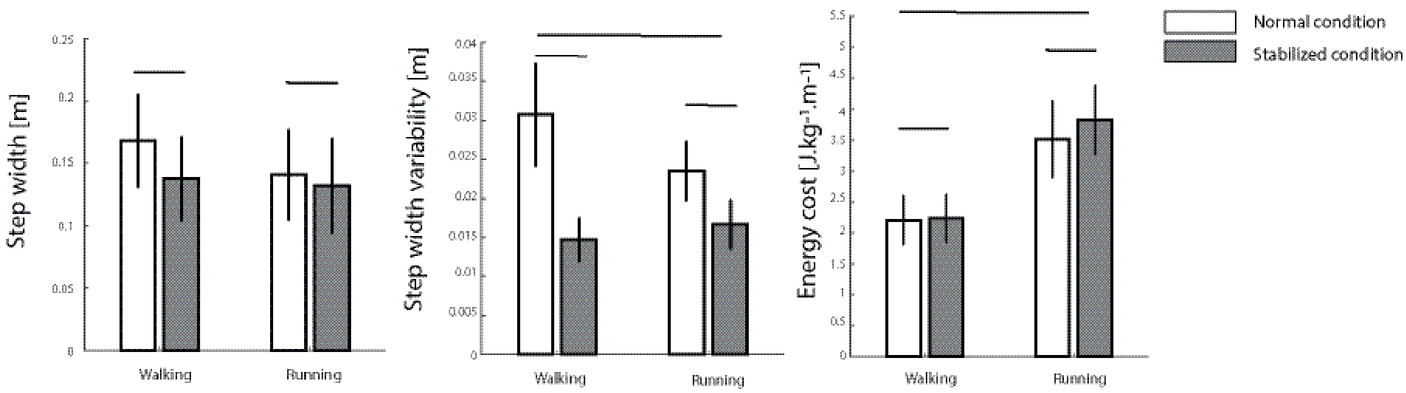

Figure 5

Fig. 5. A. The effect of condition (normal & stabilized) on R^2. B. The effect of mode of locomotion (walking & running) in both conditions (normal & stabilized) on R^2 C. The interaction effect (condition x mode of locomotion) on R[sup]