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Reduced center of pressure modulation elicits foot placement adjustments, but no additional trunk motion during anteroposterior-perturbed walking

M. Vlutters *, E.H.F. van Asseldonk, H. van der Kooij

Department of Biomechanical Engineering, University of Twente, The Netherlands

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A B S T R A C T

Understanding balance during human gait is complicated by the abundance of recovery options. Among all possible recovery options, three main strategies are often considered for human balance control, being the ankle, hip, and foot placement strategies. All can be addressed when balance is threatened during walking, but their relative importance remains uncertain. We have previously shown that healthy human subjects did not significantly adjust their foot placement relative to the body's center of mass (COM) in the first recovery step following anteroposterior pelvis perturbations, as compared to unperturbed walking. An ankle strategy could have contributed to the recovery instead.

Here the goal is to further elucidate balance strategy preferences by investigating the stepping and hip strategies following these anteroposterior perturbations, but with an ankle strategy made ineffective. This was achieved by physically blocking each ankle and minimizing the support area of each foot through a pair of modified ankle-foot orthoses. These "pin-shoes" enabled stilt-like walking and ensured that foot placement adjustment was the only way to modulate the center of pressure location, comparable to "footless" inverted pendulum models of walking.

Despite the pin-shoes, subjects did not additionally address a hip strategy compared to normal walking, but relied on foot placement adjustments instead. The observed foot placement adjustments were furthermore in line with concepts derived from a linear inverted pendulum model of walking. These results suggest low hip strategy priority when a foot placement strategy is available, while the latter can be predicted with concepts derived from a simple walking model.

1. Introduction

Balance during walking can be maintained in numerous ways, given the abundance of motor control options. For example, the body's center of mass (COM) motion can be influenced by foot placement adjustments ('stepping strategy') (MacKinnon and Winter, 1993), ankle moment modulation ('ankle strategy') (Kim and Collins, 2015; Pijnappels et al., 2005) and hip moment modulation ('hip strategy') (Horak, 1987). In the latter case, hip moments either oppose or aid gravity in angular accelerations of the trunk. The net trunk angular acceleration, together with the trunk's inertial properties, leads to sheer-forces under the foot or feet on the floor which affect horizontal COM motion.

Foot placement adjustment is often considered the most important strategy during walking, especially in the frontal plane (MacKinnon and Winter, 1993). In the sagittal plane, however, we recently found that healthy subjects did not significantly adjust their foot placement relative to the COM in the first step following anteroposterior (AP) pelvis perturbations applied at toe-off (Vlutters et al., 2016). Instead, subjects modulated subsequent gait phase durations, and the center of pressure (COP) location within the new base of support (BoS) after foot placement. The COM velocity at heel strike after the perturbation was furthermore a strong predictor of the COP location at the subsequent toe-off. This relation between COM velocity and COP location was in line with the extrapolated center of mass (XCOM) concept that can be derived from a linear inverted pendulum model (Hof, 2008).

In addition to unexpected perturbations, another way to investigate compensatory actions during walking is by changing the walking dynamics (Selinger et al., 2015; Thielemans et al., 2014), and inferring to what purpose a compensation occurs. This way,

* Corresponding author at: Universiteit Twente, Horstring, W-119, Postbus 217, 7500 AE Enschede, The Netherlands.

E-mail address: m.vlutters@utwente.nl (M. Vlutters).

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underlying objectives of balance control during walking it might be extracted. Fully eliminating a balance recovery strategy might lead to more pronounced effects in other strategies. A clear example is provided in (Horak, 1987), where subjects utilize a hip strategy when the availability of both the ankle and stepping strategies become severely limited when standing on a narrow ridge.

In this study, we examine balance strategy preferences by investigating whether subjects adjust the stepping and hip strategies in the recovery from AP perturbations during walking, while the ankle strategy is made ineffective. By eliminating the ankle strategy, the stepping and hip strategies might need to be additionally addressed as a compensation. Conversely, if in such a challenging condition a strategy is not additionally addressed while it is an available recovery option, it is reasonable to assume that this strategy has low priority in normal walking as well. This can provide insight in strategy preferences. In the presented experiment, the ankle strategy is made ineffective by physically blocking each ankle joint and reducing the BoS of each foot through a pair of 'pin-shoes'. These ensure a gait that resembles stilt-walking. Second, we investigate whether under such conditions the stepping strategy will be in line with predictions based on the XCOM concept. Due to the pin-shoes the COP can no longer modulate within the BoS of each foot. This way a stepping strategy becomes the only way to realize COP modulation in response to variations in COM velocity, in line with (Hof, 2008). Such comparisons might further validate concepts derived from simple inverted pendulum models of walking, and aid in the development of controllers for balance assisting devices such as lower-extremity exoskeletons.

2. Materials and methods

2.1. Apparatus

Set-up details can be found in (Vlutters et al., 2016). Briefly, an instrumented treadmill (MotekForce Link, Culemborg, Netherlands) and a motor (Moog, Nieuw-Vennep, Netherlands) at the rear of the treadmill were used to perturb subjects forward and backward during walking. The motor could be connected to the subject's pelvis through a rod and a brace (Distrac Wellcare, Hoegaarden, Belgium). Additionally, in this study a pair of ankle-foot orthoses (Airselect Elite, Aircast Inc., Summit NJ, USA) were used to restrain ankle movement, and reduce the area of the BoS through a modification, see Fig. 1. An aluminum pin was attached perpendicular to the sole, with a solid, aluminum hemisphere located at the end of the pin to ensure a minimal BoS. The weight of one pin-shoe was approximately 1.5 kg.

Fig. 1. Schematic drawing of one pin-shoe: an ankle-foot orthosis that blocks the ankle joint, modified to strongly reduce the support area of the foot. The vertical pin can be positioned along the foot.

2.2. Protocol

The experimental protocol was approved by the local ethics committee. All participants gave written informed consent prior to the experiment, in accordance with the Declaration of Helsinki. Ten subjects (5 men, age 20 ± 1 years, weight 71 ± 7 kg, height 1.76 ± 0.08 m, mean ± s.d.) walked on the treadmill while wearing the pin-shoes. For each subject, the pins were fixated at a position approximately halfway the longitudinal arches of each foot. We used this position based on a pilot study, in which subjects found it difficult to walk with the pin shifted too far toward the toes or toward the heel, making them either fall backward, or make the knee prone to buckling during the stance phase.

Subjects were given time to adapt to the pin-shoes until they could walk with their arms crossed over the abdomen. Walking speed was $0.63 * \sqrt{l}\ \mathrm{ms}^{-1}$, where $\sqrt{l}$ is the square root of the total leg length, including pin length, following (Hof, 1996). This speed corresponds to the 'slow' walking speed in (Vlutters et al., 2016), and is considered reasonable for walking on the pin-shoes. Subjects randomly received 150 ms forward (+) and backward (−) pelvis perturbations at the instance of toe-off right (TOR) using the motor behind the treadmill. Perturbation magnitude was 4, 8, 12, or 16% of the subject's body weight. Each perturbation condition was repeated 8 times, yielding 64 perturbations per subject.

Kinematic data of landmarks on the feet, lower legs, upper legs, pelvis, trunk, head, and the pin-shoes were collected at 100 Hz using a motion capture system (Visualeyez II, Phoenix Technologies Inc, Vancouver, Canada). The landmark on each shoe was the pin-tip. Ground reaction forces and moments were collected at 1000 Hz.

2.3. Data processing

For comparison, previously collected data in perturbed walking without pin-shoes at a similar speed (Vlutters et al., 2016) were also processed as described below. In each data set, with and without pin-shoes, the trunk data of one subject was removed from analysis due to motion artifacts of the trunk cluster. Landmark data were used to estimate the total-body COM position (Dumas et al., 2007). The COM position was differentiated, and the belt velocity was added for a COM velocity relative to the walking surface. Trunk kinematic data were used to find sagittal plane trunk angular accelerations and angular excursions. Trunk angular accelerations were used as an indication of the hip strategy, containing both passive (gravitational) and active (human) contributions. Trunk angular excursions indicate the changes in trunk angle relative to that at the instance of perturbations onset.

Instances of TOR, heel strike right (HSR), toe-off left (TOL) and heel strike left (HSL) were detected using vertical ground reaction forces (threshold 20 N). All data were cut, and re-sampled to five 50-sample sequences spanning one gait cycle: (1) TOR (perturbation onset) to TOR + 150 ms (perturbation end), (2) TOR + 150 ms to HSR, (3) HSR to TOL, (4) TOL to HSL, (5) HSL to TOR. Sequence duration was calculated before re-sampling. Sequences and their corresponding durations were sorted, and averaged over the 8 repetitions per perturbation type to obtain repetition-average sequences per subject.

As the pin-shoes change the way of walking, even without perturbations, a direct comparison of the perturbation effects between walking with and without pin-shoes can be difficult. We therefore quantified the effects of the perturbations on the COM velocity, trunk angular acceleration, and trunk angular excursion data relative to the unperturbed case, for both the with and without pin-shoes conditions. For each subject, the repetition-average unperturbed sequences were subtracted from all repetition-average perturbed sequences. This makes the data relative to the baseline in both the with and without pin-shoes conditions. These differences with the baseline were integrated for a 'deviation area'. The integration time step size was obtained by dividing the repetition-average duration of each perturbed sequence by its 50 samples, therefore yielding a different time step size for every perturbed condition. The total integral for all conditions spanned from the end of the perturbation to time t_A = 1.108 s, which was the longest subject-average gait cycle duration among all perturbed conditions with pin-shoes. The obtained areas quantify how much a subject deviates from unperturbed walking up to time t_A as a result of the perturbations, and can be compared between the with and without pin-shoes conditions. To obtain the deviation areas for the trunk angular excursion, we integrated over the absolute difference between perturbed and unperturbed conditions instead, as the integral of an angle over time has little physical meaning while positive and negative deviations can cancel out. Finally, the repetition-average data and deviation areas per subject were averaged across subjects for subject-averages and standard deviations.

Linear mixed models were used to assess the effect of the pin-shoes (fixed effect, with intercept), and its interaction with perturbation magnitude (fixed effect, with intercept) on the deviation areas. Tests were done separately for forward and backward perturbations. To account for correlated data resulting from repeated measures within a single subject, a random subject factor (intercept) was included. The significance level was set to 0.05, and a Bonferroni correction was applied during *post hoc* analysis to correct for multiple comparisons. In the *post hoc* analysis we focus on the effect of the pin-shoes on the areas, and do not mutually compare the area sizes for the different perturbation magnitudes and directions. Statistical analysis was performed with SPSS statistics 21 (IBM Corporation, Armonk, NY, USA).

To investigate COM-velocity effects on foot placement adjustments, linear least-squares fits were made to the subject-average data at previously investigated gait events (Vlutters et al., 2016), see Table 1. For comparison with the fits, an XCOM proportionality constant $\omega_0{}^{-1} = \sqrt{(l/g)}$ and an offset $b_{AP} = S_c/(e^{T_c\sqrt{(g/l)}} - 1)$ were calculated for each subject following (Hof, 2008), where l is the subject's leg length plus pin height, g is the Earth's gravitational constant, S_c is some desired step length, and T_c is some desired single support duration. In a linear inverted pendulum model of walking, proportionality constant $\omega_0{}^{-1}$ relates the model's horizontal COM velocity to a stepping location that terminates model movement (i.e. the XCOM). Each subject's $\omega_0{}^{-1}$ was used to find a subject-average $\omega_0{}^{-1}$, and a subject-average AP XCOM = $\omega_0{}^{-1} * \dot{x}$, where the XCOM is relative to the COM, and $\dot{x}$ corresponds with the horizontal AP COM velocity. Offset b_{AP} can be used in 'constant offset control' proposed in (Hof, 2008), to make a linear inverted pendulum model converge toward a desired gait with a step length S_c and single support duration T_c. Each subject's average step length (AP pin-pin distance at heel strike) and single support duration (time between toe-off and subsequent heel strike) during unperturbed pin-shoe walking were used to calculate this offset.

Table 1
Slope, intercept and coefficient of determination of the linear least squares (LLSQ) fit made to the subject-average data at specific instances of the gait cycle after the perturbation.

Dependent variable →	AP COM velocity, at HSR		
Independent variable ↓	Slope	Intercept	R^2
AP distance right pin – COM, at HSR	0.421	−0.148	0.926
AP distance right pin – COM, at TOL	0.409	−0.188	0.986
AP distance COP – COM, at TOL	0.325	−0.158	0.986

3. Results

3.1. COM velocity

During unperturbed walking, the step frequency was higher in subjects wearing pin-shoes (1.88 ± 0.11 Hz) than in subjects without pin-shoes (1.40 ± 0.08 Hz) at similar speeds. The perturbations lead to additional movements that temporary deviate from the unperturbed case. The deviation areas quantify the degree to which subjects deviated from the unperturbed walking condition within the chosen time window, for both the with and without pin-shoes conditions.

The pin-shoes affect the COM velocity the most for backward perturbations. Backward perturbed subjects with pin-shoes showed significantly more deviations from the unperturbed walking velocity than subjects without pin-shoes ($F_{1,18}$ = 146.200, p < .001), as reflected by the deviation areas in Fig. 2I. These areas were furthermore significantly affected by the interaction between pin-shoes and perturbation magnitude ($F_{3,54}$ = 18.639, p < .001). Post hoc comparison showed that the areas were significantly larger (more negative) in the pin-shoes condition for all backward perturbations (p ≤ .019). In contrast, for forward perturbations there was no significant effect of the pin-shoes ($F_{1,18}$ = 1.951, p = .179), unless when interacting with perturbation magnitude ($F_{3,54}$ = 3.368, p = .025). Only for the two smaller magnitude forward perturbations were the deviation areas significantly larger in the pin-shoes condition (p ≤ .040).

3.2. Trunk response

During the perturbations, the trunk initially obtains a backward (+) angular acceleration for forward perturbations and a forward (−) angular acceleration for backward perturbations, attributed to a passive response. Subsequently, subjects must actively accelerate the trunk in the opposite direction to prevent it from toppling over. Although in terms of magnitude these trunk accelerations tend to deviate more from those in unperturbed walking in the pin-shoe condition, the sequence within which these deviations occur (P.End – HSR) is generally also of shorter duration in the pin-shoe condition. Integrating the angular acceleration deviations leads to deviation areas which are significantly smaller in the pin-shoe condition for backward perturbations ($F_{1,16}$ = 12.108, p = .003), but no different for forward perturbations ($F_{1,16}$ = 0.025, p = .876). Neither for forward nor for backward perturbations there was an interaction effect between pin-shoes and perturbation magnitude (p > .658). Similarly, the pin-shoes had no significant effects on the trunk angular excursion areas for both forward ($F_{1,16}$ = 0.006, p = .940) and backward ($F_{1,16}$ = 0.865, p = .366) perturbations, nor where there any interaction effects with perturbation magnitude ($F_{3,48}$ ≤ 2.048, p ≥ .120). Overall, the results suggests that COM velocity is affected by the pin-shoes for various perturbations, while the trunk response remains similar between both conditions, or even tends to decrease in the pin-shoe condition.

3.3. Foot placement

Following perturbations at TOR, subjects made the first recovery step with the right leg. On average, subjects increased the AP distance between the COM and the tip of the right pin-shoe with increasing forward COM velocity at HSR, and decreased it with decreasing COM velocity, eventually placing the pin behind the COM. This foot placement modulation did not occur in subjects that walked without pin-shoes (Vlutters et al., 2016). Following contact, subjects shifted their weight to the newly placed foot,

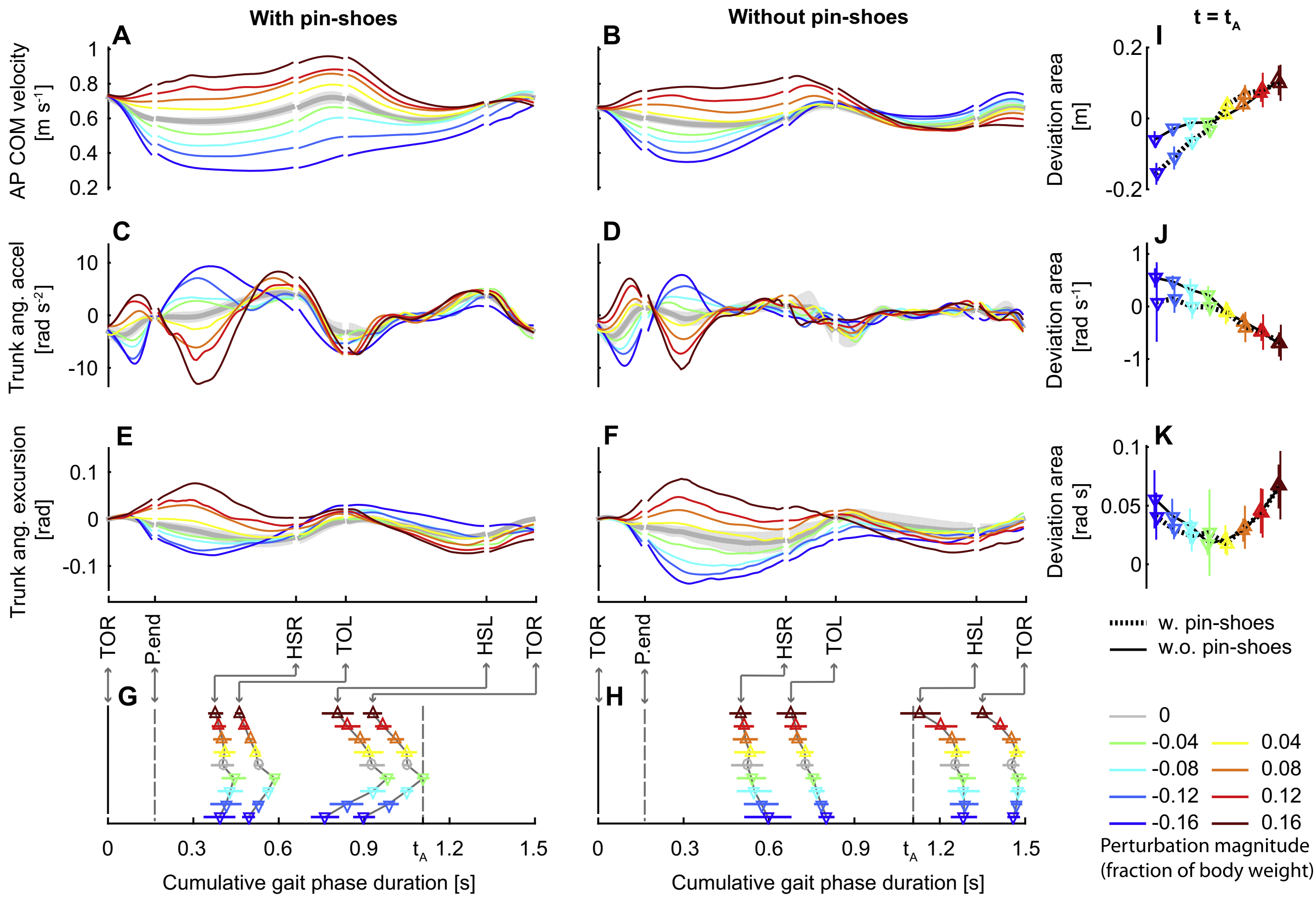

Fig. 2. Subject-average responses to the perturbations during walking, with (A, C, E, G) and without (B, D, F, H) pin-shoes over one gait cycle. (A, B) AP COM velocity. (C, D) Sagittal-plane trunk angular acceleration. (E, F) Sagittal-plane trunk angular excursion. Positive corresponds with backward trunk angular accelerations and excursions. (G, H) Duration from perturbation onset at TOR to subsequent gait events. Horizontal bars indicate the subject standard deviation. Vertical dashed lines indicate the integration time window for the deviation areas. (I, J, K) Deviation area for COM velocity, trunk angular acceleration, and trunk angular excursion per perturbation. Vertical bars indicate the subject standard deviation. Dotted line corresponds to data with pin-shoes, solid line to data without pin-shoes. Colors indicate the perturbation magnitudes as fraction of the subject's body weight. In (A-F) standard deviations (shaded area) of the perturbed conditions are not shown to prevent image cluttering.

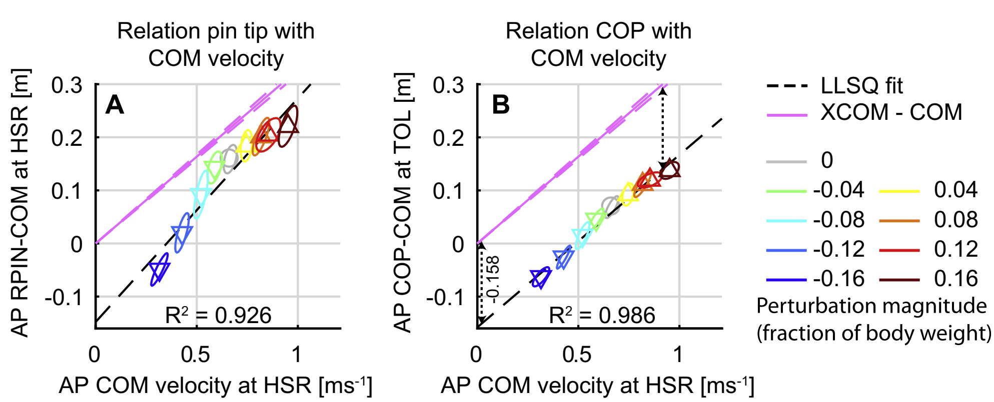

Fig. 3. Subject-average foot placement responses. (A) AP COM velocity at HSR against the distance between the COM and the right pin-tip at HSR. (B) AP COM velocity at HSR against the distance between the COM and the COP at TOL. Triangles show subject averages and indicate the perturbation direction. Ellipses represent the subject standard deviation. Black dashed line is a linear least square fit to the data. R^2 indicates the coefficient of determination of the fit. Pink solid line has slope ω_0^{-1} and no intercept, and indicates the subject average of the XCOM relative to the COM. Pink dashed lines indicate the corresponding standard deviation. The two vertical dotted arrows have the same length, and indicate the magnitude of the intercept of the fit. The intercept corresponds to a constant offset between the fit and the pink XCOM line, assuming the slopes of both lines are equal. Colors indicate the perturbation magnitudes as fraction of the subject's body weight. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

resulting in a COP location at TOL that is beneath the right pin. The relation of the COM velocity at HSR with both the distance between the pin and the COM at HSR, and with the distance between the COP and the COM at TOL, can be quantitatively explained with a linear relation, see Table 1 and Fig. 3. The slope of the latter relation in Table 1 is about equal to the subject-average ${\omega_0}^{-1}$ (0.327 ± 0.008, mean ± std). Under the condition that the slope of this fit is equal to that of the XCOM line in Fig. 3 (i.e. ${\omega_0}^{-1}$), the intercept of the fit provides a constant offset between the COP and the XCOM of −0.158 m. For comparison, the offset b_{AP} calculated according to (Hof, 2008) was −0.129 ± 0.021 m, using the step length (0.311 ± 0.04) and single support duration (0.402 ± 0.036) during unperturbed pin-shoe walking. From the perspective of a linear inverted pendulum model this negative offset would make the pendulum topple over the foot, which is indeed required in the sagittal plane to continue walking forward.

4. Discussion

We investigated balance recovery responses to AP pelvis perturbations during walking with constrained COP modulation within the feet to examine preferences in balance recovery strategies, and to compare the foot placement strategy with concepts derived from a simple inverted pendulum model. The results suggest that the trunk is not additionally addressed despite the more challenging pin-shoe condition. Foot placement adjustments are used in the recovery instead, in line with the XCOM concept.

4.1. Velocity recovery

Step frequency was higher in subjects with pin-shoes, possibly due to the inability to deliver a push-off, and to reduce the collision costs at heel strike (Kuo, 2002). This higher frequency could partially explain why it takes more steps to counteract the perturbation-induced velocity changes. Based on the deviation areas, deviations from the unperturbed walking velocity up to t_A tend to be of larger magnitude for subjects with pin-shoes. This is presumably caused by the inability to deliver an ankle moment. There is no intrinsic stiffness or damping about the point of rotation with the floor that affects the passive response to the perturbation, nor can an active moment be generated to influence body movement during single support. Hence, in the pin-shoe conditions the perturbations tend to cause larger deviations in COM velocity, while at the same time the ankle strategy is ineffective in counteracting these deviations. Two alternative recovery options are trunk motion and foot placement adjustment.

4.2. Trunk response

The presented results suggest that a hip strategy has a low priority in perturbation recovery when foot placement adjustments are possible. Maintaining an upright upper body appears to be an important objective even under destabilizing conditions. Subjects cannot instantly react to the perturbations, hence the trunk response during the perturbation likely contains passive effects. The lower peak accelerations during the perturbations for subjects with pin-shoes compared to those without pin-shoes suggest an increased trunk impedance prior to perturbation onset. This possibly results from proactive trunk stabilization (Winter, 1995) in response to wearing the pin-shoes. Trunk accelerations directly after the perturbations must contain active contributions from hip moments. These both help the trunk return toward its unperturbed state, and initially help mitigate the perturbation effects on the COM velocity through inertial effects. While it is difficult to discriminate between these two purposes, it appears that subjects try to prevent deviations from the unperturbed trunk state, rather than using a hip strategy to influence COM motion. This is because subjects with pin-shoes do not show increased trunk angular acceleration or angular excursion areas, even though they cannot use an ankle strategy to counteract the perturbations. Similarly, findings in (Oddsson et al., 2004) suggest that humans tend to stabilize the trunk, and mainly use foot placement to recover from diagonal support surface perturbations. If instead the perturbation would have been given briefly before foot contact, there is little time to correct through foot placement adjustments (Hof et al., 2010). In such a scenario subjects might be more inclined to use the trunk for maintaining balance, or transfer the recovery to the subsequent step. The trunk findings therefore hold under the condition that at least foot placement adjustment is an available recovery option.

4.3. Velocity-related foot adjustment

Contrary to our previous findings in subjects without pin-shoes (Vlutters et al., 2016), subjects with pin-shoes actively adjusted their AP foot placement relative to the COM in the first recovery step after the perturbations. Still, linear relations previously observed between the AP COM velocity at HSR and the AP COM-COP distance at TOL remain intact. Without pin-shoes this can be partially realized through ankle moment modulation, which provides a mechanism to displace the COP within the area of the foot. With pin-shoes, COP displacement is only possible through foot placement adjustments. Hence, a similar relation is observed through strategies that differ between the with and without pin-shoes conditions.

The observed foot placement adjustments resemble velocity-based strategies previously applied in low-dimensional models of walking (Hof, 2008) and running (Peuker et al., 2012). The slope of the aforementioned relation is about equal to ω_0^{-1}, which is the proportionality constant between COM velocity and the XCOM (Hof, 2008). Furthermore, the intercept of the fit provides a (near) constant offset from the XCOM, which is comparable to the calculated offset b_{AP}. Both of these findings support the use of the linear inverted pendulum model and constant offset control for predictions in the first recovery step.

It must be noted, however, that there are some important differences between human and model. In humans, it takes until the end of the double support phase for the COP to shift toward the leading foot, which is required for the relation presented in Fig. 3b. In the linear inverted pendulum model, a double support phase does not occur and the COP shifts instantaneously at heel strike. To calculate b_{AP} we used the average unperturbed single support duration, which does not include the time that is required to shift the COP forward during double support. If this time would be included, T_c increases and b_{AP} would decrease to deviate more from the intercept presented in Table 1. Furthermore, b_{AP} is only meaningful in constant offset control proposed in (Hof, 2008) if the single support time is kept constant, while humans will modulate this time (Fig. 2G, H). The model therefore does not provide a complete explanation of the data.

4.4. Implications

The presented findings have implications for the design and control of assistive devices, such as lower-extremity exoskeletons and prostheses. Humans might adopt different balance recovery strategies if no ankle control is present, such as when using passive prostheses, or in gait with impaired control of the distal joints. In unilateral amputees, shifts in AP COP were significantly lower in the prosthetic limb compared to the intact limb in six different walking conditions (Kendell et al., 2010). The inclusion of ankle actuation could increase the similarities with an intact limb, improve balance control during walking (Kim and Collins, 2015), and possibly result in balance control that is more similar to that of a healthy human. When considering an able-bodied user of a lower-extremity exoskeleton, such as a rescue or construction worker, a conflict might occur between the natural response of the user and the control of the device if the device has not the same degrees of freedom as its user. As demonstrated in this work, balance control is dependent on the constraints on the human body. Having such an exoskeleton assist in maintaining balance therefore requires at least human-like capabilities of the device, without impeding the user.

5. Conclusions

Foot placement adjustments are essential for perturbation recovery when ankle contributions are ineffective. Considering that a hip strategy was available but not additionally addressed in a challenging pin-shoes condition, we might conclude that this strategy has low priority during walking in general when foot placement adjustments are an option. Hence, humans prefer to keep their trunk upright when facing a balance threat.

By introducing the pin-shoes we furthermore showed that an earlier presented relation between the COM velocity and the COP is still established by the subjects, but now with the help of foot placement adjustments rather than with ankle moment modulation. This relation was in line with a concept derived from a linear inverted pendulum model, and therefore motivates the use of low dimensional models as a basis for mimicking concepts of reactive balance during human walking.

Conflict of interest statement

The authors declare that they have no conflicts of interest.

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