Journal of Biomechanics 136 (2022) 111073

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech

The effect of constraining mediolateral ankle moments and foot placement on the use of the counter-rotation mechanism during walking

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A R T I C L E I N F O

Keywords:
Gait stability
Constrained walking
Center of mass acceleration
Foot placement
Ankle moments
Counter-rotation mechanism

A B S T R A C T

During walking, the center of mass (CoM) position can be controlled relative to the base of support by shifts of the center of pressure through modulation of foot placement and ankle moments (CoP-mechanism). An additional mechanism is the counter-rotation mechanism, i.e. changing the angular momentum of segments around the CoM to change the direction of the ground reaction force. It is unknown if, and how, humans use the counter-rotation mechanism to accelerate the CoM during walking and how this interacts with the CoP-mechanism. Thirteen healthy adults walked on a treadmill, while full-body kinematic and force plate data were obtained. The contributions of the CoP and the counter-rotation mechanisms to CoM-acceleration during steady-state walking, walking on LesSchuh (i.e. constraining mediolateral CoP shifts underneath the stance foot) and walking on LesSchuh at 50% of normal step width, constraining both foot placement and ankle mechanisms (LesSchuh50%) were calculated. The within-stride variance in CoM-acceleration due to the CoP-mechanism was smaller and the within-stride variance in CoM-acceleration due to the counter-rotation mechanism was larger during LesSchuh50% compared to steady-state walking. This suggests that the counter-rotation mechanism is used to stabilize gait when needed, but the CoP-mechanism was the main contributor to the total CoM-acceleration. The use of the counter-rotation mechanism may be limited, because angular accelerations ultimately need to be reversed and because of interference with other task constraints, such as head stabilization and preventing interference with the gait pattern.

1. Introduction

Stable gait, defined as gait that does not lead to falls (Bruijn et al., 2013), requires control of the state of the body center of mass(CoM) relative to the base of support(BoS), i.e. the area within an outline of all points on the body in contact with the support surface, or vice versa. In gait, the BoS, at any point in time, is formed by the parts of the feet that are in contact with the floor (Bruijn et al., 2018). Perturbations of the CoM state and actions to control the CoM state relative to the BoS are reflected in accelerations of the CoM. The most extensively studied mechanism to accelerate the CoM is foot placement (Bauby et al., 2000; Bruijn et al., 2018; Rankin et al., 2014; Wang et al., 2014). A second mechanism to accelerate the CoM is the application of moments around the ankle of the stance foot(i.e. ankle mechanism) (Horak et al., 1986). These ankle moments are reflected in a shift of the center of pressure of the ground reaction force(CoP). Foot placement primarily moves the BoS to accommodate the state of the CoM, but also constrains the location of the CoP during the subsequent single-leg stance phase. So, foot placement and ankle moments combined determine the resulting position of the CoP relative to the CoM, which in turn determines CoM-acceleration. Shifts of the CoP through modulation of foot placement and ankle moments will here be referred to as the CoP-mechanism. An additional mechanism to accelerate the CoM, is the counter-rotation mechanism, i.e. changing the angular momentum of segments around

Abbreviations: CoM, center of mass; BoS, base of support; ML, mediolateral; CoP, center of pressure; *m*, body mass; $CoM_{vertical}$, vertical position of the CoM; CoM_{ML}, ML position of the CoM; *t*, time; $\ddot{CoM}_{ML}$, double derivative of; *g*, gravitational acceleration; CoP_{ML}, ML position of the CoP; $\dot{H}_{fr}$, change in total body angular momentum in the frontal plane; F_z, vertical ground reaction force.

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https://doi.org/10.1016/j.jbiomech.2022.111073
Accepted 29 March 2022
Available online 1 April 2022

the CoM to change the direction of the ground reaction force (Hof, 2007; Hof et al., 2007; Otten, 1999). A leftward rotational acceleration of the trunk, for example, results in a rightward acceleration of the CoM and vice versa. Accelerations of other body segments, for example the arms or head can be used in the same way.

The main mechanism to accelerate the CoM in the frontal plane has been suggested to be foot placement, with a smaller role for the ankle mechanism (Bauby et al., 2000; MacKinnon et al., 1993; van Leeuwen et al., 2021). Theoretically, the counter-rotation mechanism may also play a role, but research on its role is limited. Previously, we investigated the relative contribution of the CoP and counter-rotation mechanism to CoM-acceleration in the anteroposterior direction during a normal step and during the first recovery step after a perturbation (van den Bogaart et al., 2020). A limited use of the counter-rotation mechanism to accelerate the CoM was found, likely because using the counter-rotation mechanism would have interfered with the gait pattern (van den Bogaart et al., 2020). The current study addressed the role of the counter-rotation mechanism for mediolateral stability.

When aiming to improve gait stability, it is important to unravel the use of the CoP and counter-rotation mechanisms and their interplay. Simply studying unconstrained walking may not be sufficient, as during normal walking, the foot placement mechanism is dominant (Bauby et al., 2000; MacKinnon et al., 1993; van Leeuwen et al., 2021). Constrained or perturbed walking might provide insights in the use of the CoP and counter-rotation mechanisms and their interplay. Previous research has investigated the interplay between the two components of the CoP-mechanism by constraining one of them. Constrained foot placement to a fixed step width did not lead to ankle moment adjustments (van Leeuwen et al., 2021). Therefore, it is likely that the counter-rotation mechanism played a role, but this was not investigated. Conversely, constraining ankle moments, e.g. by narrowing the surface area underneath the shoe or by walking on prosthetic legs, led to wider foot placement (Hof et al., 2007; van Leeuwen et al., 2020), but did not lead to a significant increase in the range of the frontal plane angular momentum(<15%) (D'Andrea et al., 2014; Miller et al., 2018; Pickle et al., 2014; Sheehan et al., 2015; Silverman et al., 2011). Assuming similar timing of the minima and maxima of the angular momentum, increased(decreased) angular momentum corresponds to an increased (decreased) rate of change of the angular momentum, and therefore suggests increased(decreased) use of the counter-rotation mechanism. Thus, most likely, the counter-rotation mechanism did not, or only minimally compensated for limited ankle moments in people with a prosthetic leg. These studies determined the mean range of the angular momentum over all strides. However, the within-stride and stride-to-stride variance in CoM-acceleration due to the counter-rotation mechanism could still be increased due to reduced modulation of ankle moments. Constraints in the CoP-mechanism could be adjusted by such an increase in variation. Therefore, in this study, we determined the within-stride and stride-to-stride variance in CoM-acceleration caused by the counter-rotation mechanism.

To study if, and how, humans use a counter-rotation mechanism to accelerate the CoM during walking, it may be needed to provoke them to do so, e.g. by constraining foot placement in addition to constraining ankle moments. Studies that assessed the use of the counter-rotation mechanism after medial foot placement or platform perturbations when ankle moments were constrained, did find increased whole-body angular momentum (Miller et al., 2018; Sheehan et al., 2015). It thus seems that by constraining both components of the CoP-mechanism, the use of the counter-rotation mechanism can be provoked (Miller et al., 2018). While the authors interpreted this increased whole-body angular momentum as a sign of instability (Miller et al., 2018; Sheehan et al., 2015), it could be that the increased angular momentum is a sign of increased control of the CoM through the counter-rotation mechanism.

Overall, it should be noted that accelerating the CoM affects the location of the CoM relative to the BoS. However, CoM-accelerations (induced by the CoP or counter-rotation mechanism) due to control of the CoM or due to internal perturbations cannot be distinguished.

In the current study, we assessed the contribution of the CoP and counter-rotation mechanisms to CoM-acceleration in the mediolateral direction during steady-state walking, during walking with constrained ankle moments and during walking with constrained ankle moments and foot placement(i.e. constrained CoP-mechanism). We hypothesized that the mean contribution of the counter-rotation mechanism over strides would be similar during all conditions, because angular accelerations ultimately need to be reversed, leading to the opposite effect on CoM-acceleration. We also hypothesized limited use of the counter-rotation mechanism compared to the CoP-mechanism, because rotational accelerations of body parts may interfere with other task constraints, such as stabilizing the head and performing leg swing (van den Bogaart et al., 2020). Moreover, we hypothesized that the within-stride and stride-to-stride variance in CoM-acceleration due to the counter-rotation mechanism will increase when ankle moments are constrained, and even more so when also foot placement is constrained. This hypothesis is based on a previous study, showing an increased range of whole-body angular momentum when perturbing mediolateral foot placement in people with a prosthetic leg (Miller et al., 2018).

2. Methods

2.1. Subjects

Thirteen healthy adults(6 males, age 23.8 ± 3.7 years old, BMI 23.3 ± 2.2 kg/m^2) participated in this study. Sample size was calculated for a two-tailed paired sample *t*-test analysis using G*Power (1-β = 0.9, α = 0.05) and an effect size of 1.13 based on a previous study of Miller at al., 2018. This effect size was based on frontal plane whole-body angular momentum during medially perturbed(0.04 ± 0.0080) and unperturbed steps(0.032 ± 0.0025) and a drop-out rate of 10%.

Potential participants were excluded if they reported neurological or orthopedic disorder(s), had uncorrectable visual impairments, were unable to walk without difficulty for ≥ 45 min, had undergone surgery of the lower extremities during the last two years, or took medication that might affect their gait pattern. Participants gave written informed consent prior to the experiment. Ethical approval (VCWE-2020–202) was granted by the ethics review board of the faculty of Behavioral and Movement Sciences at 'Vrije Universiteit Amsterdam'.

2.2. Materials and software

Participants walked on an instrumented dual-belt treadmill (Motek-Forcelink, Amsterdam). Full-body kinematics were measured using two Optotrak-cameras (Northern Digital Inc, Waterloo Ontario) directed at the center of the treadmill (sampling rate:50 Hz). Clusters of 3 markers were attached to the shoes (feet), shanks, thighs, pelvis, trunk, upper arms, forearms and head. The force plate sampling rate was 200 Hz.

During the experiment, participants wore shoes to constrain

Fig. 1. LesSchuh, shoe constraining the mediolateral shift of the center of pressure (CoP) underneath the stance foot through the narrow (1 cm) ridge attached to the sole.

mediolateral CoP shifts underneath the stance foot through a narrow(1 cm) ridge attached to the sole (Fig. 1). The so-called LesSchuh limits ankle moments, while anteroposterior roll-off and subsequent push-off remain possible, because the ridge bends with the sole in anteroposterior direction.

2.3. Procedures

Participants walked on the left belt of an instrumented dual-belt treadmill (width:49.5 cm), in three conditions at a constant speed of 2.5 km/h. In view of the initial plan to also measure children and older adults, we chose a walking speed of 2.5 km/h as this speed was attainable for all age groups during all conditions. Participants were required to walk on one belt to prevent that the LesSchuh could get stuck in the gap between the two belts. Additionally, walking on two belts could result in a wider step as precaution (van Leeuwen et al., 2020). The first condition consisted of 10 min steady-state walking on normal shoes of the same type as the LesSchuh, but without the ridge. A trial duration of 10 min should ensure habituation to treadmill walking (Meyer et al., 2019). The second condition consisted of 15 min walking on the LesSchuh (Fig. 1) as walking on LesSchuh for 10 min was not always enough to reach a steady state (Hoogstad et al., 2022; van Leeuwen et al., 2020). Participants were asked to walk on the ridge, not touching the ground with the sides of the shoe's sole. Participants were also instructed to place their feet in a similar orientation as they would do without LesSchuh, to avoid a "toeing-out strategy" potentially inducing a mediolateral CoP shift after foot placement, despite the narrow base of support (Rebula et al., 2017). The third and final condition consisted of 5 min walking on the LesSchuh at 50% of the average step width as measured during the first condition, constraining both components of the CoP-mechanism(i.e., foot placement and ankle mechanisms;LesSchuh50%). LesSchuh50% for longer than 5 min was not feasible, as it is fatiguing and attention-demanding. The step width during the third condition was imposed by projecting beams on the treadmill, and participants were instructed to place their foot(ridge of the LesSchuh) in the middle of the beam. During all conditions, participants wore a safety harness connected to a rail at the ceiling, that did not provide weight support.

2.4. Data analysis

Kinematic and force plate data were low-pass filtered at 10 Hz using a second-order Butterworth filter, as recommended by Bisseling et al. 2006 (Bisseling et al., 2006). Kinematic data were analyzed using a 13-segment kinematic model. Full-body CoM was calculated by combining the CoM of all segments. CoP and full-body CoM data were high-pass filtered at 0.1 Hz using a second-order Butterworth filter, to correct a force plate offset error. Heel-strikes and toe-offs were determined based on CoP time series (Roerdink et al., 2008). One full gait cycle started at heel-strike of the left foot and ended at heel-strike of the same foot.

The contributions of the CoP and counter-rotation mechanism to CoM-acceleration in the frontal plane, were calculated using Eq. (1), as described by Hof (2007).

$$\ddot{CoM}_{ML}(t) = \frac{-F_z(CoP_{ML}(t) - CoM_{ML}(t))}{m \bullet CoM_{vertical}(t)} - \frac{\dot{H}_{fr}(t)}{m \bullet CoM_{vertical}(t)} \quad (1)$$

in which m is body mass, CoM_{ML} is mediolateral(ML) position of the CoM, $CoM_{vertical}$ is vertical position of the CoM, $\ddot{CoM}_{ML}$ is the double derivative of CoM_{ML} with respect to time, t is time, F_z is vertical ground reaction force, CoP_{ML} is ML position of the CoP, and $\dot{H}_{fr}$ is change in total body angular momentum (around the CoM) in the frontal plane. The first part of the right-hand term, $\frac{-F_z(CoP_{ML}(t)-CoM_{ML}(t))}{m \bullet CoM_{vertical}(t)}$, refers to ML CoM-acceleration induced by the CoP-mechanism, whereas the second part, $\frac{\dot{H}_{fr}(t)}{m \bullet CoM_{vertical}(t)}$, is ML CoM-acceleration induced by the counter-rotation mechanism.

For each individual stride, time normalized curves of the total CoM-acceleration and of the contributions of the CoP and counter-rotation mechanisms to the total CoM-acceleration in the frontal plane were calculated. For each variable, we calculated the averaged curve across all strides during each trial (Fig. 2).

We expressed within-stride variance in CoM-accelerations (due to the CoP and counter-rotation mechanism) by the standard deviation(SD) of the time series for each single-leg stance phase. Then we averaged across all single-leg stance phases during each trial, to obtain the average variance within single-leg stance per trial (Fig. 2). The within-stride variance in CoM-acceleration due to the CoP-mechanism can only be explained by the ankle mechanism, as foot placement largely determines the average CoP in stance and this measure only considers the variance around this average position.

We also expressed the stride-to-stride variance in CoM-acceleration (due to the CoP and counter-rotation mechanism) by the SD over the strides per time sample, after which the average SD was calculated per trial (Fig. 2).

To determine to role of foot placement to the contribution of the CoP-mechanism, we assessed the step width as the difference between the mediolateral CoP positions halfway during two subsequent single-leg stances and calculated the average step width per trial.

2.5. Statistics

Average time normalized curves of the total CoM-acceleration and of CoM-acceleration induced by the CoP and counter-rotation mechanism were compared between conditions using SPM repeated measures ANOVA(SPM1d vM.0.4.5, https://www.spm1D.org). Parametric statistical tests were used, as the D'Agostino-Pearson K2-test revealed that the values were normally distributed. If the main effect was significant, post-hoc SPM(t) maps were calculated. A Bonferroni correction was applied for comparisons within one variable.

One-way repeated measures ANOVAs were used to determine the effect of Condition on the within-stride and stride-to-stride variance in CoM-acceleration (due to the CoP and counter-rotation mechanism) and on the average step width. The data were normally distributed as tested with the Shapiro-Wilk test. Post-hoc analyses were performed to determine differences between the conditions (using a Bonferroni correction). Except the SPM analyses, statistical analyses were performed with SPSS (v25), and for all analysis we used $\alpha < 0.05$.

3. Results

On average participants took 414(SD = 29) strides during steady-state walking, 648(SD = 43) strides during LesSchuh and 225(SD = 32) strides during LesSchuh50%. Despite the different trial durations, the variances in outcomes did not differ significantly between the three conditions.

3.1. Total CoM-acceleration

The average time normalized curve of total CoM-acceleration over strides was similar during steady-state walking and walking on LesSchuh, but was decreased in magnitude around left and right heel-strike during LesSchuh50%, compared to steady-state walking and was increased in magnitude during the single-leg stance phases during LesSchuh compared to LesSchuh50% (Fig. 3). The variance in total within-stride CoM-acceleration did not differ significantly between conditions ($F(2,24) = 2.494$, $p = 0.104$;Fig. 4A). The stride-to-stride variance in total CoM-acceleration differed significantly between conditions($F(2,24) = 43.562$, $p < 0.001$). The stride-to-stride variance was larger during LesSchuh and LesSchuh50% compared to steady-state walking($p < 0.001$ and $p < 0.001$, respectively;Fig. 4B).

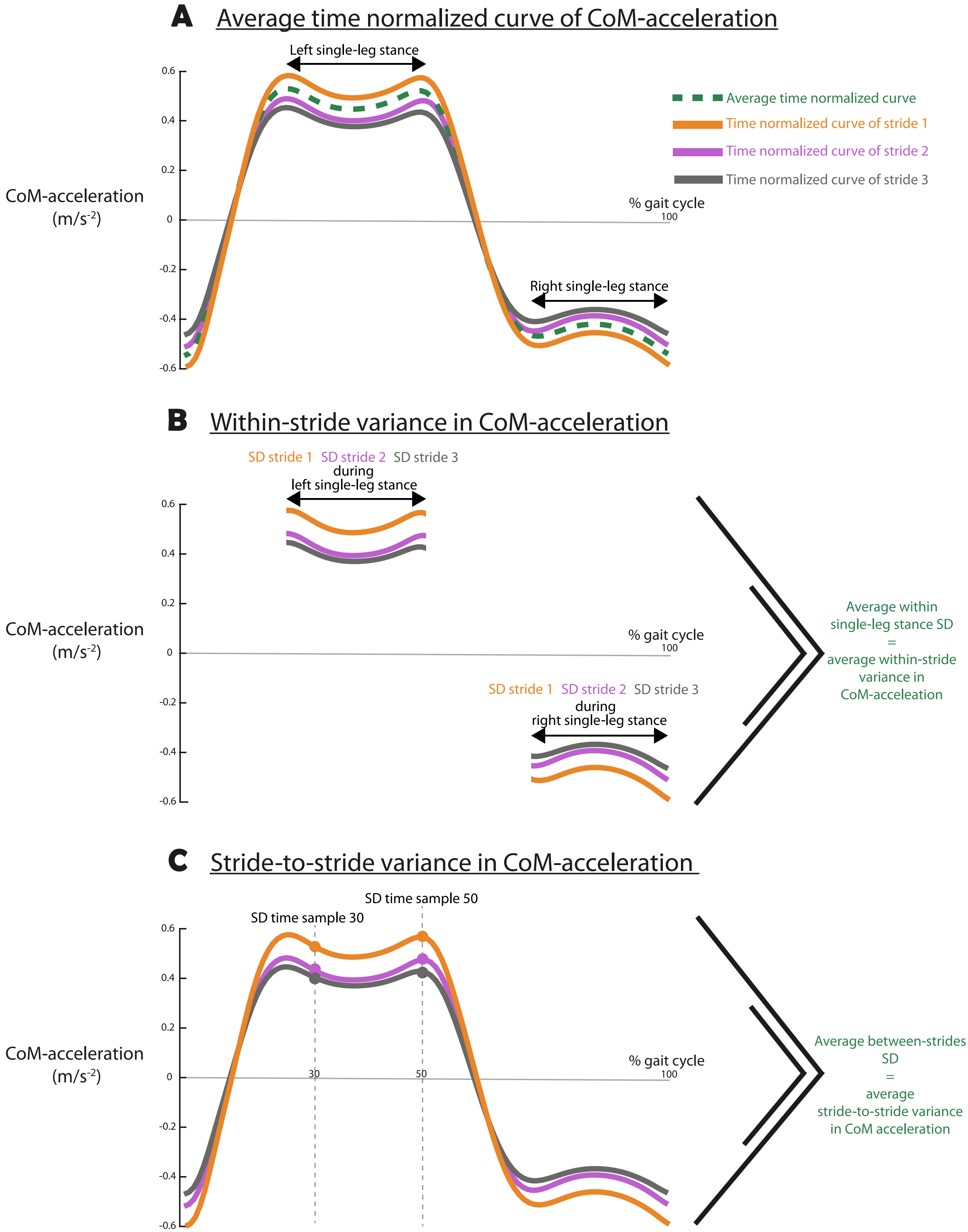

Fig. 2. Illustration of the calculation of A) the average time normalized curve of the center of mass (CoM) acceleration (dashed green line), which was determined by calculating the averaged curve across the time normalized curves of CoM-acceleration of in this example three strides (solid lines in orange, purple and grey), B) the within-stride variance in CoM acceleration (in green), which was expressed by the average of the standard deviations (SD) of the time series of the single-leg stance phases (solid lines in orange, purple and grey) and C) the stride-to-stride variance (in green), which was expressed by the SD over strides per time sample (dots), after which the SDs were averaged across all time samples (solid lines in orange, purple and grey). The dots correspond to data points at two of the 100 samples per stride. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

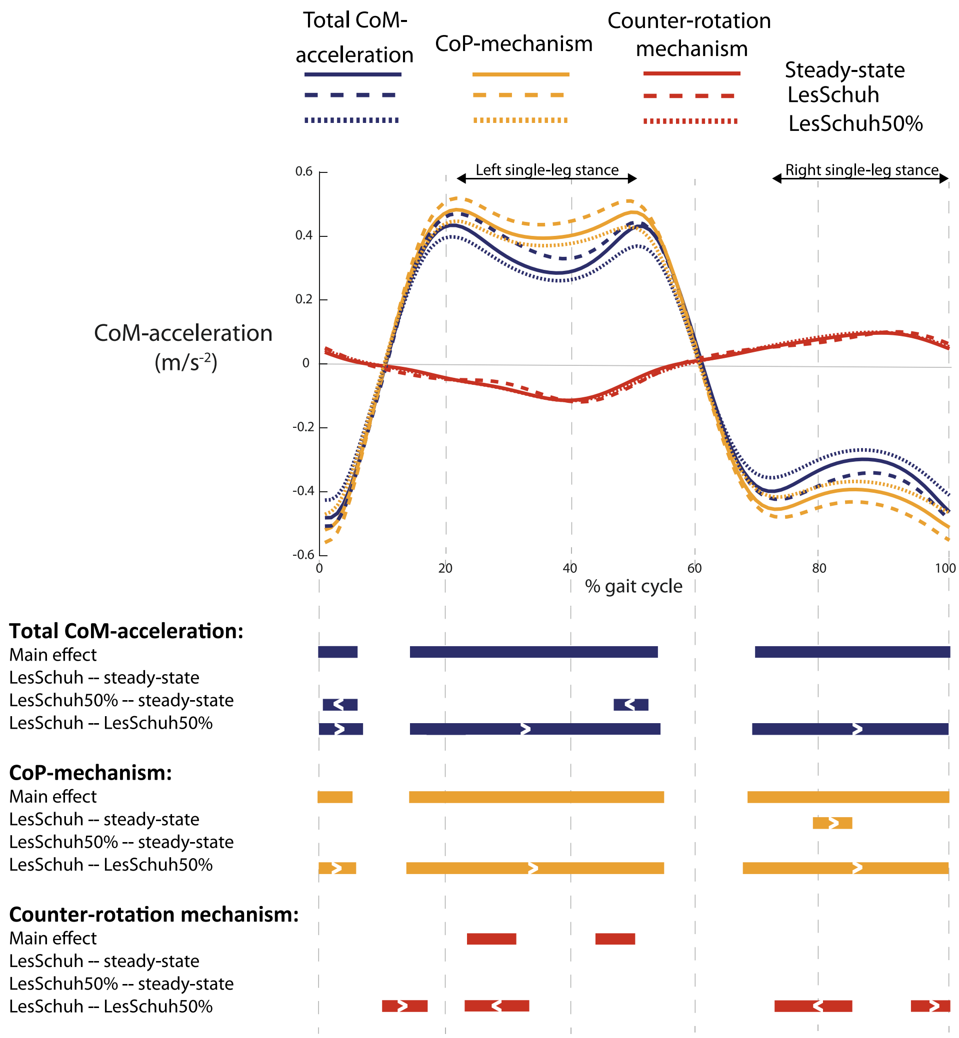

Fig. 3. Average time series (N = 13) of the center of mass (CoM) acceleration (in blue) and the contribution of the center of pressure mechanism (CoP-mechanism) (in yellow) and the counter-rotation mechanism (in red) during normal walking (steady-state, solid lines), walking on LesSchuh constraining the ankle mechanism (LesSchuh, dashed lines) and walking on LesSchuh at 50% of the average step width constraining the ankle mechanism and foot placement (LesSchuh50%, dotted lines). The gait cycle (0–100%) started at heel-strike of the left foot (0%) and ended at heel-strike of the same foot (left) (100%). The bars indicate gait phases with significant differences between conditions (main effects and post-hoc test results). The greater then (>) and less then (<) sign within the bars indicate if the CoM-acceleration during the first mentioned condition is greater or less then during the latter condition. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

3.2. CoP-mechanism

The contribution of the CoP-mechanism to the total CoM-acceleration, was increased in magnitude around 80% of the gait cycle during LesSchuh compared to steady-state walking, was similar between LesSchuh50% and steady-state walking and was increased in magnitude during the single-leg stance phase during LesSchuh compared to LesSchuh50% (Fig. 3).

The within-stride variance in CoM-acceleration due to the CoP-mechanism during single-leg stance differed significantly between conditions($F(2,24) = 20.411$, $p < 0.001$). It was smaller during LesSchuh50% compared to steady-state walking and LesSchuh($p < 0.001$ and $p = 0.001$, respectively;Fig. 4A).

The stride-to-stride variance in CoM-acceleration due to the CoP-mechanism differed significantly between the conditions($F(2,24) = 24.811$, $p < 0.001$). It was larger during LesSchuh50% and LesSchuh compared to steady-state walking($p = 0.011$ and $p < 0.001$, respectively) and was smaller during LesSchuh50% compared to LesSchuh ($p = 0.045$;Fig. 4B).

Step width differed significantly between conditions($F(1.316,15.793) = 58.234$, $p < 0.001$). It was smaller by about 28% during LesSchuh50% compared to LesSchuh and steady-state walking($p < 0.001$ and $p = 0.001$, respectively;Fig. 4C) and was larger during LesSchuh compared to steady-state($p = 0.003$;Fig. 4C).

3.3. Counter-rotation mechanism

The contribution of the counter-rotation mechanism to total CoM-acceleration was similar when comparing steady-state walking with LesSchuh and LesSchuh50% (Fig. 3). It was increased in magnitude around 15% and 95% of the gait cycle, and decreased in magnitude around 30% and 80% of the gait cycle during LesSchuh compared to LesSchuh50% (Fig. 3). The within-stride variance in CoM-acceleration due to the counter-rotation mechanism differed significantly between conditions($F(2,24) = 46.148$, $p < 0.001$). It was larger during LesSchuh50% compared to steady-state walking and LesSchuh($p < 0.001$ and $p < 0.001$, respectively;Fig. 4A). The stride-to-stride variance in CoM-acceleration due to the counter-rotation mechanism also differed significantly between conditions($F(1.210,14.515) = 120.775$, $p < 0.001$). It increased from steady-state walking to LesSchuh to

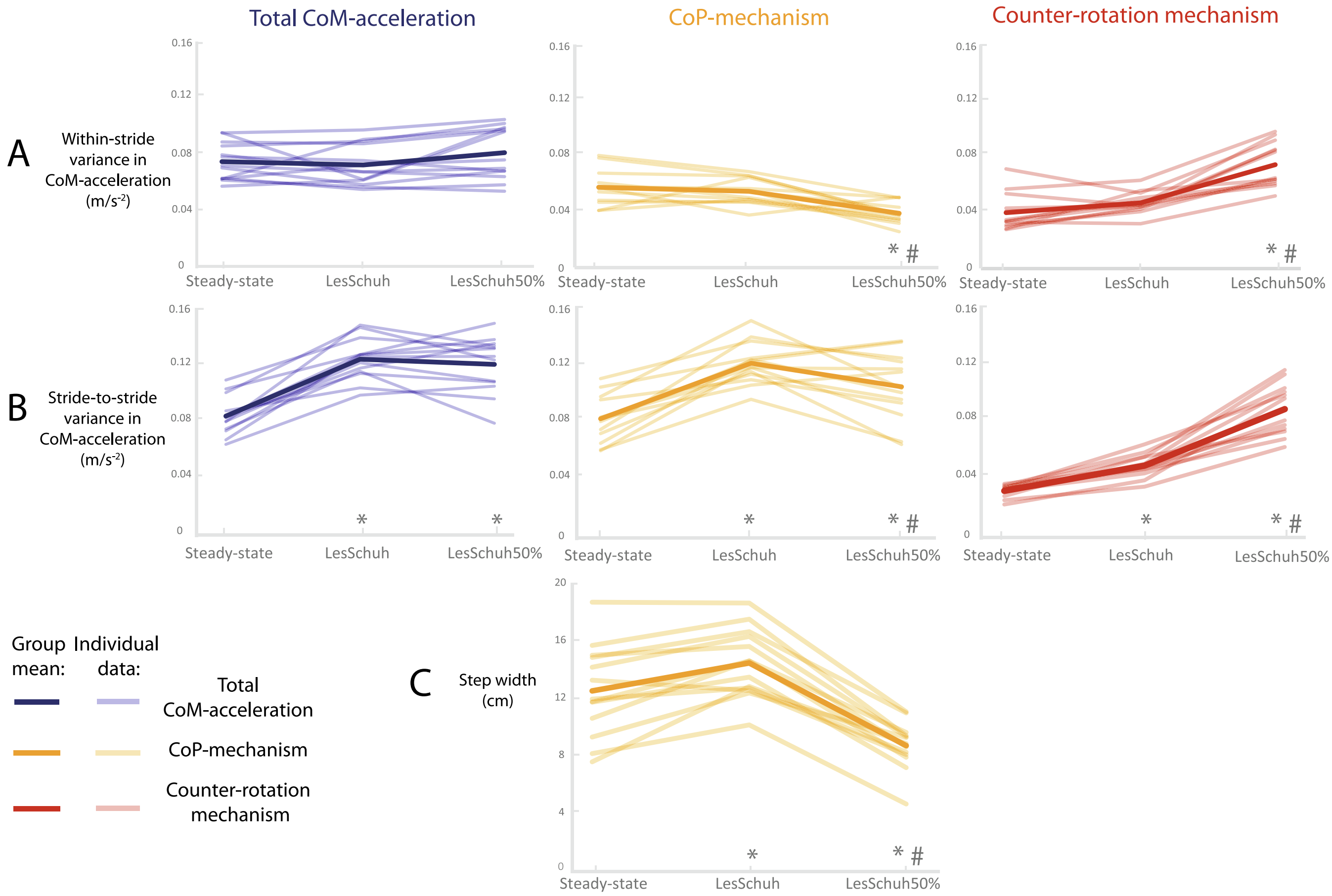

Fig. 4. The effect of Condition on A) the within-stride variances in the center of mass (CoM) accelerations during single-leg stance: total CoM-acceleration (in blue), CoM-acceleration due to the center of pressure mechanism (CoP-mechanism) (in yellow), and CoM-acceleration due to the counter-rotation mechanism (in red), B) the stride-to-stride variances in CoM-accelerations: total CoM-acceleration, CoM-acceleration due to the CoP-mechanism and CoM-acceleration due to the counter-rotation mechanism and C) the average step width. * represents a significant difference compared to steady-state walking. # represents a significant difference compared to LesSchuh. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

LesSchuh50%(all $p < 0.001$;Fig. 4B).

4. Discussion

We assessed the contribution of the CoP and counter-rotation mechanism to CoM-accelerations in normal walking and in conditions in which ankle moments (using a customized shoe(LesSchuh)) and foot placement were constrained in the mediolateral direction.

Walking on LesSchuh led to a larger step width compared to steady-state walking. This is consistent with previous literature, showing that adjusting foot placement can compensate for limited ankle moments due to narrow contact surface underneath a shoe or when walking on a prosthetic leg (Hof et al., 2007; van Leeuwen et al., 2020). Unexpectedly, we found that the within-stride variance in CoM-acceleration due to the CoP-mechanism only significantly decreased during walking with ankle and foot placement constraints(LesSchuh50%) compared to steady-state walking. Toeing-out could be a possible explanation for this mechanism not being increasingly used in the LesSchuh condition. Despite the instruction to place the feet facing forward, participants walked with a toe-out angle of about three degrees larger on LesSchuh compared to steady-state walking. This would allow for an additional CoP shift of approximately 1.3 cm, to compensate for the ankle moment constraint (assuming a shoe length of 25 cm, $\sin(3\ deg) * 25 = 1.3\ cm$).

Since within-stride variance in CoM-acceleration due to the counter-rotation mechanism was similar during walking on LesSchuh and steady-state walking, it seems that only constraining the ankle moments was not sufficient to provoke an increased contribution of the counter-rotation mechanism compared to steady-state walking. It appears not to be necessary to use the counter-rotation mechanism in addition to adjusting foot placement (i.e. walking with wider steps and toeing out).

As expected, we found smaller step width and smaller within-stride variance in CoM-acceleration due to the CoP-mechanism, during LesSchuh50% compared to steady-state walking and LesSchuh. The within-stride and stride-to-stride variance in CoM-acceleration due to the counter-rotation mechanism were larger during LesSchuh50% compared to steady-state walking and LesSchuh, which is consistent with our hypothesis. Compared to steady-state walking and LesSchuh, during LesSchuh50% the within-stride variance in CoM-acceleration due to the CoP-mechanism was smaller and the within-stride variance in CoM-acceleration due to the counter-rotation mechanism was larger. Consequently, no difference in the total within-stride variance in CoM-acceleration between conditions was found. It should be noted that an altered within-stride variance in CoM-acceleration (due to the CoP or counter-rotation mechanism) does not necessarily reflect increased or decreased control actions of the CoM, as these cannot be distinguished from internal perturbations.

The larger within-stride variance in CoM-acceleration due to the counter-rotation mechanism did not coincide with an increased contribution of the counter-rotation mechanism to CoM-acceleration during LesSchuh50% compared to steady-state walking and LesSchuh, and even a decrease around 15% and 95% of the gait cycle during LesSchuh50% compared to LesSchuh. The absence of significant differences between the conditions in mean contribution of the counter-rotation mechanism to total CoM-acceleration over strides may reflect that angular accelerations, with the possible exception of angular acceleration of the arm around the shoulder, ultimately need to be reversed, leading to the opposite effect on the total CoM-acceleration and thus allowing relatively high-frequency (within-stride) modulation only. Overall, the contribution of the counter-rotation mechanism to CoM-acceleration was around three times smaller and in opposite direction compared to that of the CoP-mechanism, while the within-stride variance in CoM-accelerations due to the CoP and counter-rotation mechanisms were more or less similar. This is consistent with previous findings, where we also found around three times smaller and counteracting contributions of the counter-rotation and CoP-mechanisms to CoM-acceleration in the anteroposterior direction during a normal step and the first recovery step after perturbation (van den Bogaart et al., 2020). The CoP-mechanism was the main contributor to total CoM-acceleration, possibly because the use of the counter-rotation mechanism may interfere with other task constraints, such as stabilizing the orientation of the head in space and preventing interference with the gait pattern.

In future studies, it is worthwhile to assess the use of the counter-rotation mechanism in different (patient) populations, which might use the CoP and counter-rotation mechanisms differently. Future studies should determine the link between fall risk and the use of the counter-rotation mechanism. Training the use of specific mechanisms could be implemented in therapeutic interventions that aim to decrease fall incidence. However, whether and how a specific mechanism can be trained also needs further investigation.

5. Conclusion

We found a smaller within-stride variance in CoM-acceleration due to the CoP-mechanism and a larger within-stride variance in CoM-acceleration due to the counter-rotation mechanism during LesSchuh50% compared to steady-state walking. This suggests that the counter-rotation mechanism is used to stabilize gait when needed. However, the mean contribution of the counter-rotation mechanism to CoM-acceleration over strides did not increase during LesSchuh50% compared to steady-state walking. Overall, the CoP-mechanism was the main contributor to the total CoM-acceleration. The use of the counter-rotation mechanism may be limited, because angular accelerations ultimately need to be reversed and because of interference with other task constraints, e.g. interference with the gait pattern.

CRediT authorship contribution statement

Maud van den Bogaart: Conceptualization, Investigation, Formal analysis, Methodology, Visualization, Writing – original draft, Writing – review & editing. **Sjoerd M. Bruijn:** Conceptualization, Methodology, Supervision, Writing – review & editing. **Joke Spildooren:** Supervision, Writing – review & editing. **Jaap H. van Dieën:** Conceptualization, Methodology, Supervision, Writing – review & editing. **Pieter Meyns:** Conceptualization, Methodology, Supervision, Writing – review & editing.

Declaration of Competing Interest

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

Acknowledgments

The authors are thankful for all participants, Leon Schutte for technical support and Moira van Leeuwen and Marijne Nieuwelink for assistance during data collection.

Funding

Sjoerd M. Bruijn was supported by a grant from the Netherlands Organization for Scientific Research (NWO#451-12-041).

Data availability

The data that support the findings of this study are available on request from the corresponding author[SMB].

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2022.111073.

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