Does ankle push-off correct for errors in anterior-posterior foot placement relative to center-of-mass state?

Jian Jin[1], Jaap van Dieën[1], Dinant Kistemaker[1], Andreas Daffertshofer[1,2], Sjoerd M. Bruijn[1,2]

[1] *Department of Human Movement Sciences, Faculty of Behavioural and Movement Sciences,* Amsterdam, The Netherlands

[2] *Institute of Brain and Behavior Amsterdam*, Amsterdam, The Netherlands

I. ABSTRACT

Understanding the mechanisms humans use to stabilize walking is vital for predicting falls in elderly. Modelling studies identified two of them: foot placement control and push-off control. Foot placement control relies on center-of-mass (CoM) state, by stepping in the direction of CoM movement, which has been found in human gait. Push-off control has been suggested to modulate the deviations of CoM trajectories from the desired trajectories in perturbed human walking, but it is unknown whether this mechanism is also used in steady-state (unperturbed) walking. To test this, we considered the covariance between CoM state and anterior-posterior foot placement as a measure of foot placement accuracy and investigated whether the difference between the actual and the ideal foot placement based on the CoM state at heel strike may serve as a feedback signal for push-off control. We found the (ankle) push-off torque to be correlated to the foot placement error in 30 participants when walking at normal and slow speed, with mean correlation values of up to 0.45. Our study hence suggests that humans use a push-off strategy for correcting foot placement errors in steady-state walking.

II. INTRODUCTION

Walking is a seemingly simple task for most of us. Yet, it involves intricate feedback control and the continuous integration of sensory inputs and generation of motor outputs (Rossignol et al., 2006; Warren et al., 2001; Zehr & Stein, 1999). A solid understanding of the mechanisms that underlie the stabilization of walking is crucial for identifying causes of falls in elderly (McGibbon, 2003). In-silico studies singled out basic principles of locomotion. Two examples are feedback control of push-off and of foot placement. Both of them been shown to stabilize comparably simple walking models in the presence of perturbations. Hobbelen & Wisse, (2008) illustrated that modulating ankle push-off torque via feedback from the leading leg's angle substantially improved robustness against perturbations in a sagittal plane flat-feet walker. According to Byl & Tedrake, (2008), feedback-controlled foot placement with a constant push-off magnitude appears more robust than only modulating the push-off magnitude in a simple point-feet walker model. In inverted pendulum models, combined push-off and foot placement control based on feedback from mid-stance speed demonstrated good robustness (Bhounsule, 2015; Kelly & Ruina, 2015; Zaytsev, Wolfslag, & Ruina, 2018). In humans, there is evidence for foot placement control with feedback from the CoM state (position and velocity) both in steady-state (Wang & Srinivasan, 2014) and perturbed walking (Joshi & Srinivasan, 2019; Vlutters et al., 2018). There is also evidence of push-off control with feedback from the deviations of CoM trajectories from the desired trajectories (Afschrift et al., 2021; van Mierlo et al., 2021; Vlutters et al., 2016), but this evidence is restricted to perturbed walking. In view of the aforementioned modelling and perturbed walking studies, we here subsume that humans may also use a feedback-controlled push-off strategy in steady-state walking.

In human walking, foot placement in anterior-posterior (AP) direction is correlated to the CoM state. The CoM states at mid-stance and at heel strike both predict about 33% of the variance in AP foot placement (Wang & Srinivasan, 2014). By the same token, the ankle push-off during the double-stance phase seems to modulate with AP pelvis perturbations, as van Mierlo et al., 2021 found adjustments of the double-stance duration and the AP center-of-pressure (CoP) trajectory to correct perturbed CoM state towards a desired state at the end of the double-stance phase. Afschrift et al., (2021) demonstrated that for both AP pelvis perturbations and treadmill belt speed perturbations, deviations in CoM velocity from the steady-state trajectory accurately predicted the ankle moment after a neural delay of 100 ms. In steady-state walking, the percentage of explained variance of AP foot placement from the CoM state increases from 33% to 70% during the double-stance phase (Wang & Srinivasan, 2014). This clearly suggests some modulation of the CoM state through push-off control. Since the CoM state covaries with AP foot placement, neither of them can be isolated

as feedback signal for push-off control. A more appropriate feedback signal is the *foot placement error*, defined as the difference between actual foot placement and the ideal foot placement based on the CoM state at heel strike. Here, the ideal foot placement can be linearly predicted via the CoM state at heel strike, where the linear regression coefficients may be obtained by a least square fit between CoM state and foot placement (Wang & Srinivasan, 2014). Positive and negative foot placement errors at heel strike are illustrated in Fig. 1. Note that the covariance between CoM state and foot placement has been suggested to subserve gait stability in both simple walking models and in humans (Hof, 2008; Joshi & Srinivasan, 2019; Redfern & Schumann, 1994; Wang & Srinivasan, 2014). Assuming the presence of foot placement control, deviations from this covariance can be viewed as foot placement *errors* that need to be corrected.

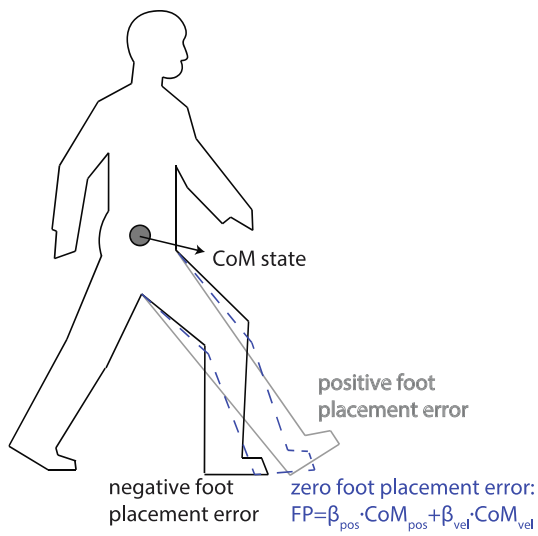

Fig. 1. Illustration of positive and negative foot placement errors. For a particular CoM state (CoM AP position and velocity at heel strike), a too large step leads to a positive foot placement error, a too small step leads to a negative foot placement error. A vanishing foot placement error corresponds to linear predictions of foot placement from the CoM state.

We hypothesized that push-off of the trailing leg corrects for AP foot placement errors at heel strike. We expected push-off correction to be proportional to the AP foot placement error; specifically, a stronger push-off for larger positive foot placement errors, and less push-off for larger (more negative) foot placement errors. The latter may be motivated by the fact that the total energy of the CoM after heel strike is lower for a positive foot placement error than for a negative one. Lower energy requires stronger push-off to accelerate the CoM, irrespective of being caused by a too large collision loss (from too large step length), by a too backward CoM position, and/or by a too low CoM velocity.

We investigated the relationship between foot placement errors and three kinetics time series (combined AP ground reaction force (GRF), trailing leg AP GRF and ankle moment) that affect AP CoM acceleration in 30 participants. We also compared for each participant the mean kinetics of ten most positive foot placement

errors with the mean kinetics of ten most negative foot placement errors to see how push-off may differ when correcting positive and negative foot placement errors. We selected these specific time series because they represent the summed effect of push-off by the trailing leg and braking by the leading leg; the CoM is mainly accelerated by the trailing leg (Hernández et al., 2009). Since ankle push-off provides most of the trailing leg's propulsive force (Toney & Chang, 2016) and modelling studies suggest ankle push-off may stabilize gait (Hobbelen & Wisse, 2008; Kerimoglu et al., 2021; Kim & Collins, 2017), we finally evaluated the correlations between foot placement errors and ankle moment.

III. METHODS

We used existing data from steady-state normal and slow treadmill walking experiments in a recent study of our group (van Leeuwen et al., 2020a). The methods section below briefly describes how data were collected and preprocessed. For further details, we refer to the original study by van Leeuwen et al., (2020b). All data used in the current study can be found at https://doi.org/10.5281/zenodo.4229851, and the code for the analyses presented here can be found at: https://surfdrive.surf.nl/files/index.php/s/piXp5iYKxMPi7MU.

Participants. 30 participants were included (19 female, 11 male, age 30 ± 7 years, weight 70 ± 13 kg, height 1.73 ± 0.08m; mean ± sd). None of them reported injuries or balance issues that could affect their gait pattern. All participants signed informed consent before the experiment. Ethical approval (VCWE-2018-159) had been granted by the ethics board of the Faculty of Behavioural and Movement Sciences, Vrije Universiteit Amsterdam, prior to conducting the experiment.

Protocol. Every participant walked on a treadmill at a constant belt speed of $1.25 \cdot \sqrt{\text{leg length}}$ m/s (*normal* walking condition) and $0.63 \cdot \sqrt{\text{leg length}}$ m/s (*slow* walking condition). A metronome served to impose stride frequency. Participants were asked to match their right heel strikes to the metronome beat. The imposed frequency was customized as the average preferred stride frequency during the last hundred steps of the familiarization trial at each speed. The normal walking trials lasted five minutes and the slow walking trials ten minutes each to ensure that data of at least 200 consecutive strides were collected. Normal and slow speed trials were randomized in order and separated by sufficient breaks to prevent fatigue.

Data collection and processing. Participants walked on an instrumented dual-belt treadmill (Motek-Force-link, Amsterdam, Netherlands). Full-body kinematics were recorded using an active 3D motion analysis system and cluster markers on all segments (Optotrak, Northern Digital Inc, Waterloo ON, Canada). For every participant and condition, we analyzed the last 200 consecutive strides without data quality issues (e.g., limited marker visibility, large noise, etc.). GRF data were collected using a force plate integrated in the treadmill. Gait events (heel strike and toe-off) were detected based on the "butterfly pattern" of the combined CoP trajectory (Roerdink et al., 2008).

A left (right) stride was defined from left (right) heel strike to left (right) heel strike and a left (right) step from left (right) toe-off to right (left) toe-off. Ankle joint moments were calculated using the GRFs and the lower body kinematics via a bottom-up inverse dynamics approach (Kingma et al., 1996). For subsequent analysis, the kinematics and kinetics time series of a stride and a step were time-normalized. To account for the variations in timing of heel strike and subsequent push-off within a step[1], we segmented the time window of a step to single-stance and double-stance sub-windows with fixed time length per walking condition.

Linear Foot placement model. Similar to previous studies (van Leeuwen et al., 2020b; Wang & Srinivasan, 2014), we fitted a linear model between the CoM state and the AP foot placement. This full body CoM state was derived from a weighted sum of the body segments' CoM, which in turn was estimated from the longitudinal axis of the body segments' CoM (De Leva, 1996; van Leeuwen et al., 2020b). The predictor CoM state included the AP position (the horizontal distance from stance foot to CoM) and the AP velocity, i.e. $\mathrm{CoM_{pos}}$ and $\mathrm{CoM_{vel}}$, respectively. The predictor and dependent variables were de-meaned prior to regression to ensure a zero intercept. We used the following model:

$$\mathrm{FP} = \beta_{\mathrm{pos}} \cdot \mathrm{CoM_{pos}}(\mathrm{i}) + \beta_{\mathrm{vel}} \cdot \mathrm{CoM_{vel}}(\mathrm{i}) + \varepsilon(\mathrm{i}), \tag{1}$$

where the foot placement (FP) is the (de-meaned) AP distance between trailing and leading leg, β_{pos} and β_{vel} are regression coefficients obtained from the least square fit, and $\varepsilon(\mathrm{i})$ denotes the residual from the linear regression, or, in short the foot placement error. The index i represents the time instant of a stride or a step. We used the time instant just after contralateral heel strike to compute foot placement (FP) and foot placement error $\varepsilon(\mathrm{i})$, which we assume to be corrected by the subsequent push-off.

[1] The total samples of time window for single-stance phase, double-stance phase in normal and slow walking were different on average (over all steps from all participants).

Correction of foot placement error. We evaluated how well the foot placement error of the leading leg at heel strike predicted combined AP GRF, trailing leg AP GRF, and trailing leg ankle moment during the double-stance, through regression analysis. In addition, we compared the kinetics time series of the 'most positive' with that of the 'most negative' foot placement error. Specifically, the foot placement errors of each participant (200 strides) were sorted from low to high, and the ten most positive and ten most negative foot placement error strides were analyzed, by extracting the mean kinetics time series from these sets of ten strides.

Statistics. All statistical tests and analysis were performed in Matlab (v2017b, The Mathworks Inc., Natick, MA), including the multilinear regression of the foot placement model. To test the correlation between foot placement error and the kinetics time series, we identified correlations between foot placement error and the evaluated kinetic time series for every participant and condition. Correlation values are presented as group level means and individual data. The mean correlations were obtained using Fishers r-to-z transformation prior to averaging and then do inverse transformation. We tested for significance using statistical parametric mapping (Friston et al., 2007) as implemented in the SPM1D toolbox (https://spm1d.org/). In brief, SPM1D allows for statistical testing of time series data, considering the interdependence (smoothness) between time samples. To test for differences between the kinetics time series of the 'most positive' and 'most negative' foot placement errors, we also performed an SPM1D paired t-test. The significance level of every test was set to α=0.05.

IV. RESULTS

The variance in AP foot placement explained by the CoM state (R^2) increased over the single-stance phase (Fig. 2, left of the dashed vertical line) and remained mostly between 50% and 80% during the double-stance phase in both normal and slow walking (Fig. 2, right of the dashed vertical line).

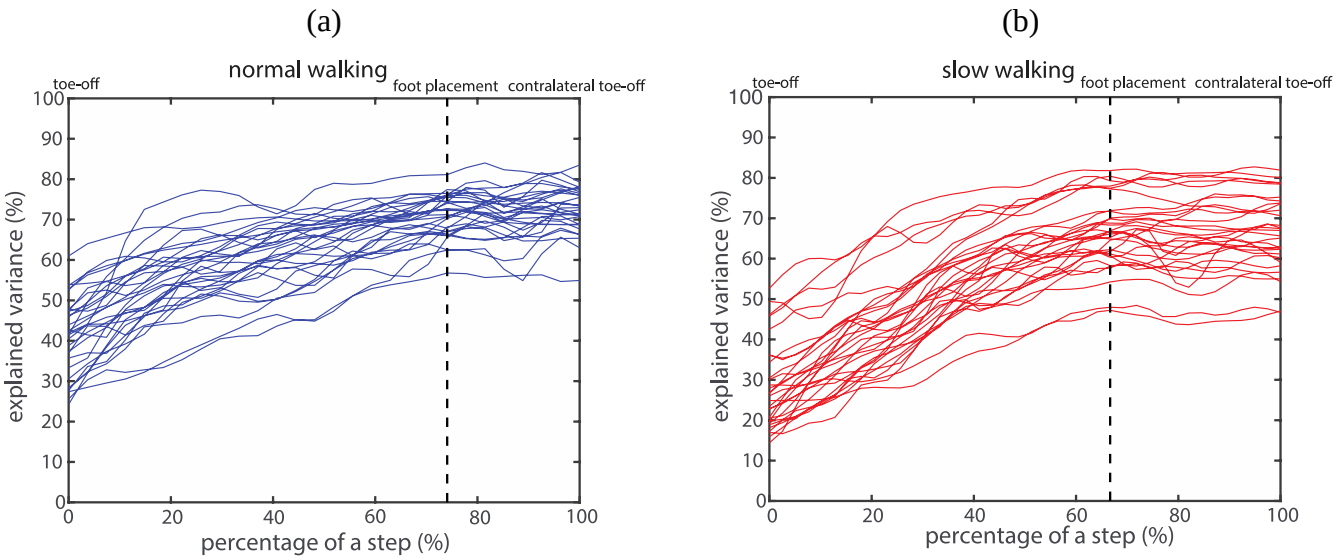

Fig. 2. Proportion of foot placement variance explained by CoM state, for normal (a) and slow walking (b). Every curve represents the explained variance for a participant. The dashed vertical lines correspond to foot placement, and the phases after heel strike correspond to the double-stance phase.

Although a large proportion of foot placement variance could be explained by the CoM state at heel strike, a fair amount of unexplained variance prevailed, i.e., there were substantial foot placement errors. The distributions of foot placement errors across all participants for normal and slow walking were close to Gaussian with zero mean (Fig. 3a). The variance of foot placement errors in slow walking was larger than in normal walking (Fig. 3a). The group means of the ten most positive and ten most negative foot placement errors in slow walking had larger magnitudes compared to those in normal walking (Fig. 3b).

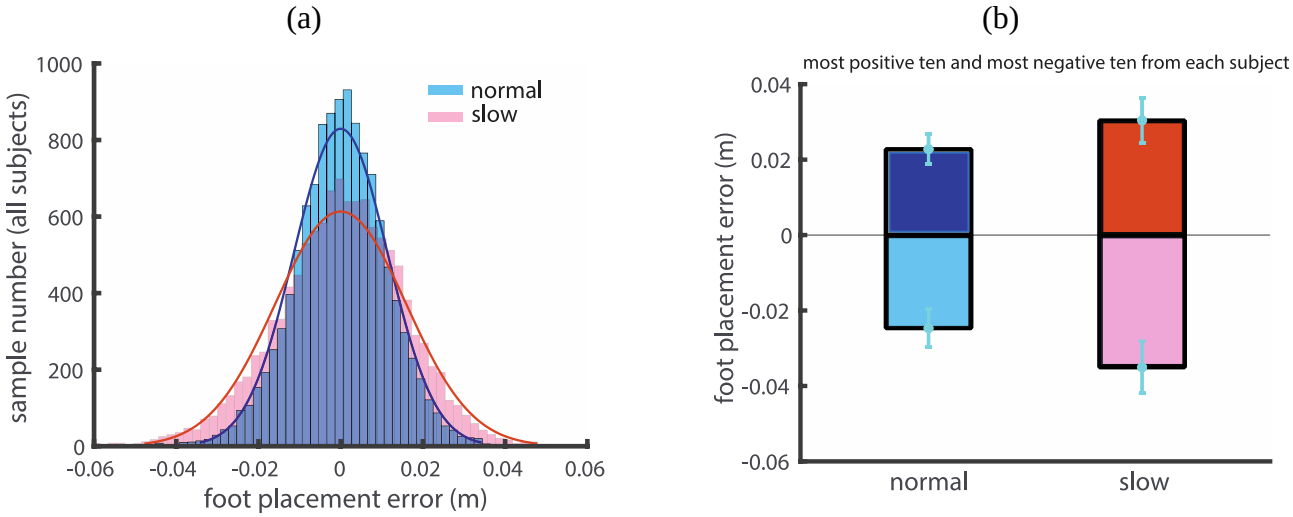

Fig. 3. (a) Distributions of foot placement errors for normal and slow walking for all participants. Gaussian distributions are fitted to the data in two conditions. (b) The group mean values of the averaged most positive ten and most negative ten foot placement errors per participant. Error bars indicate the standard deviation across participants.

Foot placement errors were significantly correlated with the combined AP GRF for all participants during the double-stance phase in normal and slow walking (mean correlations up to 0.34 and 0.28, respectively; see Fig. 4). In both cases, the high positive correlations occurred in the first half of the double-stance phase, and decreased to negative correlations over the latter half of the double-stance phase. Correlations were close to zero prior to foot placement, implying that the accuracy of foot placement cannot be predicted by the combined AP GRF before foot placement. Correlations were also close to zero at foot placement, suggesting that pre-emptive push-off indicated by the combined AP GRF did not cause foot placement errors. Correlations were significant and negative at the end of the stride cycle during normal walking (before the subsequent heel strike, mean values reached -0.43, Fig. 4). See the *Supplementary Material* S2 for further details on the regression.

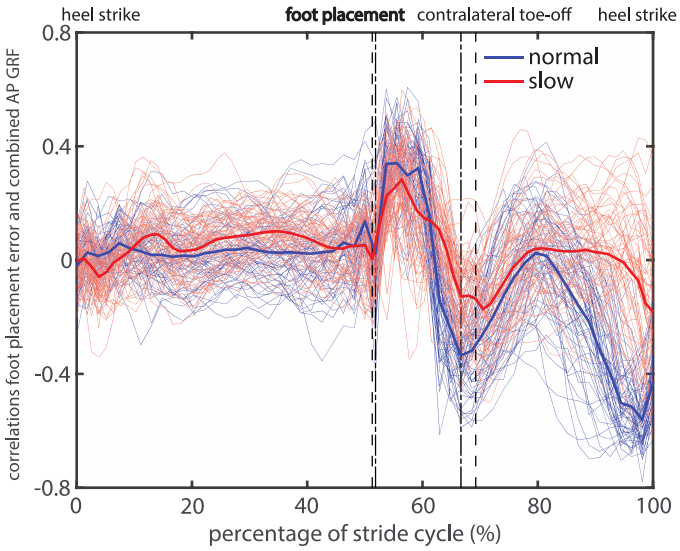

Fig. 4. Correlations between foot placement error (at heel strike indicated by the left dashed line) and the *combined* AP GRF, for normal and slow walking. The thin curves are the correlations for each participant and each foot (left, right), and the thick curves are the mean correlations over participants. The phases between the dashed lines correspond to double-stance for normal (dash-dotted) and slow (dashed) walking, respectively.

Likewise, the trailing leg AP GRF displayed significant correlations with foot placement error at heel strike in normal and in slow walking (mean correlations up to 0.45 and 0.41, respectively; see Fig. 5 and *Supplementary Material* S3 for the corresponding statistical tests). Correlations were highest in the middle of the double-stance phase, i.e., push-off modulations occurred mostly around this phase. Correlations were close to zero at the beginning and at the end of the double-stance phase, suggesting that pre-emptive, early and late push-off did not contribute to (the cause and/or correction of) foot placement errors.

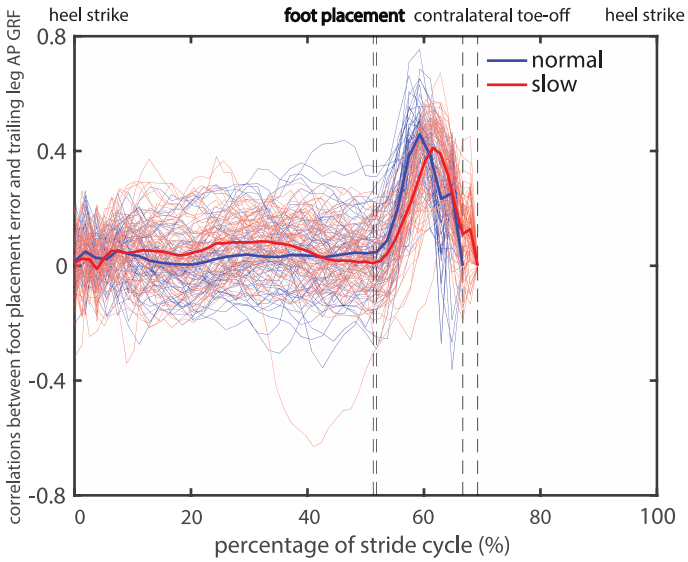

Fig. 5. Correlations between the foot placement error (at heel strike indicated by the left dashed line) and *trailing leg* AP GRF, for normal and slow walking. The thin curves are the correlations for each participant and each foot (left, right), and the thick curves are the mean correlations over participants. The phases between the dashed lines correspond to double-stance for normal (dash-dotted) and slow (dashed) walking, respectively. The trailing leg GRF was zero during the swing phases at the end of the stride cycle, thus the correlations were not shown.

The ankle moments during double-stance were significantly correlated with foot placement errors at heel strike in normal and slow walking (mean correlations up to 0.39 and 0.37, respectively; see Fig. 6 and *Supplementary Material* S4). Note that we observed one outlier with negative correlations. This participant's right leg ankle moment in slow walking was negatively correlated to foot placement error, and his/her foot placement of the right leg was consistently longer than of the left leg. We suspect this was a different and uncommon asymmetric walking strategy in slow walking to meet the imposed treadmill and metronome constraints.

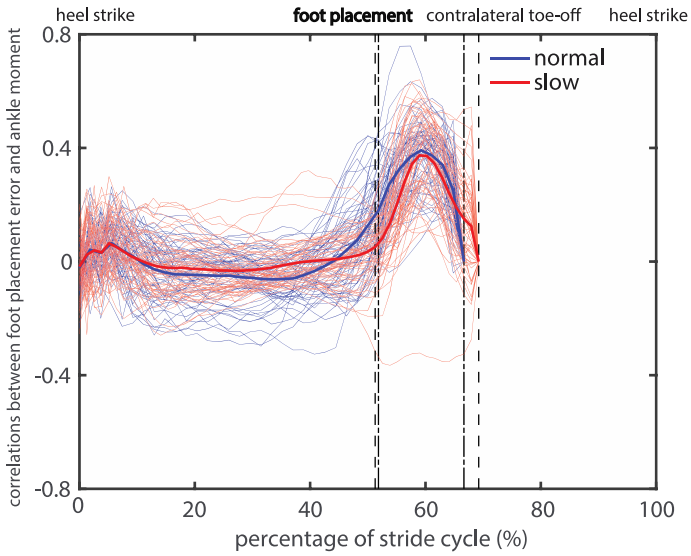

Fig. 6. Correlations between the foot placement error at heel strike, i.e., the left dashed line) and the contralateral ankle moment, for normal and slow walking. The thin curves are the correlations for each participant and each foot (left, right), and the thick curves are the mean correlations over participants. The phases between the dashed lines correspond to double-stance for normal (dash-dotted) and slow (dashed) walking, respectively.

In line with the correlations we found, all the kinetic time series during the double-stance phase were significantly larger for 'most positive' foot placement errors than for 'most negative' foot placement errors in both walking conditions (Fig. 7a-c and *Supplementary Material* S5). In line with Figs. 4-6, the combined AP

GRF was higher in the first half and lower in the end of the double-stance phase for ‘most positive’ foot placement errors; the trailing leg AP GRF and ankle moment were higher in the middle of the double-stance phase for ‘most positive’ foot placement errors.

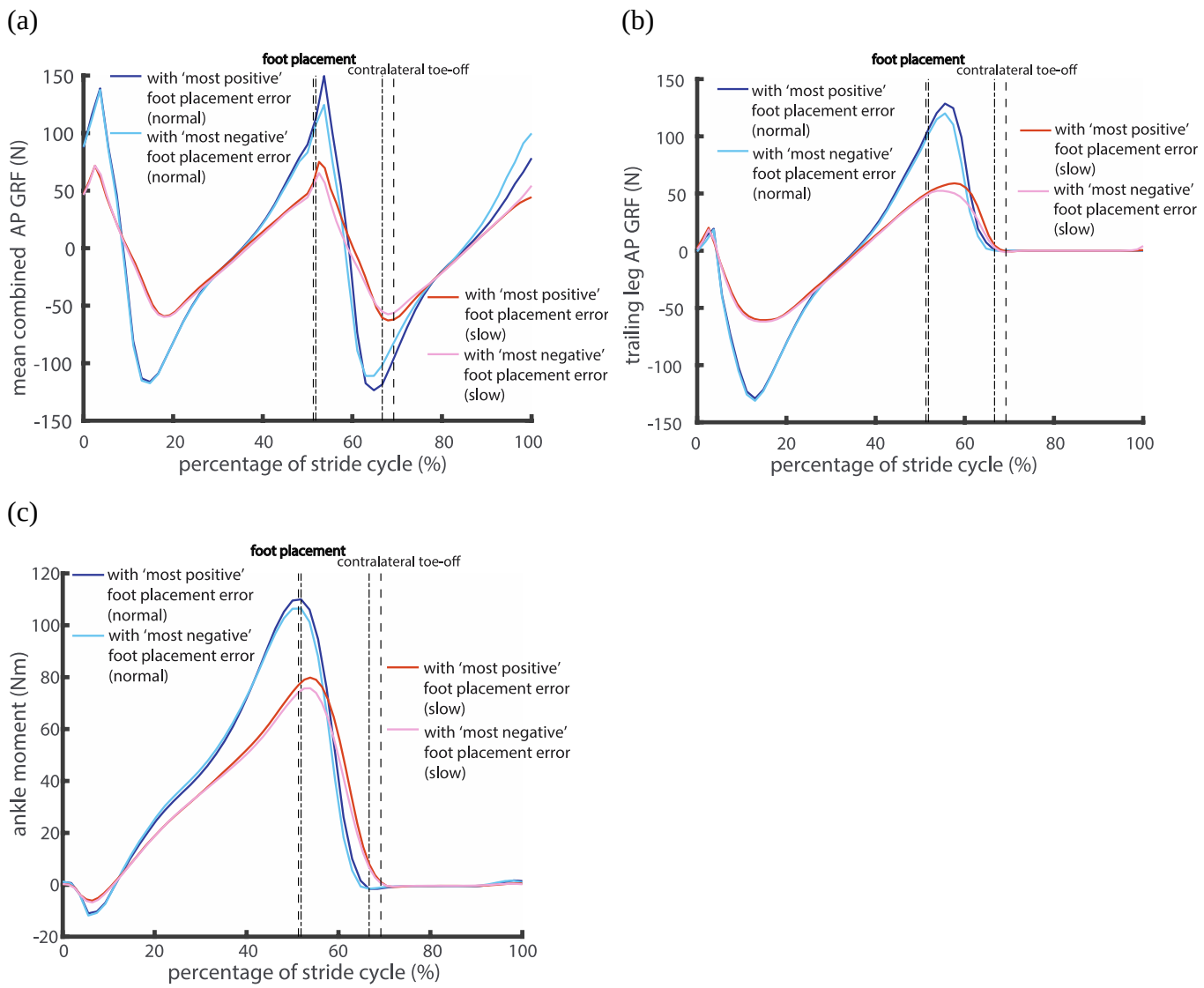

Fig. 7. Averaged kinetics time series corresponding to the ‘most positive’ and ‘most negative’ foot placement errors in normal and slow walking. (a) combined AP GRF; (b) trailing leg AP GRF; (c) ankle moment. The phases between the dashed lines correspond to double-stance for normal (dash-dotted) and slow (dashed) walking, respectively.

V. DISCUSSION

We sought to test whether push-off control is being used to compensate for errors in AP foot placement in steady-state walking in humans. We found combined and trailing leg AP GRF and ankle moment to be correlated to foot placement errors at heel-strike. Feedback control of push-off has previously been demonstrated to improve robustness against perturbations in modelling studies (Hobbelen & Wisse, 2008; Kim & Collins, 2013, 2017) and was found to be used by humans during walking to correct for AP perturbations (Afschrift et al., 2021; Joshi & Srinivasan, 2019; van Mierlo et al., 2021; Vlutters et al., 2016). Recently, van Leeuwen et al., (2021) reported that errors in mediolateral foot placement are being corrected by ankle moments during the subsequent stance phase. We here supplemented these findings by showing evidence for a push-off mechanism used during steady-state gait in the AP direction.

High correlations between CoM state and foot placement can be achieved by foot placement in the direction of the CoM movement (Wang & Srinivasan, 2014) or by modulating the CoM state. The modulation of CoM state during the double-stance phase can be achieved by (ankle) push-off control, or could be a mechanical consequence of foot impact with the ground. Previous studies evaluated the correlations between ankle moment and CoM state in AP perturbed walking (Afschrift et al., 2021; van Mierlo et al., 2021). Studies combining this evaluation of correlations (between ankle moment and CoM state) with analysis of the proceeding foot placement are few and far between to our knowledge (Vlutters et al., 2018). As foot placement cannot be accurate at every step, its difference with the predicted foot placement from CoM state at heel strike, i.e., the foot placement error, may be a candidate input for feedback control. This idea finds support by an increase in (linear) correlations between CoM state and AP foot placement over the double-stance phase, shown in our study (see Fig. 2) and reported earlier by Wang & Srinivasan, (2014). The identified push-off strategy in our study may be interpreted as a way to control the CoM state towards a desired value relative to the stance foot position at the end of the double-stance phase, in order to maintain gait stability. In the following, we discuss the main mechanisms to adjust the AP CoM state during the double-stance phase by kinetics variables like combined AP GRF, trailing leg AP GRF, and ankle moment.

We found that in steady-state normal walking, the correlations between foot placement error and trailing leg AP GRF are highest (up to 0.45), followed by the correlations with ankle moment (up to 0.39), and then with combined AP GRF (up to 0.34). Arguably, feedback control based on foot placement error is mainly used for control of the push-off force by the trailing leg which is strongly determined by the ankle moment. Push-off may be generated by ankle, knee, and hip muscles (Kuo, 2002). At the instant of maximum trailing leg power,

the gain of the ankle moment to the trailing leg AP GRF (by computing the Jacobian matrix from moment to GRF) is ten times larger than the gain of the hip joint moment (Toney & Chang, 2016). It is hence not unexpected that correlations for the ankle moment are close to the correlations for the trailing leg AP GRF.

Peaks in the correlations between foot placement error and kinetic variables all occurred around the middle of the double-stance phase (Figs. 4-6), while the peak combined/trailing leg AP GRF and the peak ankle moment all occurred already in the beginning of the double-stance phase (Fig. 7). These two observations imply that modulations of the CoM state are less prevalent during the peak combined/trailing leg AP GRF or peak ankle moment, but mostly around the middle of the double-stance phase, at which time the CoM AP velocity is highest (Hernández et al., 2009; Kuhman & Hurt, 2019; Lipfert et al., 2014). Modulations of the CoM state may be achieved either by adjusting the trailing leg AP GRF and ankle moment magnitude (e.g., Fig. 7c), or by modulating their timings (duration of double-stance phase and relative duration of particular force/torque in the double-stance phase) (Kuhman & Hurt, 2019; Toney & Chang, 2016; van Mierlo et al., 2021; Williams & Martin, 2019). For instance, in slow walking the peak in combined AP GRF occurs later in too large foot placement error steps than in too small foot placement error steps (Fig. 7b), which may imply a prolonged activation of the muscles contributing to this force. Yet, here the prolonged activation mechanism in slow walking cannot be directly verified or falsified from Fig. 4-7 due to the normalization of the double-stance phase to a fixed window. The time delay between heel strike and the peak in the correlations between foot placement errors and ankle moment in normal walking was about 80 ms. Feedback delay in control of human ankle moments is arguably longer (≈ 100 ms) due to signal transmission, sensory integration in the nervous system and the electromechanical delay (Afschrift et al., 2021; Welch & Ting, 2008). Rather than directly using foot placement error at heel strike, humans might predict their foot placement error shortly before heel-strike as input for the (ankle) push-off modulation. The correlations between the CoM state and the AP foot placement shortly before heel strike were close to the ones at heel strike (Fig. 2). This implies more or less similar foot placement accuracy predicted prior to or at foot placement. Whether and how humans estimate foot placement error before, at, or after foot placement, however, requires further investigation.

Our finding that push-off modulation (indicated by the peak correlations in Figs. 4-6) does not occur earlier than contralateral heel-strike in steady-state walking is consistent with Kuhman & Hurt, (2019). Humans seemingly rely less on the more energy efficient preemptive push-off strategy to stabilize gait than suggested by modelling studies (Kuo, 2002; Ruina et al., 2005). Potentially, they opt for producing push-off somewhat later to incorporate feedback from foot placement error.

We showed that steady-state gait relies on more or less the same push-off strategy as perturbed walking, where a forward push led to more forward foot placement (Vlutters et al., 2018), followed by CoP modulation (shorter double-stance duration and longer CoP distance travelled, see van Mierlo et al., 2021) from either weight shift of the leading leg (Hof, 2007) or plantarflexion ankle moment modulations (Gruben & Boehm, 2014; Vlutters et al., 2018), contributing to a reduction of the CoM AP velocity over the double-stance phase. Van Mierlo et al., (2021) found CoP modulation to be absent for backward perturbations, presumably because adjustments in foot placement sufficiently reduced the foot placement error (see Fig. 5 from Vlutters et al., 2018). According to Vlutters et al., (2018b) changing magnitude and direction of AP perturbations on the pelvis near foot contact hardly affected the AP foot placement in both the first and second recovery step. This suggests that a limited response time for foot placement modulation primarily calls for push-off modulation. Apparently, humans cope with perturbations by complementing foot placement with push-off control to correct the remaining foot placement errors.

Older adults have poorer foot placement accuracy, at least in the mediolateral direction (Arvin et al., 2018). They have a reduced (ankle) push-off power generation (Franz, 2016; Hernández et al., 2009). One can therefore expect the older adults have poorer gait robustness due to larger foot placement errors and a reduced capacity to correct for these. Future work may compare age-related differences in foot placement error at the beginning and the end of the double-stance phase and relate them to differences in push-off modulation. This will further clarify gait adaptations in the elderly, such as the distal-to-proximal redistribution of push-off power production (Franz, 2016).

We interpreted the modulation of the push-off force by the trailing leg in terms of gait stability. Admittedly, the modulation of push-off force might also be explained by speed regulation. On a treadmill, speed is controlled from step to step and may results in an over-correction of speed errors with respect to the treadmill speed (Dingwell & Cusumano, 2015). When walking at different steady-state speeds, humans can adjust the relative duration of double-stance phase (cf. the percentage for normal and slow walking in Fig. 7), which may lead to changes in magnitudes of AP GRF (Williams & Martin, 2019). Here, we also observed over-correction of foot placement errors by the combined AP GRF at the subsequent step (see the end of the stride cycle in Fig. 4 and Fig. 7a). The foot placement error is usually determined by two interacting variables: CoM state and foot placement at heel-strike. For an inverted pendulum model walking at a nominal speed, perturbations leading to positive foot placement errors reduced walking speed and perturbations leading to

negative foot placement errors increased walking speed (Hof, 2008). Corrections of foot placement errors probably also correct speed errors with respect to the treadmill. The present data do not allow for distinguishing whether maintaining constant speed or gait stability were the primary control goal. Either way, it is likely to be a false dichotomy given their interdependence (Hof, 2008).

Our results suggest that push-off control corrects foot placement errors in steady-state walking. Push-off, however, also contributes to leg swing (Zelik & Adamczyk, 2016) and to the foot trajectory in the early swing phase. This is obvious from humans' responses to stepping target perturbations (Barton et al., 2019). Contrary to push-off control, swing leg control for foot placement is largely passive except at the beginning and end of the swing phase, as indicated by quick bursts of leg swing and retraction impulses (Doke et al., 2005). This passive ('predictable') pendular dynamics in the swing phase may hence be exploited by the feedforward/anticipative control of push-off for the desired foot placement. It is currently unknown whether in steady-state walking feedforward control of push-off is also used. As such is remains opaque whether push-off corrects foot placement error or push-off plans for the next foot placement or both. To further unravel the stabilizing mechanisms of push-off control, modelling approaches can elucidate when implementing feedforward and/or feedback push-off controllers based on foot placement errors for simple bipedal walkers under noisy conditions. Ryu & Kuo, (2021) showed pure feedback and pure feedforward control to be susceptible to sensor noise and process noise (e.g., uncertainly in the environment), respectively, and the combined feedforward and feedback controller was best for stable and robust walking.

VI. CONCLUSION

Push-off control has been suggested to modulate the deviations of CoM state with respect to desired values in perturbed human walking, leaving its relevance in steady-state walking an open issue. By testing the linear correlations between foot placement errors and kinetics time series that are indicative of push-off control, we found that (ankle) push-off torque corrects AP foot placement error both in steady-state normal and slow walking, with mean correlations from 30 participants up to R=0.45. With this we can confirm our hypothesis that push-off of the trailing leg corrects for AP foot placement errors at heel strike in steady-state walking to maintain a stable gait pattern. Our results suggest that humans use a push-off strategy to control CoM state and correct foot placement errors in steady-state walking.

VII. Acknowledgements

S.M.B. and J.J. are funded by a VIDI grant no. (016.Vidi.178.014) from the Dutch Organization for Scientific Research (NWO).

VIII. References

Afschrift, M., de Groote, F., & Jonkers, I. (2021). Similar sensorimotor transformations control balance during standing and walking. *PLoS Computational Biology*, *17*(6). https://doi.org/10.1371/journal.pcbi.1008369

Arvin, M., Hoozemans, M. J. M., Pijnappels, M., Duysens, J., Verschueren, S. M., & van Dieën, J. H. (2018). Where to step? Contributions of stance leg muscle spindle afference to planning of mediolateral foot placement for balance control in young and old adults. *Frontiers in Physiology*, *9*(1134). https://doi.org/10.3389/fphys.2018.01134

Barton, S. L., Matthis, J. S., & Fajen, B. R. (2019). Control strategies for rapid, visually guided adjustments of the foot during continuous walking. *Experimental Brain Research*, *237*(7). https://doi.org/10.1007/s00221-019-05538-7

Bhounsule, P. A. (2015). Control of a compass gait walker based on energy regulation using ankle push-off and foot placement. *Robotica*, *33*(6). https://doi.org/10.1017/S0263574714000745

Byl, K., & Tedrake, R. (2008). Approximate optimal control of the compass gait on rough terrain. In *Proceedings - IEEE International Conference on Robotics and Automation* (pp. 1258–1263). https://doi.org/10.1109/ROBOT.2008.4543376

De Leva, P. (1996). Adjustments to zatsiorsky-seluyanov's segment inertia parameters. *Journal of Biomechanics*, *29*(9). https://doi.org/10.1016/0021-9290(95)00178-6

Dingwell, J. B., & Cusumano, J. P. (2015). Identifying stride-to-stride control strategies in human treadmill walking. *PLoS ONE*, *10*(4). https://doi.org/10.1371/journal.pone.0124879

Doke, J., Donelan, J. M., & Kuo, A. D. (2005). Mechanics and energetics of swinging the human leg. *Journal of Experimental Biology*, *208*(3). https://doi.org/10.1242/jeb.01408

Franz, J. R. (2016). The Age-Associated Reduction in Propulsive Power Generation in Walking. *Exercise and Sport Sciences Reviews*, *44*(4). https://doi.org/10.1249/JES.0000000000000086

Friston, K. J., Ashburner, J. T., Kiebel, S. J., Nichols, T. E., & Penny, W. D. (2007). *Statistical Parametric Mapping: The Analysis of Functional Brain Images*. *Statistical Parametric Mapping The Analysis of Functional Brain Images* (Vol. 8).

Gruben, K. G., & Boehm, W. L. (2014). Ankle torque control that shifts the center of pressure from heel to toe contributes non-zero sagittal plane angular momentum during human walking. *Journal of Biomechanics*, *47*(6). https://doi.org/10.1016/j.jbiomech.2014.01.034

Hernández, A., Silder, A., Heiderscheit, B. C., & Thelen, D. G. (2009). Effect of age on center of mass motion during human walking. *Gait and Posture*, *30*(2). https://doi.org/10.1016/j.gaitpost.2009.05.006

Hobbelen, D. G. E., & Wisse, M. (2008). Ankle actuation for limit cycle walkers. *International Journal of Robotics Research*, *27*(6). https://doi.org/10.1177/0278364908091365

Hof, A. L. (2007). The equations of motion for a standing human reveal three mechanisms for balance. *Journal of Biomechanics*, *40*(2). https://doi.org/10.1016/j.jbiomech.2005.12.016

Hof, A. L. (2008). The "extrapolated center of mass" concept suggests a simple control of balance in walking. *Human Movement Science*, *27*(1). https://doi.org/10.1016/j.humov.2007.08.003

Joshi, V., & Srinivasan, M. (2019). A controller for walking derived from how humans recover from perturbations. *Journal of the Royal Society Interface*, *16*(157). https://doi.org/10.1098/rsif.2019.0027

Kelly, M., & Ruina, A. (2015). Non-linear robust control for inverted-pendulum 2D walking. In *Proceedings - IEEE International Conference on Robotics and Automation* (Vol. 2015-June). https://doi.org/10.1109/ICRA.2015.7139800

Kerimoglu, D., Karkoub, M., Ismail, U., Morgul, O., & Saranli, U. (2021). Efficient bipedal locomotion on rough terrain via compliant ankle actuation with energy regulation. *Bioinspiration & Biomimetics*, *16*(5). https://doi.org/10.1088/1748-3190/ac13b1

Kim, M., & Collins, S. H. (2013). Stabilization of a three-dimensional limit cycle walking model through step-to-step ankle control. In *IEEE International Conference on Rehabilitation Robotics* (pp. 1–6). https://doi.org/10.1109/ICORR.2013.6650437

Kim, M., & Collins, S. H. (2017). Once-Per-Step Control of Ankle Push-Off Work Improves Balance in a Three-Dimensional Simulation of Bipedal Walking. *IEEE Transactions on Robotics*, *33*(2). https://doi.org/10.1109/TRO.2016.2636297

Kingma, I., De Looze, M. P., Toussaint, H. M., Klijnsma, H. G., & Bruijnen, T. B. M. (1996). Validation of a full body 3-D dynamic linked segment model. *Human Movement Science*, *15*(6). https://doi.org/10.1016/S0167-9457(96)00034-6

Kuhman, D., & Hurt, C. P. (2019). The timing of locomotor propulsion in healthy adults walking at multiple speeds. *Human Movement Science*, *68*. https://doi.org/10.1016/j.humov.2019.102524

Kuo, A. D. (2002). Energetics of actively powered locomotion using the simplest walking model. *Journal of Biomechanical Engineering*, *124*(1). https://doi.org/10.1115/1.1427703

Lipfert, S. W., Günther, M., Renjewski, D., & Seyfarth, A. (2014). Impulsive ankle push-off powers leg swing in human walking. *Journal of Experimental Biology*, *217*(8). https://doi.org/10.1242/jeb.097345

McGibbon, C. A. (2003). Toward a better understanding of gait changes with age and disablement: Neuromuscular adaptation. *Exercise and Sport Sciences Reviews*, *31*(2). https://doi.org/10.1097/00003677-200304000-00009

Redfern, M. S., & Schumann, T. (1994). A model of foot placement during gait. *Journal of Biomechanics*, *27*(11). https://doi.org/10.1016/0021-9290(94)90043-4

Roerdink, M., Coolen, B. (H)., Clairbois, B. (H)E, Lamoth, C. J. C., & Beek, P. J. (2008). Online gait event detection using a large force platform embedded in a treadmill. *Journal of Biomechanics*, *41*(12). https://doi.org/10.1016/j.jbiomech.2008.06.023

Rossignol, S., Dubuc, R., & Gossard, J. P. (2006). Dynamic sensorimotor interactions in locomotion. *Physiological Reviews*, *86*(1). https://doi.org/10.1152/physrev.00028.2005

Ruina, A., Bertram, J. E. A., & Srinivasan, M. (2005). A collisional model of the energetic cost of support work qualitatively explains leg sequencing in walking and galloping, pseudo-elastic leg behavior in running and the walk-to-run transition. *Journal of Theoretical Biology*, *237*(2). https://doi.org/10.1016/j.jtbi.2005.04.004

Ryu, H. X., & Kuo, A. D. (2021). An optimality principle for locomotor central pattern generators. *Scientific Reports*, *11*(1). https://doi.org/10.1038/s41598-021-91714-1

Toney, M. E., & Chang, Y. H. (2016). The motor and the brake of the trailing leg in human walking: leg force control through ankle modulation and knee covariance. *Experimental Brain Research*, *234*(10). https://doi.org/10.1007/s00221-016-4703-8

van Leeuwen, A. M., van Dieën, J. H., Daffertshofer, A., & Bruijn, S. M. (2020). Active foot placement control ensures stable gait: Effect of constraints on foot placement and ankle moments. *Plos One*, *15*(12). https://doi.org/10.1371/journal.pone.0242215

van Leeuwen, A. M., van Dieën, J. H., Daffertshofer, A., & Bruijn, S. M. (2021). Ankle muscles drive mediolateral center of pressure control to ensure stable steady state gait. *Scientific Reports*, *11*(1). https://doi.org/10.1038/s41598-021-00463-8

van Leeuwen, M., Bruijn, S. M., van Dieën, J. H., & Daffertshofer, A. (2020). Dataset and analyses for: Active foot placement control ensures stable gait: Effect of constraints on foot placement and ankle moments. Zenodo. https://doi.org/https://doi.org/10.5281/zenodo.4229851

van Mierlo, M., Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2021). Centre of pressure modulations in double support effectively counteract anteroposterior perturbations during gait. *Journal of Biomechanics*, *126*. https://doi.org/10.1016/j.jbiomech.2021.110637

Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2016). Center of mass velocity-based predictions in balance recovery following pelvis perturbations during human walking. *Journal of Experimental Biology*, *219*(10). https://doi.org/10.1242/jeb.129338

Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2018a). Foot placement modulation diminishes for perturbations near foot contact. *Frontiers in Bioengineering and Biotechnology*, *6*(48). https://doi.org/10.3389/fbioe.2018.00048

Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2018b). Lower extremity joint-level responses to pelvis perturbation during human walking. *Scientific Reports*, *8*(1). https://doi.org/10.1038/s41598-018-32839-8

Wang, Y., & Srinivasan, M. (2014). Stepping in the direction of the fall: The next foot placement can be predicted from current upper body state in steady-state walking. *Biology Letters*, *10*(9). https://doi.org/10.1098/rsbl.2014.0405

Warren, J., Kay, B. A., Zosh, W. D., Duchon, A. P., & Sahuc, S. (2001). Optic flow is used to control human walking. *Nature Neuroscience*, *4*(2). https://doi.org/10.1038/84054

Welch, T. D. J., & Ting, L. H. (2008). A feedback model reproduces muscle activity during human postural responses to support-surface translations. *Journal of Neurophysiology*, *99*(2). https://doi.org/10.1152/jn.01110.2007

Williams, D. S., & Martin, A. E. (2019). Gait modification when decreasing double support percentage. *Journal of Biomechanics*, *92*. https://doi.org/10.1016/j.jbiomech.2019.05.028

Zaytsev, P., Wolfslag, W., & Ruina, A. (2018). The Boundaries of Walking Stability: Viability and Controllability of Simple Models. *IEEE Transactions on Robotics*, *34*(2). https://doi.org/10.1109/TRO.2017.2782818

Zehr, E. P., & Stein, R. B. (1999). What functions do reflexes serve during human locomotion? *Progress in Neurobiology*, *58*(2). https://doi.org/10.1016/S0301-0082(98)00081-1

Zelik, K. E., & Adamczyk, P. G. (2016). A unified perspective on ankle push-off in human walking. *Journal of Experimental Biology*, *219*(23). https://doi.org/10.1242/jeb.140376