Sagittal-plane balance perturbations during very slow walking
Strategies for recovering linear and angular momentum

van Mierlo, M.; Vlutters, M.; van Asseldonk, E. H.F.; van der Kooij, H.

DOI
10.1016/j.jbiomech.2023.111580

Publication date
2023

Document Version
Final published version

Published in
Journal of Biomechanics

Citation (APA)
van Mierlo, M., Vlutters, M., van Asseldonk, E. H. F., & van der Kooij, H. (2023). Sagittal-plane balance perturbations during very slow walking: Strategies for recovering linear and angular momentum. *Journal of Biomechanics*, *152*, Article 111580. https://doi.org/10.1016/j.jbiomech.2023.111580

Journal of Biomechanics 152 (2023) 111580

Contents lists available at ScienceDirect

Journal of Biomechanics

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

Sagittal-plane balance perturbations during very slow walking: Strategies for recovering linear and angular momentum

M. van Mierlo [a,*], M. Vlutters [a], E.H.F. van Asseldonk [a], H. van der Kooij [a,b]

[a] Department of Biomechanical Engineering, University of Twente, Enschede, The Netherlands
[b] Department of Biomechanical Engineering, Delft University of Technology, Delft, The Netherlands

ARTICLE INFO

Dataset link: http://dx.doi.org/10.4121/21370095

Keywords:
Human balance
Gait
Very slow walking
Whole-body angular momentum
Centre of pressure modulation
Ground reaction force vector

ABSTRACT

Spatiotemporal gait characteristics change during very slow walking, a relevant speed considering individuals with movement disorders or using assistive devices. However, we lack insights in how very slow walking affects human balance control. Therefore, we aimed to identify how healthy individuals use balance strategies while walking very slow. Ten healthy participants walked on a treadmill at an average speed of $0.43\,\mathrm{m\,s^{-1}}$, while being perturbed at toe off right by either perturbations of the whole-body linear momentum (WBLM) or angular momentum (WBAM). WBLM perturbations were given by a perturbation on the pelvis in forward or backward direction. The WBAM was perturbed by two simultaneous perturbations in opposite directions on the pelvis and upper body. The given perturbations had magnitudes of 4, 8, 12 and 16 % of the participant's body weight, and lasted for 150 ms. After perturbations of the WBLM the centre of pressure placement was modulated using the ankle joint, while keeping the moment arm of the ground reaction force (GRF) with respect to the centre of mass (CoM) small. After the perturbations of the WBAM a quick recovery was initiated, using the hip joint and adjusting the horizontal GRF to create a moment arm with respect to the CoM. These findings suggest no fundamental differences in the use of balance strategies at very slow walking compared to normal speeds. Still as the gait phases last longer, this time was exploited to counteract perturbations in the ongoing gait phase.

1. Introduction

Healthy individuals have an excellent ability to maintain balance. Many researchers explored human balance strategies to handle external perturbations during standing and walking (Madehkhaksar et al., 2018; Martelli et al., 2016; van Mierlo et al., 2022; Rietdyk et al., 1999; Vlutters et al., 2016; Young et al., 2012). However, little is known about the use of these strategies during very slow walking ($< 0.63\,\mathrm{m\,s^{-1}}$), this is a relevant speed considering individuals with movement disorders or using assistive devices (Nymark et al., 2005). Nowadays, individuals with a spinal cord injury using a powered lower limb exoskeleton have an average walking speed of about $0.26\,\mathrm{m\,s^{-1}}$ (Louie et al., 2015; Miller et al., 2016). At this low speed, various gait characteristics have shown changes, compared to normal walking (Smith and Lemaire, 2018; Otter et al., 2004). The use of balance recovery strategies might therefore change as well. To assist balance during very slow walking, a better understanding is needed of human balance strategies while walking at this low speed.

Spatiotemporal gait characteristics change when walking very slow (Lu et al., 2017; Smith and Lemaire, 2018; Wu et al., 2019). Step time, stride time and the time spent in double support (DS) increase (Wu et al., 2019). For example, when slowing down from $1.25\,\mathrm{m\,s^{-1}}$ to $0.63\,\mathrm{m\,s^{-1}}$ the time spent in DS doubles (Vlutters et al., 2016). Step lengths become shorter and the mediolateral (ML) centre of mass (CoM) excursion increases (Lu et al., 2017; Wu et al., 2019). Some of these characteristics are linear related to walking speed, also at very low speeds. However, an inflection point was found around $0.5\,\mathrm{m\,s^{-1}}$ for stride-, step-, stance- and DS time, presenting a stronger negative slope for the very low speeds (Smith and Lemaire, 2018). For walking speeds below $0.28\,\mathrm{m\,s^{-1}}$ the muscle activity of the peroneus longus and rectus femoris muscles increases again Otter et al. (2004) and also frontal plane ankle and hip joint moments increased again during very slow walking (Best and Wu, 2020). A possible explanation is that there is a higher demand on our muscles in order to maintain postural stability, to counteract gravity during the elongated swing phases and to compensate for a decreased ML margin of stability at this low walking speed (Otter et al., 2004; Best and Wu, 2020). It remains unclear whether our balance responses also change when walking very slowly.

* Corresponding author.
E-mail address: m.vanmierlo@utwente.nl (M. van Mierlo).

https://doi.org/10.1016/j.jbiomech.2023.111580
Accepted 4 April 2023
Available online 10 April 2023

Nomenclature

AP	Anteroposterior
BLP	Backward linear perturbation
BPP	Backward pitch perturbation
CoM	Centre of mass
CoP	Centre of pressure
DS	Double support
EndP	End perturbation
FLP	Forward linear perturbation
FPP	Forward pitch perturbation
GRF	Ground reaction force
HSL	Heel strike left
HSR	Heel strike right
ML	Mediolateral
QTM	Qualisys track manager
$r_{GRF-CoM}$	Moment arm of the GRF with respect to the CoM
SS	Single support
TOL	Toe off left
TOR	Toe off right
WBAM	Whole-body angular momentum
WBLM	Whole-body linear momentum

Whether slow walking increases or decreases stability compared to normal speeds is an open discussion (Bruijn et al., 2009; Dingwell and Marin, 2006; Reimann et al., 2018; Best and Wu, 2020; Hak et al., 2012). Reimann et al. (2018) suggest that increasing the cadence increases safety, because it shortens the falling phase between steps. This suggests that walking with a low cadence, which is usually the case for very slow walking, might decrease safety. The strategy of increasing cadence was shown by Hak et al. (2012) by applying ML treadmill perturbations. When balance was challenged, the participants continued walking with a higher step frequency, while walking speed did not increase since step length decreased (Hak et al., 2012). After ML pelvis perturbations individuals relied more on a hip strategy when walking very slow ($0.4\,\mathrm{m\,s^{-1}}$), compared to walking at $0.8\,\mathrm{m\,s^{-1}}$. Also, the use of a stepping strategy did not seem to be involved as much as during the faster walking speed (Matjačić et al., 2019). After AP pelvis perturbations healthy individuals have shown to follow the constant offset control of a simple linear inverted pendulum model during walking at $1.25\,\mathrm{m\,s^{-1}}$ and even better at $0.63\,\mathrm{m\,s^{-1}}$ (Hof, 2008; Vlutters et al., 2016). All these findings show that there is not one clear conclusion whether the ability of maintaining balance is decreased or increased by adjusting walking speed.

The above described pelvis perturbations result in a change of the whole-body linear momentum (WBLM). However, these are not the only type of perturbations we can encounter, other perturbations could also result in a change of the whole-body angular momentum (WBAM). The theoretical options to recover the WBLM and/or WBAM by modulations of the ground reaction force (GRF) direction and/or centre of pressure (CoP) are presented in Fig. 1. Previous research already showed some fundamental differences and priorities in balance recovery from perturbations of the WBLM and WBAM during walking at low and/or normal speeds (Vlutters et al., 2016; van Mierlo et al., 2022). Recovery of the WBAM is prioritized over WBLM regulation and achieved by modulations of the GRF direction directly after perturbations of the WBAM given at toe off right (TOR) (van Mierlo et al., 2022). Recovery from perturbations of the WBLM given at TOR was achieved through modulations of the CoP. These were achieved via ankle joint modulations since foot placement was not adjusted (Vlutters et al., 2016, 2018).

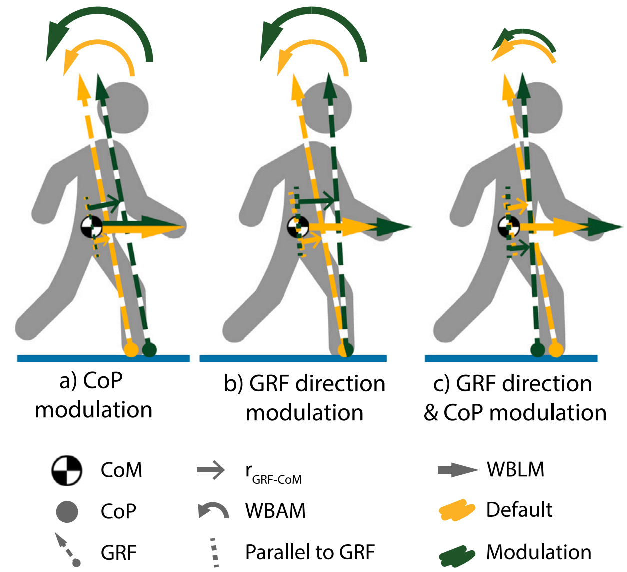

Fig. 1. Options for recovery of the WBLM and/or WBAM. A default configuration is presented in yellow, and the effects of GRF direction and/or CoP modulations are presented in green. a) CoP modulation: the more forward CoP results in an increase of the $r_{GRF-CoM}$. This results in an increase of the WBAM. b) GRF direction modulation: another method increasing the $r_{GRF-CoM}$ resulting in an increase of the WBAM. This GRF direction modulation also changes the horizontal GRF, and thereby the WBLM as well. c) Combining the GRF direction and CoP modulation: this can result in the condition where the $r_{GRF-CoM}$ remains similar and only the horizontal GRF changes, and thereby the WBLM, but not the WBAM.

We aimed to identify how healthy individuals use balance strategies after perturbations of the WBLM and WBAM during very slow walking at $0.44\,\mathrm{m\,s^{-1}}$. This speed was selected because (van Hedel, 2009) showed this to be a threshold for independent community ambulation using an aid, such as a lower limb exoskeleton. For speeds below this threshold, the aiding solution might not be preferred over a wheelchair. With respect to normal walking, this very low walking speed increases the time spent in DS and single support (SS). Therefore, we hypothesized that a larger part of the recovery will already be done in the phase directly after the perturbations given at TOR, which was SS in our study. This results in the use of balance strategies that can be specifically used during this phase, like the ankle and hip strategy. We also hypothesized that minimal foot placement adjustments were used at the following heel contact, since this was also not reported for WBLM and WBAM perturbations during slow and normal walking. Modulations of the CoP and GRF direction below the stance foot were hypothesized to be similar to those reported during walking at low or normal speeds.

2. Methods

Data was used from the same set of experiments performed for van Mierlo et al. (2022), more details can be found in this paper.

2.1. Participants

Ten healthy volunteers (age 24 ± 3 year) participated in the study (van Mierlo et al., 2022). The EWI/ET ethics committee of the University of Twente approved the experimental setup and protocol with reference number RP 2019–88. All participants gave written informed consent prior to the experiment in accordance with the Declaration of Helsinki.

2.2. Setup

The experiments were performed on a dual-belt instrumented treadmill (custom Y-Mill, Motekforce Link, The Netherlands). Two motors (SMH60, Moog, The Netherlands) on the rear were used to apply perturbations to the upper body and/or pelvis via horizontal rods attached to braces worn by the participants. A safety harness was worn to secure the participant in case of a fall. An admittance controller (described in van der Kooij et al., 2021) was used to minimize the interaction forces during walking and to track the desired forces at the instant a perturbation was given.

2.3. Data collection

Kinematic data were recorded with an 8-camera Qualisys motion capture system (Oqus 600+, Qualisys, Göteborg, Sweden). Ten rigid bodies containing four reflective markers each and 33 single markers were placed on the body segments and bony landmarks respectively. Motion capture data was recorded at 128 Hz with Qualisys Track Manager (QTM, Qualisys, Göteborg, Sweden). Via an analog interface data from the force plates was received at 2048 Hz in sync with the motion capture data. Data from the load cells integrated in the horizontal rods were recorded at 1000 Hz via the main computer controlling the motor, and synchronized with the other data via a synchronization signal.

2.4. Experimental protocol

The participants walked on the treadmill at a very low speed scaled to their leg length ($\sqrt{l} \cdot 0.44\,\mathrm{m\,s^{-1}}$). The experiment was divided into three blocks. The first block consisted of 3 minutes unperturbed walking to set a baseline. During the second and third block the participants received either: 1) perturbations of the WBLM, consisting of a perturbation on the pelvis (backward linear perturbation (BLP) and forward linear perturbation (FLP)); or 2) perturbations of the WBAM, consisting of two simultaneous perturbations of the same magnitude in opposite direction on the pelvis and upper body (forward pitch perturbation (FPP) rotating the upper body forward and backward pitch perturbation (BPP) rotating the upper body backward). The order of the second and third block was randomized between the participants. All perturbations were given at the moment of TOR and lasted for 150 ms. The individual perturbations had a magnitude of 4 , 8 , 12 , or 16 % of body weight. Each combination of magnitude and direction was repeated 6 times, resulting in 48 perturbations per perturbation type (linear or pitch), which were given in a completely randomized order. Between each perturbation there was a random interval ranging from 3 to 6 strides.

2.5. Data processing

QTM software was used to label the markers and interpolate the missing samples of the marker data with the polynomial gap filling tool. Processing of the data was done with Matlab (R2019b, MathWorks). The marker and force data were filtered with a 6th order zero phase low pass Butterworth filter with a cut off frequency based on the participant's cadence ($cadence \cdot 6.25$ Hz) (Rácz and Kiss, 2021). OpenSim 4.2 (Delp et al., 2007) was used to scale a full body model (Rajagopal et al., 2016) for each participant. The OpenSim inverse kinematics, inverse dynamics, and analyze tool were used to derive the joint moments and CoM positions, velocities and orientations of the total body and individual segments. The GRFs and moments were used to calculate the CoP positions and the horizontal component of the GRF. The moment arm ($r_{GRF-CoM}$) of the GRF was calculated with respect to the whole-body CoM position. The WBAM was calculated with Eq. (1), where i presents each body segment of the OpenSim model, x^i_{CoM} and x_{CoM} the position of each ith segment and the whole-body CoM respectively, m_i each ith segment's mass, $\dot{x}^i_{CoM}$ and $\dot{x}_{CoM}$ the velocity of each ith segment and the whole-body CoM respectively, I^i each ith segment's inertia tensor, and ω^i the angular velocity about the ith segment's CoM (Herr and Popovic, 2008). All measures are given in the global frame and in anteroposterior (AP) direction or about the ML axis.

$$H = \sum_{i=1}^{22} [(x^i_{CoM} - x_{CoM}) \times m_i(\dot{x}^i_{CoM} - \dot{x}_{CoM}) + I^i \omega^i] \tag{1}$$

All measures were scaled for the individual participants using a scaling factor based on the participant's leg length, mass and/or height (depending on the measure), making the values dimensionless (Hof, 1996). To bring the values back to the original order of magnitude, the scaled measures were multiplied with a measure-specific scaling factor calculated from the average leg length, mass and/or height over all participants. All measures were normalized over time by resampling each (gait) phase to 50 samples, synchronizing the instants of toe off, heel strike and the end of the perturbation (EndP). This allowed for averaging the data over all repetitions within each participant and across all participants.

2.6. Outcome measures

The outcome measures were defined as the maximal deviation of a measure due to the perturbation with respect to the baseline value, which was recorded during unperturbed walking. This maximal deviation was taken at an instant during the left SS, between EndP and the instant the measure crossed the baseline value (if this latter occurred). The WBLM is the body's mass times the velocity and since the mass does not change we used the linear CoM velocity and the indicator of WBLM. Another outcome measure was the relationship between the CoM velocity at heel strike right (HSR) as an independent variable and the distance between the CoP and CoM at toe off left (TOL) as a dependent variable, since this relationship has shown good predictive values before during higher walking speeds (Vlutters et al., 2016). Linear least squares fits were made to examine this relationship. To analyze whether the perturbations affected foot placement, the AP distance was taken between the CoM and the lateral malleolus of the right foot at HSR. The last outcome measures were the durations of the first SS and DS after the perturbation. Averages were taken over all repetitions and participants.

2.7. Statistics

Linear mixed models were used to evaluate the dependence of the outcome measures on the perturbations. The statistical analysis was performed in R4.1.2 (R. Core Team, 2021, Vienna, Austria). Linear mixed models were fitted based on a maximum-likelihood estimation for each of the following outcome measures: WBAM, CoM velocity, $r_{GRF-CoM}$ and CoP position with respect to the CoM, horizontal GRF, joint moments of the left hip and ankle, foot placement at HSR and gait phase durations of the first SS and DS. Separate models were made for linear and pitch perturbations. The AIC, BIC and likelihood-ratio test were used to select the final model structure (Bates et al., 2015). The final structure included perturbation magnitude (continuous variable) and perturbation direction (categorical variable) as fixed effects together with their interaction. Random effects for the intercept and slope were added to take into account the participant effects. Baseline measurements were included in the models for gait phase duration. The residuals were checked for normality and heteroscedasticity. The main effects were tested with a significance level of $\alpha = 0.05$ using the Wald t-test with a Kenward-Roger correction for the degrees of freedom.

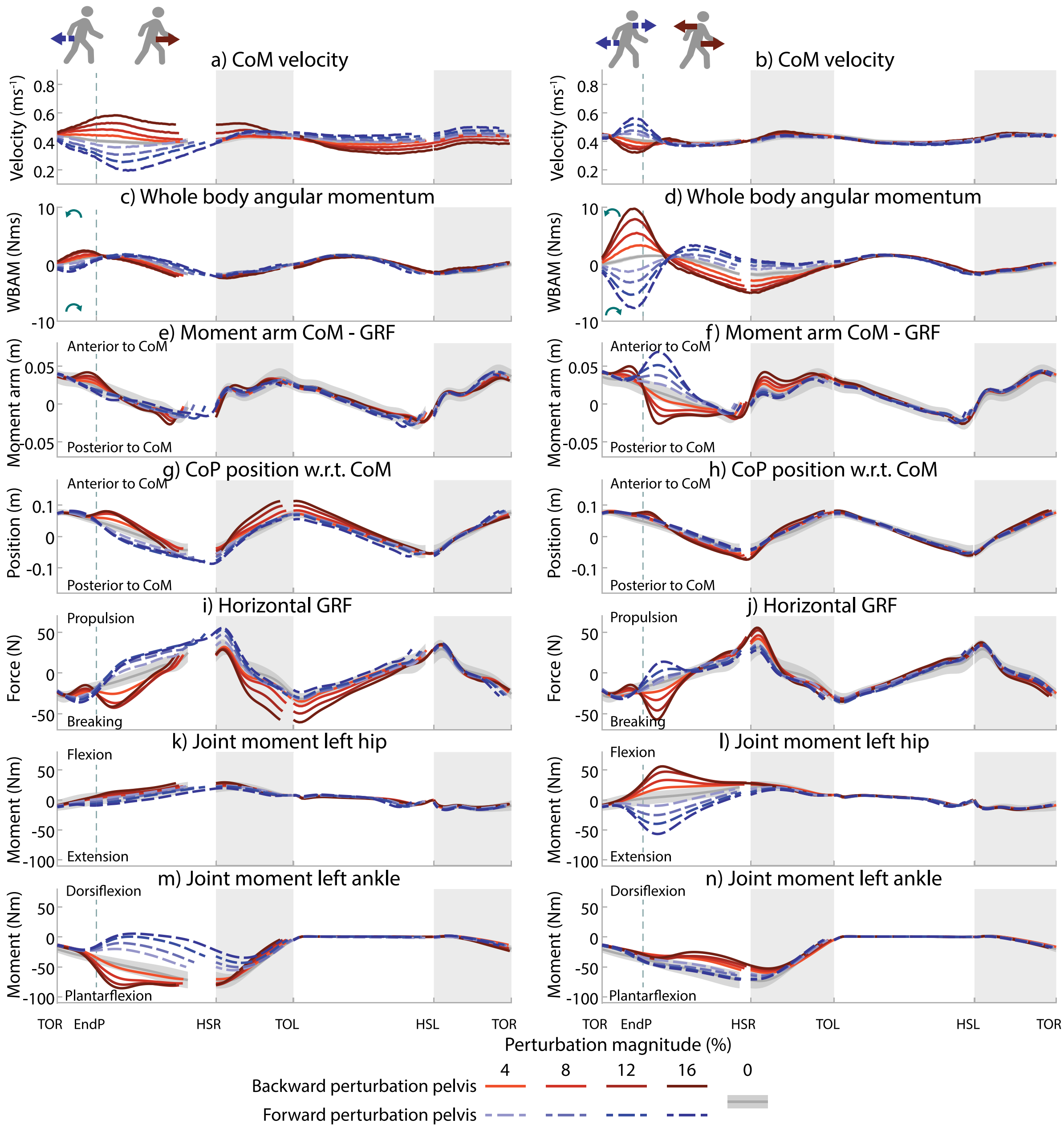

Fig. 2. Averaged time series of one gait cycle from TOR to TOR. The perturbation starts at the first TOR and ends at EndP. The data of the different perturbation magnitudes is aligned at the start of the marked events: heel strike right and left (HSR, HSL) and toe of left (TOL). Shaded areas indicate DS. In the left column the results are presented of the linear perturbations, with the BLP in blue and FLP in red. The right column presents the pitch perturbations with the FPP presented in blue and BPP in red. Unperturbed walking is presented in gray with an area covering the standard deviation. The $r_{GRF-CoM}$ and the CoP position are both taken with respect to the CoM.

3. Results

3.1. Perturbation effect

The two different perturbation types (linear and pitch) and directions (forward and backward) affected the WBAM and WBLM differently. The linear perturbations significantly affected the CoM velocity during SS directly after the perturbation (Figs. 2a & 3a), while the WBAM was hardly affected by these perturbations (Figs. 2c & 3b). Details of all the statistical outcomes can be found in Table 1. As intended the pitch perturbations had a significant effect on the WBAM (Figs. 2b & 3b). The CoM velocity was also affected, however this could also have been caused by a measurement artifact, since the total external horizontal forces sum up to zero.

3.2. Balance recovery

A significant change in the horizontal GRF helped in recovering from linear perturbations during SS, by increasing the backward GRF in case of a FLP or increasing the forward GRF after a BLP (Figs. 2i & 3e). This occurred together with a CoP shift (Figs. 2g & 3d), such that the $r_{GRF-CoM}$ remained small (Figs. 2e & 3c). After perturbations of the WBAM the horizontal GRF also changed significantly (Figs. 2j & 3e), but together with minimal changes of the CoP position (Figs. 2h & 3d). This created a significant change of the $r_{GRF-CoM}$ (Figs. 2f & 3c), such that the GRF assisted in recovering the perturbed WBAM. Most effects of the initial disturbance seems to be negated at the end of the first DS after the perturbation. However, for the linear perturbations deviations of the CoM velocity, CoP position and horizontal GRF were still present throughout the whole gait cycle after the perturbation. At

Table 1
Estimated model parameters and model fits of the linear mixed models for the different outcome variables and perturbation types (LP = linear perturbations & PP = pitch perturbations). The default direction is either the backward perturbation for the linear perturbation type, and forward pitch perturbation for the pitch perturbation type. All models except those for foot placement and the SS and DS duration had an N-value of 80 (4 magnitudes, 2 directions and 10 participants). The models for foot placement and the SS and DS duration had an N-value of 100 (5 magnitudes, 2 directions and 10 participants).

	CoM velocity ($m\,s^{-1}$)		Angular momentum (N m s)		Moment arm $r_{GRF-CoM}$ (m)	
Fixed effect	LP	PP	LP	PP	LP	PP
Intercept	0.014	−0.014	0.257	0.718	−0.001	0.000
Magnitude	−0.014 ***	0.009 ***	−0.041	−0.535 ***	−0.001	0.004 ***
Direction	−0.019	0.003	−0.463	−0.676	0.000	0.009
Magnitude*Direction	0.027 ***	−0.012 ***	0.031	0.987 ***	0.001 ***	−0.007 ***
Model fit						
R^2-marginal	0.99	0.62	0.01	0.87	0.21	0.79
R^2-conditional	0.99	0.67	0.22	0.87	0.80	0.82
	CoP (m)		Horizontal GRF (N)		Hip joint moment (N m)	
Fixed effect	LP	PP	LP	PP	LP	PP
Intercept	−0.018 ***	0.010	8.192	−8.852	4.914 *	−0.617
Magnitude	−0.002 ***	0.001 **	1.521 ***	2.245 **	−1.144 ***	−4.032 ***
Direction	0.040 ***	−0.013	−23.649 ***	10.562	−7.532 **	4.224
Magnitude*Direction	0.004 ***	0.001	−2.829 ***	−4.901 ***	1.911 ***	7.326 ***
Model fit						
R^2-marginal	0.94	0.20	0.79	0.56	0.55	0.98
R^2-conditional	0.97	0.51	0.89	0.70	0.79	0.99
	Ankle joint moment (N m)		Single support time (s)		Double support time (s)	
Fixed effect	LP	PP	LP	PP	LP	PP
Intercept	21.805 ***	10.602 *	0.482 ***	0.492 ***	0.305 ***	0.292 ***
Magnitude	2.600 ***	−0.998 *	0.007 ***	0.001	−0.003 **	−0.002 *
Direction	−15.479 ***	9.590	0.011	0.013	−0.014 *	0.008
Magnitude*Direction	−5.777 ***	1.304 **	−0.010 ***	0.002 *	−0.001	0.000
Model fit						
R^2-marginal	0.95	0.33	0.35	0.05	0.21	0.09
R^2-conditional	0.97	0.80	0.87	0.84	0.77	0.69
	Foot placement (cm)					
Fixed effect	LP	PP				
Intercept	16.74 ***	17.31 ***				
Magnitude	0.04	0.11				
Direction	0.46	−0.63				
Magnitude*Direction	−0.23 ***	−0.34 ***				
Model fit						
R^2-marginal	0.08	0.18				
R^2-conditional	0.67	0.85				

*The statistical significance of $p<0.05$.
**The statistical significance of $p < 0.01$.
***The statistical significance of $p < 0.001$.

HSR the first foot placement after the perturbation was only significant affected for the BLP and FPP. The leading foot was placed closer to the CoM (Table 1).

3.3. Hip and ankle joint contributions

Created moments of the hip and ankle joint contribute to achieving the obtained changes in GRF, CoP and $r_{GRF-CoM}$. The left ankle joint strongly contributed to recovering from linear perturbations by applying a plantarflexion moment after the FLP and a dorsiflexion moment after the BLP (Figs. 2m and 3g). A significant linear trend was observed for the hip moments as well (Figs. 2k and 3f), but the deviations were small. For the pitch perturbations an opposite pattern was observed. The hip joint had a significant strong contribution (Figs. 2l and 3f). The ankle joint was significantly involved as well after these perturbations, however the slope was small and there was a large variation among the participants (Fig. 3g).

3.4. CoM velocity relationships

Linear least square fits have been made to the four different perturbation conditions (2x type and 2x direction) to see if the CoP position at TOL was related to the CoM velocity at the previous HSR (Fig. 4). Only for the FLP this resulted in a good relationship ($y = 0.35x - 0.07$, $R^2 = 0.96$). The slope of this line also approximates the average proportionality constant over all participants: $\omega_0^{-1} = 0.31$ s, defined as the reciprocal of the eigenfrequency (Hof, 2008; Vlutters et al., 2016). For the BLP, and even stronger for the pitch perturbations, there was not that much variation (anymore) in the CoM velocity at HSR, resulting in the lack of strength ($R^2 < 0.9$) for this relationship.

3.5. Gait phase durations

All perturbation conditions significantly affected the duration of the SS and/or DS directly after the perturbation (Fig. 5 and Table 1). The strongest effects were seen for the linear perturbations and during the SS, which was the phase in which the perturbations were given. The SS was prolonged for the BLP and shortened for the FLP. For both these perturbation directions the SS was followed by a shortened DS.

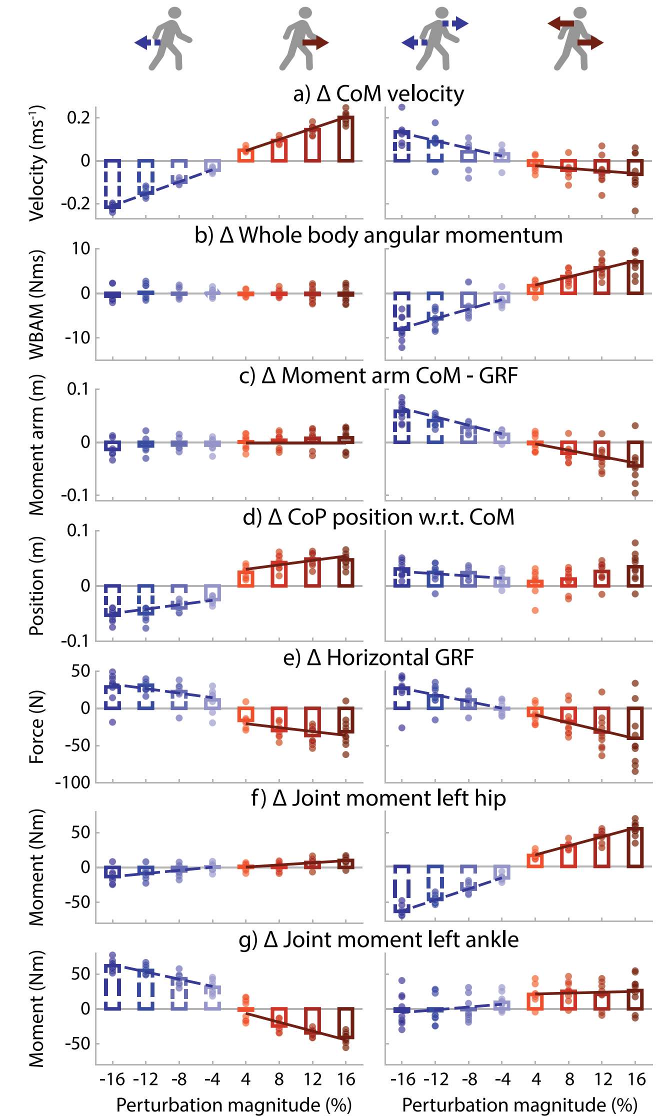

Fig. 3. Barplots presenting the maximum difference of the perturbed condition with respect to baseline during SS for the different outcome measures. The left column presents the results for the linear perturbations and the right column for the pitch perturbations. The dots present the averages per participant. Fitted models are included with a line if the perturbation magnitude or the interaction between thee perturbation magnitude and direction had a statistically significant effect ($p < 0.05$) on the outcome measure. The $r_{GRF-CoM}$ and the CoP position are both taken with respect to the CoM.

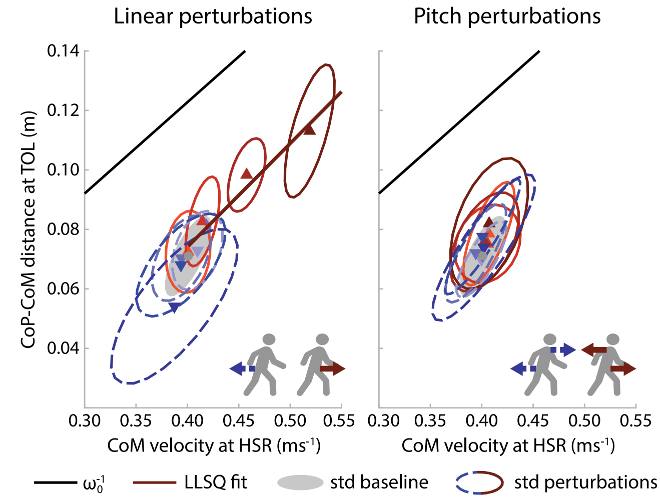

Fig. 4. Results of the linear least square fits (LLSQ) on the relationship between the CoM velocity at HSR and the CoP-CoM distance at TOL for the two different perturbation types and directions. The triangles indicate the average values over all participants with the ellipse indicating the standard deviation (std). The std of the unperturbed condition(baseline) is presented with a filled gray ellipse. The red line presents the linear least square fit for the FLP, which had an $R^2 > 0.9$. The black lines indicate the slope of the proportionality constant ω_0^{-1}.

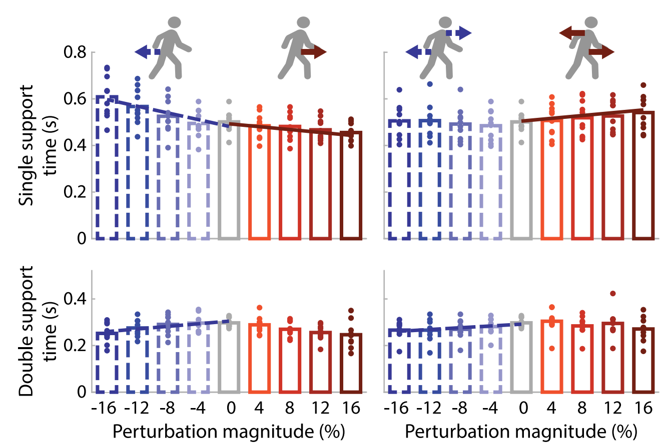

Fig. 5. Barplot presenting the gait phase durations of the first SS (top-row) and DS (bottom-row) for the linear (left column) and pitch (right column) perturbations. The dots indicate the averages for the individual participants. Lines are included if there was a significant relationship (p < 0.05) between the perturbation magnitude and direction and the gait phase duration.

Changes in the gait phase durations where less prominent after the pitch perturbations. The SS duration increased significantly for the BPP and the FPP only showed a small but significant shortening of the DS duration.

4. Discussion

We aimed to identify how balance strategies are used when walking at a very low speed when either the whole-body linear or angular momentum is perturbed in the sagittal plane. As hypothesized the majority of the perturbation was rejected during the first SS after the perturbation, especially for the pitch perturbations. Compared to normal walking speeds, during very slow walking more time is spend in each of the gait phases, allowing a further recovery earlier in the gait cycle.

The BLPs concerned perturbations given in a direction opposite to the walking direction. Therefore, more effort might have been used in compensating for these disturbances, with respect to FLP (Martelli et al., 2016). These perturbations resulted in an elongated SS duration, along with a foot placement closer to the CoM. This entailed that also a larger part of the disturbed CoM velocity could be recovered during this phase, especially compared to similar perturbations at a low or normal walking speed (Vlutters et al., 2018). Others have also shown that backward pulls elicit a fast and strong balance response in terms of muscle activations and generated joint moments (Misiaszek, 2003; Vlutters et al., 2018). The increased duration of SS, was followed by a reduced DS. This might be needed for continuing walking in the middle of the fixed speed treadmill. However, similar results were reported by Zadravec et al. (2017) during overground walking at $0.85\,\mathrm{m\,s^{-1}}$, suggesting that this was not just due to treadmill walking. Also for low or normal speeds the DS phase did not decrease after similar perturbations during treadmill walking in the study of Vlutters et al. (2016). Even though the CoM velocity disturbance was restored quickly and the deviations in the CoP position and horizontal GRF were small during the following DS, the ankle plantarflexion moment of the stance leg was still reduced during this phase. This effect could also be seen during slow and normal walking in the results of Vlutters et al. (2018).

FLPs increase the CoM velocity in the walking direction. If this does not induce an acute risk for falling, there might be no need for a quick and early recovery. A braking strategy was initiated during SS by increasing the horizontal GRF in backward direction together with a more forward positioned CoP with respect to the CoM. Since the CoM velocity was not returned to the unperturbed condition at the end of SS, this strategy was retained throughout DS, without foot placement adjustment. For the braking strategy the DS phase could efficiently be used because of the leading leg being able to create this backward GRF (van Mierlo et al., 2021). Overall the strategies used after FLP were quite similar to those reported during normal ($1.25\,\mathrm{m\,s^{-1}}$) and slow ($0.63\,\mathrm{m\,s^{-1}}$) walking (Vlutters et al., 2016, 2018).

The FPP and BPP provoked a quick response resulting in a GRF passing anterior or posterior from the CoM respectively, helping in counteracting the disturbed WBAM. This resulted in an overcompensation of the WBAM before HSR. This overshoot was gradually resolved over the following DS. A similar strategy to counteract these pitch perturbations was seen during walking with a normal walking velocity (van Mierlo et al., 2022). Very small adjustments were made to the CoP position, while the horizontal GRF was adjusted by using the hip joint of the stance leg. Unlike our hypothesis, the FPP did result in a foot placement adjustment. After BPP at TOR the ankle plantarflexion moment of the stance foot was reduced during very slow walking, which was not the case for a normal walking speed (van Mierlo et al., 2022). This might have reduced the propulsion resulting in the elongated SS after these perturbations.

Only the FLP resulted in a detectable relationship between the CoM velocity at HSR and the CoP position at the following TOL, like during slow and normal walking (Vlutters et al., 2016). Compared to the other perturbation conditions in our study, this was the only condition for which there was still a deviation in the CoM velocity with respect to baseline at HSR, which is needed to establish this relationship. Even though there was no relationship between CoM velocity at HSR and the CoP position at the following TOL for the other conditions, it does follow the expectations according to the linear inverted pendulum model. Since the pitch perturbations did not intent to perturb the CoM velocity, also no change in CoP position was expected based on this model (Hof, 2008; Vlutters et al., 2016). For the BLP the CoM velocity was already recovered at HSR. If these perturbations would have been larger or given at a different timing, a similar relationship is expected to be found.

A point to be considered is that walking at a very low speed feels unnaturally for healthy individuals. As reported by others as well, it increases the variability of many spatiotemporal gait characteristics (Wu et al., 2019). Probably because there is more thinking time, it makes you more conscious, resulting in a higher variability. This high variability is seen in the individual responses shown as the dots in Fig. 3. Some of the participants seem to have used a different response. Despite this effect, trends and balance recovery strategies could be distinguished that helped in recovering balance during very slow walking. The use of a treadmill can be considered as another limitation, due to the imposed walking speed. However, Zadravec et al. (2017) has shown similar responses to pelvis perturbations given during treadmill and overground walking. Finally, we only applied perturbations at TOR. It is an open question whether the same effects will be seen for perturbations given at other instances of the gait cycle.

Our findings revealed that strategies used during very slow walking are similar to those used during walking at a normal or low velocity (van Mierlo et al., 2022; Vlutters et al., 2016, 2018). The current results show a trend suggesting that also for walking very slowly, at $0.44\,\mathrm{m\,s^{-1}}$, still the CoP modulation together with a modulation of the GRF might be the most important recovery strategy after perturbations of the WBLM. This was done by using an ankle strategy during SS and for FLP also during DS. Perturbations of the WBAM are mainly restored from the hip joint and horizontal GRF adjustments without CoP modulation. These findings might give opportunities for extrapolating balance responses for different walking velocities. The results also suggest that the established relationship between the CoM velocity at HSR and CoP modulation in DS, based on the linear inverted pendulum model, could be used to predict where the CoP should be placed in order to maintain balance during very slow walking. However, it should not be the only measure indicating balance, since it does not reflect perturbations of the WBAM. The main difference in balance recovery during very slow walking compared to the recovery during walking with a low or normal walking speed is the time spent in the swing phase directly after the perturbations given at toe off and therefore the opportunities for the use of the different balance recovery strategies. To conclude, during very slow walking perturbations given at toe off can be handled mostly during the ongoing single support phase by using the hip and or ankle of the stance leg to recover a large part of the disturbance.

CRediT authorship contribution statement

M. van Mierlo: Conceptualization, Methodology, Investigation, Software, Formal analysis, Writing – original draft, Visualization. **M. Vlutters:** Conceptualization, Methodology, Software, Writing – review & editing. **E.H.F. van Asseldonk:** Conceptualization, Methodology, Writing – review & editing, Supervision. **H. van der Kooij:** Conceptualization, Methodology, Writing – review & editing, Supervision, Project administration, Funding acquisition.

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.

Data availability

The data used for this study is published on 4TU.ResearchData and can be found via the following DOI: http://dx.doi.org/10.4121/21370095.

Acknowledgments

This work is part of the research program Wearable Robotics with project number P16-05, which is (partly) funded by the Dutch Research Council (NWO).

Appendix A. Supplementary data

Supplementary material related to this article can be found online at https://doi.org/10.1016/j.jbiomech.2023.111580.

References

Bates, D., Mächler, M., Bolker, B.M., Walker, S.C., 2015. Fitting linear mixed-effects models using LME4. J. Stat. Softw. 67, http://dx.doi.org/10.18637/jss.v067.i01.

Best, A.N., Wu, A.R., 2020. Upper body and ankle strategies compensate for reduced lateral stability at very slow walking speeds. Proc. R. Soc. B 287, 20201685. http://dx.doi.org/10.1098/rspb.2020.1685.

Bruijn, S.M., van Dieën, J.H., Meijer, O.G., Beek, P.J., 2009. Is slow walking more stable? J. Biomech. 42, 1506–1512. http://dx.doi.org/10.1016/j.jbiomech.2009.03.047.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: Open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 54, 1940–1950. http://dx.doi.org/10.1109/TBME.2007.901024.

Dingwell, J.B., Marin, L.C., 2006. Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. J. Biomech. 39, 444–452. http://dx.doi.org/10.1016/j.jbiomech.2004.12.014.

Hak, L., Houdijk, H., Steenbrink, F., Mert, A., van der Wurff, P., Beek, P.J., van Dieën, J.H., 2012. Speeding up or slowing down?: Gait adaptations to preserve gait stability in response to balance perturbations. Gait & Posture 36, 260–264. http://dx.doi.org/10.1016/j.gaitpost.2012.03.005.

Herr, H., Popovic, M., 2008. Angular momentum in human walking. J. Exp. Biol. 211, 467–481. http://dx.doi.org/10.1242/jeb.008573.

Hof, A.L., 1996. Scaling gait data to body size. Gait & Posture 4, 222–223. http://dx.doi.org/10.1016/0966-6362(95)01057-2.

Hof, A.L., 2008. The 'extrapolated center of mass' concept suggests a simple control of balance in walking. Hum. Mov. Sci. 27, 112–125. http://dx.doi.org/10.1016/j.humov.2007.08.003.

Louie, D.R., Eng, J.J., Lam, T., 2015. Gait speed using powered robotic exoskeletons after spinal cord injury: A systematic review and correlational study. J. NeuroEng. Rehabil. 12, http://dx.doi.org/10.1186/s12984-015-0074-9.

Lu, H., Lu, T., Phil, D., Lin, H., Hsieh, H., Chan, W.P., 2017. Effects of belt speed on the body's center of mass motion relative to the center of pressure during treadmill walking. Gait & Posture 51, 109–115. http://dx.doi.org/10.1016/j.gaitpost.2016.09.030.

Madehkhaksar, F., Klenk, J., Sczuka, K., Gordt, K., Melzer, I., Schwenk, M., 2018. The effects of unexpected mechanical perturbations during treadmill walking on spatiotemporal gait parameters, and the dynamic stability measures by which to quantify postural response. PLoS One 13, 1–15. http://dx.doi.org/10.1371/journal.pone.0195902.

Martelli, D., Vashista, V., Micera, S., Agrawal, S.K., 2016. Direction-dependent adaptation of dynamic gait stability following waist-pull perturbations. IEEE Trans. Neural Syst. Rehabil. Eng. 24, 1304–1313. http://dx.doi.org/10.1109/TNSRE.2015.2500100.

Matjačić, Z., Zadravec, M., Olenšek, A., 2019. Influence of Treadmill Speed and Perturbation Intensity on Selection of Balancing Strategies During Slow Walking Perturbed in the Frontal Plane, Vol. 2019. http://dx.doi.org/10.1155/2019/1046459, Hindawi.

Miller, L.E., Zimmermann, A.K., Herbert, W.G., 2016. Clinical effectiveness and safety of powered exoskeleton-assisted walking in patients with spinal cord injury: Systematic review with meta-analysis. Med. Dev. (Auckland, N.Z.) 9, 455–466. http://dx.doi.org/10.2147/MDER.S103102.

Misiaszek, J.E., 2003. Early activation of arm and leg muscles following pulls to the waist during walking. Exp. Brain Res. 151, 318–329. http://dx.doi.org/10.1007/s00221-003-1501-x.

Nymark, J.R., Balmer, S.J., Melis, E.H., Edward, D., Millar, S., 2005. Electromyographic and kinematic nondisabled gait differences at extremely slow overground and treadmill walking speeds. J. Rehabil. Res. Dev. 42, 523–534. http://dx.doi.org/10.1682/JRRD.2004.05.0059.

Otter, A.R.D., Geurts, A.C.H., Mulder, T., Duysens, J., 2004. Speed related changes in muscle activity from normal to very slow walking speeds, 19. pp. 270–278. http://dx.doi.org/10.1016/S0966-6362(03)00071-7.

R. Core Team, 2021. R: A Language and Environment for Statistical Computing. R Foundation for statistical Computing, Vienna, Austria, URL: https://www.R-project.org/.

Rácz, K., Kiss, R.M., 2021. Marker displacement data filtering in gait analysis: A technical note. Biomed. Signal Process. Control 70, http://dx.doi.org/10.1016/j.bspc.2021.102974.

Rajagopal, A., Dembia, C.L., Demers, M.S., Delp, D.D., Hicks, J.L., Delp, S.L., 2016. Full body musculoskeletal model for muscle-driven simulation of human gait. IEEE Trans. Biomed. Eng. 63, 2068–2079. http://dx.doi.org/10.1109/TBME.2016.2586891.Full.

Reimann, H., Fettrow, T., Jeka, J.J., 2018. Strategies for the control of balance during locomotion. Kinesiol. Rev. 7, 18–25. http://dx.doi.org/10.1123/kr.2017-0053.

Rietdyk, S., Patla, A.E., Winter, D.A., Ishac, M.G., Little, C.E., 1999. Balance recovery from medio-lateral perturbations of the upper body during standing. J. Biomech. 32, 1149–1158.

Smith, A.J., Lemaire, E.D., 2018. Temporal-spatial gait parameter models of very slow walking. Gait & Posture 61, 125–129. http://dx.doi.org/10.1016/j.gaitpost.2018.01.003.

van der Kooij, H., Fricke, S.S., van t' Veld, R.C., Prieto, A.V., Keemink, A.Q.L., Schouten, A.C., van Asseldonk, E.H.F., 2021. Identification of hip and knee joint impedance during the swing phase of walking. IEEE Trans. Neural Syst. Rehabil. Eng. 30, http://dx.doi.org/10.1109/TNSRE.2022.3172497.

van Hedel, J.A., 2009. Gait speed in relation to categories of functional ambulation after spinal cord injury. Neurorehabil. Neural Repair 23, 343–350. http://dx.doi.org/10.1177/1545968308324224.

van Mierlo, M., Ambrosius, J., Vlutters, M., van Asseldonk, E., van der Kooij, H., 2022. Recovery from sagittal-plane whole body angular momentum perturbations during walking. J. Biomech. 141, 111169. http://dx.doi.org/10.1016/j.jbiomech.2022.111169.

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. J. Biomech. 126, 110637. http://dx.doi.org/10.1016/j.jbiomech.2021.110637.

Vlutters, M., van Asseldonk, E.H.F., van der Kooij, H., 2018. Lower extremity joint-level responses to pelvis perturbation during human walking. Sci. Rep. 8, 14621. http://dx.doi.org/10.1038/s41598-018-32839-8.

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. J. Exp. Biol. 219, 1514–1523. http://dx.doi.org/10.1242/jeb.129338.

Wu, A.R., Simpson, C.S., van Asseldonk, E.H.F., van der Kooij, H., Ijspeert, A.J., 2019. Mechanics of very slow human walking. Sci. Rep. 1–10. http://dx.doi.org/10.1038/s41598-019-54271-2.

Young, P.M.M., Wilken, J.M., Dingwell, J.B., 2012. Dynamic margins of stability during human walking in destabilizing environments. J. Biomech. 45, 1053–1059. http://dx.doi.org/10.1016/j.jbiomech.2011.12.027.

Zadravec, M., Olenšek, A., Matjačić, Z., 2017. The comparison of stepping responses following perturbations applied to pelvis during overground and treadmill walking. Technol. Health Care 25, 781–790. http://dx.doi.org/10.1007/978-3-319-46669-9_50.