



An effective balancing response to lateral perturbations at pelvis level during slow walking requires control in all three planes of motion



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ABSTRACT

In this study we investigated balancing responses to lateral perturbations during slow walking (0.85 m/s). A group of seven healthy individuals walked on an instrumented treadmill while being perturbed at the level of waist at left heel strike in outward and inward lateral directions. Centre of mass (COM) and centre of pressure (COP), rotation of pelvis around vertical axis, step lengths, step widths and step times were assessed. The results have shown that beside control of COP in lateral direction, facilitated by adequate step widths, control of COP in sagittal direction, slowing down movement of COM was present after commencement of lateral perturbations. Sagittal component of COM was significantly retarded as compared to unperturbed walking for both inward (4.32 ± 1.29 cm) and outward (9.75 ± 2.17 cm) perturbations. This was necessary since after an inward perturbation first step length (0.29 ± 0.04 m compared to 0.52 ± 0.02 m in unperturbed walking) and step time (0.45 ± 0.05 s compared to 0.61 ± 0.04 s in unperturbed walking) were shortened while after an outward perturbation first two step lengths (0.36 ± 0.05 m and 0.32 ± 0.11 m compared to 0.52 ± 0.03 m in unperturbed walking) were shortened that needed to be accommodated by the described modulation of COP in sagittal plane. In addition pronounced pelvis rotation assisted in bringing swing leg to new location. The results of this study show that counteracting lateral perturbations at slow walking requires adequate response in all three planes of motion.

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1. Introduction

An appropriate relationship between the centre of pressure (COP) and centre of mass (COM) must be maintained from step to step to provide stable human walking (Hof, 2007). Additionally, upsets in the form of slips, trips or bumping into other people challenge stability. Poor balance, particularly in the frontal plane, has been suggested as a major reason for falls in the elderly and neurologically impaired population (Maki and McIlroy, 2006). Studies investigating responses to lateral pushes during standing have shown that humans predominantly respond by unloading the leg ipsilateral to the side of perturbation, placing it subsequently distally in the direction of induced COM movement which requires cross stepping (Maki and McIlroy, 2006). This is a demanding manoeuvre even for healthy individuals. Several studies have investigated the balancing response of healthy individuals to lateral pushes at pelvis level at speeds close to normal walking speed (1.2 m/s) (Hof et al., 2010; Hof and Duysens, 2013; Qiao and

Jindrich, 2014; Martelli et al., 2015; Vlutters et al., 2016). These studies have revealed complex mechanisms to counteract perturbations including stepping, lateral ankle strategy and shortening of the stride time. Our previous study investigating balancing responses to lateral pushes at lower speed of walking (0.85 m/s) indicated coupling between all three planes of movement (Olenšek et al., 2016). Shorter step lengths were noticed with substantial deceleration of COM in the sagittal plane. Since our study was done during overground walking we were not able to assess COP.

The main mechanism of balance control during steady and perturbed walking is associated with the interaction of COP and COM where the COM accelerates away from the COP with the acceleration proportional to the COP-COM distance (Hof, 2007; Hof et al., 2010). Efficient control of balance during walking requires stepping that results in adequate placement of COP relative to COM position and velocity (Hof, 2008).

The aim of this study was to investigate in a group of healthy subjects the interplay between the COP and COM in sagittal and frontal directions following lateral pushes delivered at pelvis level at lower gait speed. Based on the findings of our previous study our

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hypothesis was that perturbations delivered in the frontal plane require control of COP not only in the lateral but also in the sagittal direction.

2. Methods

2.1. Subjects

Seven adult males (age 33.4 ± 8.5 years, body mass 80.1 ± 11.6 kg and height 180.6 ± 5.3 cm) free from musculoskeletal or neurological impairments participated in the study. This sample size was considered as adequate based on our previous study where similar methodology was used (Olenšek et al., 2016). The experimental protocol was approved by the local ethics committee and the subjects gave informed consent.

2.2. Perturbing apparatus and experimental procedure

Fig. 1 shows the experimental environment, which consisted of a balance assessment robot (BAR) and an instrumented treadmill. The BAR has six degrees of freedom (DOF) that interface to the pel-

vis of a walking subject. Three DOFs (translation of pelvis in the vertical direction, pelvic tilt and pelvic list) are passive while the remaining three DOFs (translation of pelvis in sagittal and lateral directions and pelvic rotation) are actuated and admittance-controlled. The BAR is capable of delivering perturbations in the directions indicated in Fig. 1. A detailed description of the BAR's architecture, control and performance is given in (Olenšek et al., 2016). COM movement was estimated from the translational movement of the BAR (Fig. 2). It has been shown that such an approximation method can be reliably applied during walking (Yang and Pai, 2014). Pelvic rotation (θ) was assessed by measuring the BAR's orientation in the transverse plane (Fig. 2). The interaction forces F_x , F_y and interaction torque T_z between a subject and the BAR were assessed by force cells (Fig. 2). COP recordings during walking were obtained by means of four force transducers placed underneath the treadmill according to the procedure described by Willems and Gosseye (2013). Spatio-temporal data were assessed by means of an Optitrack camera (NaturalPoint Inc.). The subjects' feet were equipped with reflective markers (medial malleoli, 1st metatarsal joint and 4th metatarsal joint) that constituted kinematic model (Olenšek et al., 2016).

Each subject first walked unperturbed on the instrumented treadmill for around ten minutes while being embraced with the BAR device set in a transparent mode. Another five minutes of walking was done where perturbations were delivered randomly at intervals of approximately six seconds in all six directions indicated in Fig. 1. A perturbation was initiated each time a foot switch imbedded into the shoe insole under the subject's left foot was activated. The treadmill speed was set to 0.85 m/s, this being considered a typical walking speed for individuals who have had strokes (Raja et al., 2012). The amplitude of force impulse was set to 15% of body weight while the duration was set to 150 ms. Previous studies that used similar values of perturbations have shown that repeatable responses can be obtained (Hof et al., 2010; Vlutters et al., 2016). After this initial fifteen minutes of warming-up another set of perturbations applied in all six directions (five repetitions for each perturbation direction) were repeated randomly. Perturbations to the left (outward perturbed walking condition), perturbations to the right (inward perturbed walking condition) and unperturbed walking (unperturbed walking condition) were studied.

2.3. Data processing and statistical analysis

The COM_x, COM_y, θ , F_x , F_y , T_z (Fig. 2) and COP trajectories were first segmented into strides where the gait cycle was defined as being between two consecutive left heel strikes, as detected by the left foot switch. Four full gait cycles, one prior to and three after the onset of perturbation, were analysed. The duration of one gait cycle was approximately 1.15 s. Spatio-temporal responses were investigated in terms of step lengths, steps widths and step times where left (right) step length was calculated as a AP distance between ankle markers at the moment of left (right) foot strike while left (right) step width was defined as the ML distance between the same markers. Step times were defined as the time elapsed between two consecutive left (right) and right (left) foot strikes. Step lengths, widths and times of six steps were considered (one step prior to and five steps after perturbation commencement).

All trajectories and spatio-temporal variables were averaged across five repetitions for each experimental condition (for the unperturbed experimental condition twenty subsequent gait cycles were used). The pelvis maximal rotation angle (MRA) was determined from θ , pelvis maximal lateral displacement (MLD) from COM_x, and pelvis maximal sagittal displacement (MSD) from COM_y for all experimental conditions for each subject, and then



Fig. 1. Photo of a subject walking on an instrumented treadmill while being embraced by the BAR perturbing device (up). Top view illustration of perturbation directions (down).

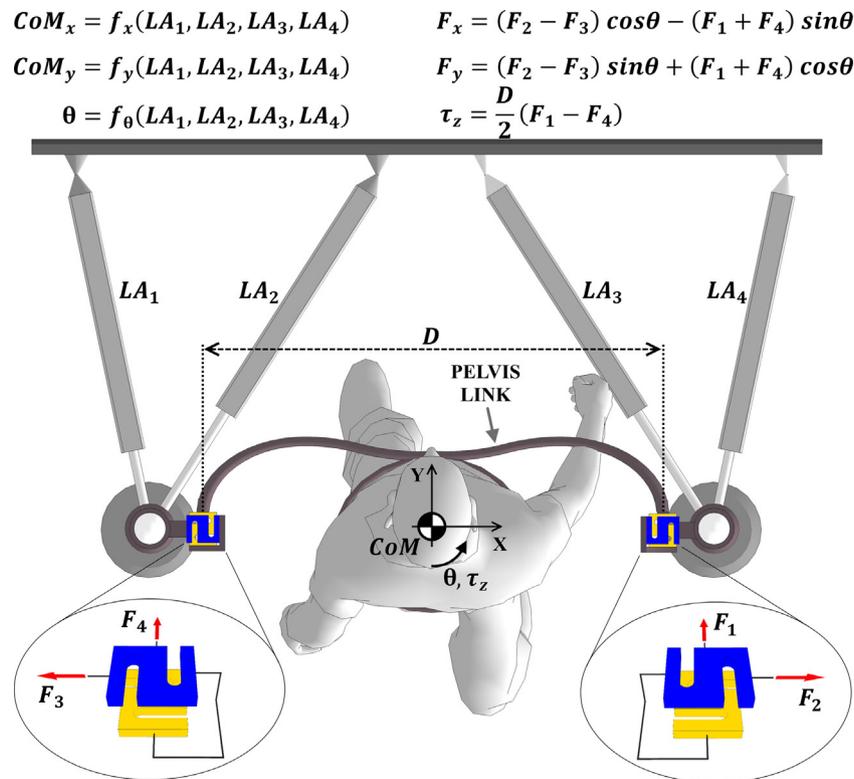


Fig. 2. Top view illustration of the experimental measurement set-up along with the indication of a laboratory coordinate system. Position and orientation of the BAR's pelvic link in the transverse plane corresponding to CoM_x , CoM_y and θ were estimated through kinematic model which is indicated by the above set of equations (left side). The lengths of linear actuators ($LA_1 - LA_4$) were measured by absolute encoders. Interaction forces F_x , F_y and torque T_z between subject's pelvis and BAR's pelvis link were calculated through the above set of equations (right side) from measurements obtained by two orthogonally oriented and serially connected pairs of load cells that connect pelvis link to the rest of BAR device as depicted in the close-up view. CoM_x , CoM_y , θ , F_x , F_y and T_z were all determined in the laboratory coordinate system. Power exchange during haptic interaction between a walking subject and the BAR was determined by multiplication of forces/torque (F_x , F_y and T_z) acting on the pelvis link and the corresponding linear/angular velocity ($d\text{CoM}_x/dt$, $d\text{CoM}_y/dt$ and $d\theta/dt$) of the pelvis link.

used to calculate the group average. The power of haptic interaction was calculated from the pelvis interaction forces (F_x , F_y and T_z) and pelvis kinematics ($d\text{CoM}_x/dt$, $d\text{CoM}_y/dt$ and $d\theta/dt$) for each movement direction (lateral translation – P_x , sagittal translation – P_y , rotation around the vertical axis – P_z). The obtained power signals were integrated over the period of perturbation action to determine the energy transferred to the walking subjects in each of the three directions (E_x , E_y and E_z). A paired t -test was used to compare MRA, MLD, MSP, E_x , E_y , E_z between each perturbed walking and unperturbed walking condition. For each of the four gait cycles considered (from -100% - 0% , 0% - 100% , 100% - 200% and 200% - 300%) the mean value of the differences between the COP and COM (COP-COM) signals were calculated for all experimental conditions for each subject. Based on the inverted pendulum model of walking (Hof, 2007; Hof et al., 2010) the COP-COM signal indicates relative displacement of COM at the end of a gait cycle in comparison to its position at the beginning of the gait cycle. The group averages were calculated for COP-COM separately for the ML (x axis) and AP directions (y axis) for each of the four gait cycles. Step lengths, step widths and step times were averaged across the group for each of the six steps considered for each experimental condition. To compare COP-COM over the four gait cycles and stepping responses over the six steps one-way repeated measures ANOVA was conducted separately for COP-COM in the lateral and sagittal planes and for step lengths, step widths and step times for each experimental condition. The Shapiro-Wilk test was used to examine normality

of distribution. Bonferroni adjusted post hoc pairwise comparisons were conducted when main effect was detected. The level of statistical significance was set to 0.05.

3. Results

Fig. 3 shows the pelvis kinematics, interaction forces and power exchange with the BAR device for a representative subject for outward perturbed and unperturbed walking. The MLD after commencement of perturbations reached around 20 cm while MSD reached around 10 cm in the backward direction. The displacement of COM was accompanied by counter clockwise pelvic rotation where the MRA reached around 5° .

Fig. 4 shows pelvis kinematics, interaction forces and power exchange with the BAR device for a representative subject for inward perturbed and unperturbed walking. The MLD after commencement of perturbations reached around 12 cm; the MSD reached around 5 cm in the backward direction while the MRA reached around 3° in the clockwise direction.

The interaction forces and power exchange in all the experimental conditions show similar performance, except for perturbation periods where a pronounced difference is observed in the lateral plane.

Fig. 5 shows the group mean MRA, MLD and MSD as well as E_x , E_y and E_z for outward and inward perturbed walking compared to unperturbed walking. Group MRA, MLD and MSD values were similar to those for representative subject in Figs. 3 and 4. Except for

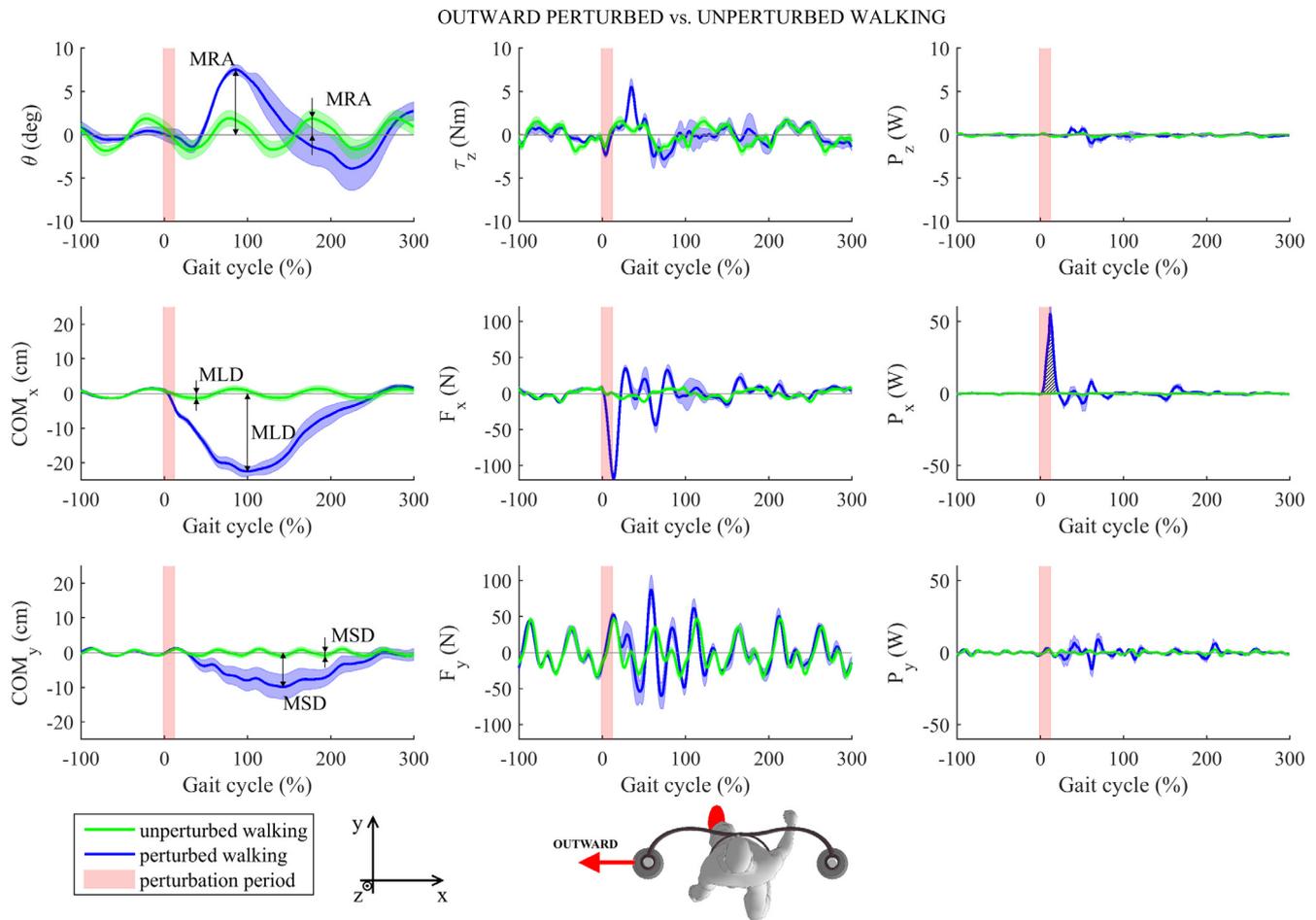


Fig. 3. Pelvis kinematics, interaction forces between BAR and subject's pelvis and power of haptic interaction between BAR and subject's pelvis are shown for a representative subject for outward perturbed walking condition along with the data of unperturbed walking condition. One stride prior and three strides following perturbation commencement are shown. Stride is defined as period between two consecutive left heel strikes as determined by footswitch. Displayed trajectories show mean values and standard deviations from five trials/five repetitions for outward perturbed walking condition and unperturbed walking condition. Within pelvis kinematics trajectories MRA, MLD and MSD are indicated for both experimental conditions. MRA, MLD and MSD were determined across all strides from zero to the maximal/minimal value of the respective curve for each experimental condition. Within power trajectories the shaded area under the power curve in the lateral direction P_x illustrates the period where the power was integrated to obtain energy of a perturbing push for both experimental conditions and for all three movement directions.

the comparison of MRA between the inward perturbed and unperturbed walking all other comparisons were statistically significant. Group means values for E_x , E_y and E_z showed that the only significant transfer of energy from the BAR to subjects occurred in the lateral plane (6.46 ± 1.81 J for outward perturbed walking condition and 3.55 ± 1.18 J for inward perturbed walking condition).

Fig. 6 shows COP, COM and COP – COM trajectories for the four considered gait cycles for all experimental conditions for a representative subject. During unperturbed walking one can observe alternating movement of all three signals where the average value over one gait cycle is around zero for both ML and AP directions. Outward perturbation initiated COM movement in the ML direction to the left which was accompanied by the movement of COP to the left being determined by placement of feet in subsequent steps. In the AP direction COP under the left foot that entered the stance phase has been moved toward the toes. This initiated posterior movement of COM in the first gait cycle following the perturbation. Later, in the second gait cycle the movement of COP was shifted posteriorly so that COM movement backward was contained and in the last gait cycle COM was moved back to the middle of the treadmill. Inward perturbation initiated COM movement in the ML direction to the

right, which was accompanied by the movement of COP to the right, and in the following two gait cycles COP was controlled in a way that brought COM back to the middle of the treadmill. In the AP direction a similar but less pronounced mechanism to that described for the outward perturbation, can be observed.

Fig. 7 shows group average COP-COM for ML and AP directions for all four gait cycles considered. The magnitudes of COP-COM in a gait cycle preceding the perturbation were close to zero for all experimental conditions in both ML and AP directions. The magnitudes of COP-COM values in the first gait cycle following perturbations were significantly larger (-3.15 ± 1.97 cm for outward perturbed and 4.5 ± 1.49 cm for inward perturbed in ML direction and 3.46 ± 1.76 cm for outward perturbed and 1.41 ± 0.59 cm for inward perturbed in AP direction) than for the gait cycle prior to perturbation while the sign was consistent with movement of COM. In the second and third gait cycles following perturbations COP-COM were consistent with the movement of COM. Statistical analysis showed that there were no statistically significant differences in COP-COM for unperturbed experimental condition across four gait cycles (Table 1). There were statistically significant differences in COP-COM across four gait cycles for outward and inward

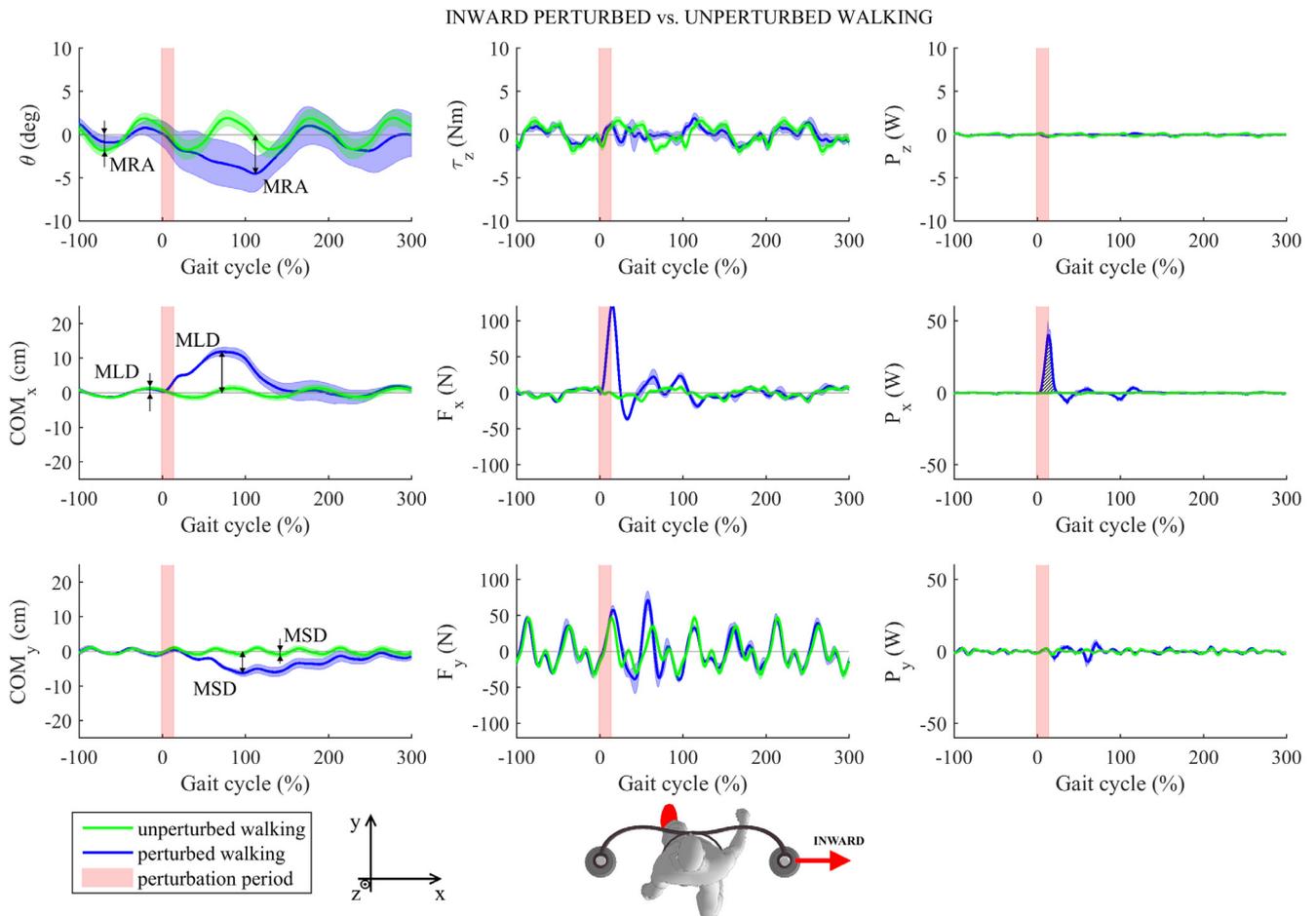


Fig. 4. Pelvis kinematics, interaction forces between BAR and subject's pelvis and power of haptic interaction between BAR and subject's pelvis are shown for a representative subject for inward perturbed walking condition along with the data of unperturbed walking condition. One stride prior and three strides following perturbation commencement are shown. Stride is defined as period between two consecutive left heel strikes as determined by footswitch. Displayed trajectories show mean values and standard deviations from five trials/five repetitions for inward perturbed walking condition and unperturbed walking condition. Within pelvis kinematics trajectories MRA, MLD and MSD are indicated for both experimental conditions. MRA, MLD and MSD were determined across all strides from zero to the maximal/minimal value of the respective curve for each experimental condition. Within power trajectories the shaded area under the power curve in the lateral direction P_x illustrates the period where the power was integrated to obtain energy of a perturbing push for both experimental conditions and for all three movement directions.

experimental conditions. The main effects and pairwise interactions are given in Table 1.

Fig. 8 illustrates stepping in all three experimental conditions for a representative subject. Outward perturbation caused the first step after perturbation commencement (RS1) to be shorter (around 35 cm) and narrower (-5 cm) than for unperturbed walking (step length around 50 cm and step width around 12 cm). The second step after perturbation (LS1) was also shorter. Subsequent steps were of similar lengths and widths. Inward perturbation caused the first step after perturbation commencement (RS1) to be shorter (around 30 cm) and wider (around 30 cm). Subsequent steps were of similar length while the second step after perturbation (LS1) was still wider than for unperturbed walking.

Fig. 9 shows step lengths, step widths and step times for the whole group. Statistical analysis showed that there were no statistically significant differences for unperturbed experimental condition (Table 1). Further, statistical analysis showed that there were statistically significant differences in step lengths and step widths for both outward and inward experimental conditions. There was no main effect for step times in the outward experimental condition while there were statistically significant differences in step

times for the inward experimental condition. The main effects and pairwise interactions are given in Table 1.

4. Discussion

Our study shows that an effective balancing response to lateral perturbation pushes applied at pelvis level to healthy subjects walking at low speed requires adequate control of COP in the ML direction as well as in the AP direction.

4.1. Balancing responses

There is a clear difference between balancing responses following inward and outward perturbations, particularly in terms of capacity of the next step to reverse the COM movement in the frontal plane. Following an outward perturbation the first foot placement (RS1) due to leg crossing cannot be displaced sufficiently away in the ML plane to enable substantial COP-COM difference, so consequently the movement of the COM in the ML direction is fully arrested only after the second step (LS1). Following an inward perturbation the first foot placement (RS1) is already well dis-

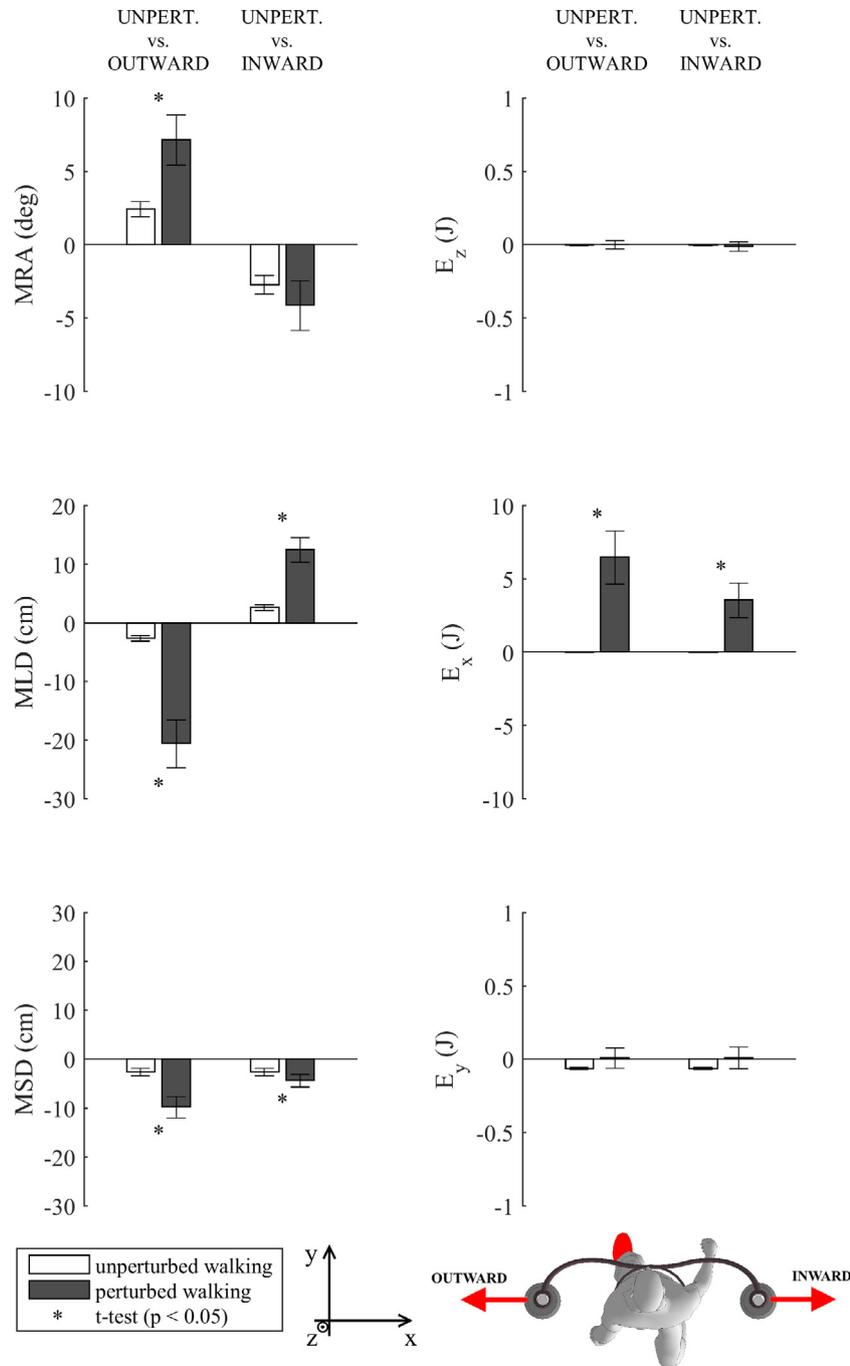


Fig. 5. Group mean values and standard deviations of MRA, MLD and MSD following outward and inward perturbations along with the unperturbed data are shown (left side). Group average of energy transferred during perturbation period following outward and inward perturbations along with the values for unperturbed walking condition are shown (right side). Positive values indicate energy transfer from the BAR to the walking subject. The asterisk (*) denotes statistically significant differences between perturbed and unperturbed experimental conditions.

placed laterally to enable the substantial COP-COM difference that is needed to arrest COM. This is one reason for observing smaller COM deviation in the lateral plane after inward perturbation than for the outward perturbation. Another reason is related to the observation that despite the magnitude of commanded push being the same in both directions, the resulting interaction force (F_x) and energy (E_x) of inward push was smaller than for the outward push. This was because the COM velocity in the lateral direction at the perturbation commencement was in both cases directed outward.

The first step (RS1) and the second step (LS1) after inward perturbation were significantly faster than for unperturbed walking. A similar trend can be observed also after outward perturbation; however this group effect did not reach significance due to substantial group variability indicating that some subjects shortened step time while others did not.

Both perturbed conditions have in common that the first step length (SL_{R1}) following perturbation was significantly shortened, requiring compensation in the form of COP displacement in the

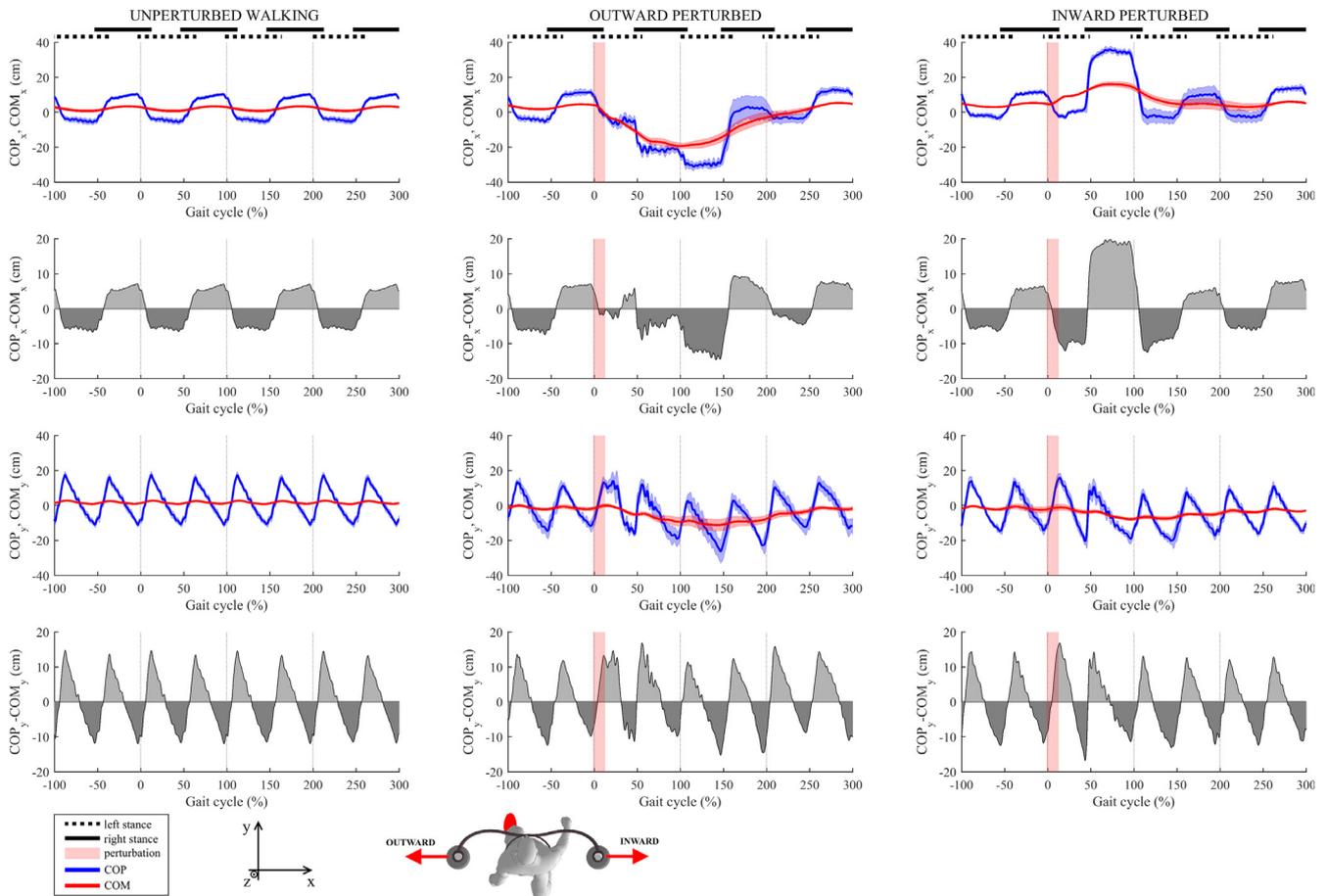


Fig. 6. COP, COM and COP-COM trajectories for ML direction (x axis) and AP direction (y axis) are shown for a representative subject. One stride prior and three strides following perturbation commencement are shown. Stride is defined as a period between two consecutive left heel strikes as determined by footswitch. Displayed trajectories show mean values and standard deviations from five trials/five repetitions for outward and inward perturbed walking conditions and unperturbed walking condition. Left – unperturbed walking condition, middle – outward perturbed walking condition and right – inward perturbed walking condition. On the top of each column left and right stance phases are indicated. Shaded areas under the COP-COM trajectories enable visual integration over each gait cycle thus indicating the value of $\overline{\text{COP-COM}}$.

sagittal direction during the stance phase of the left leg (LS0). This is in agreement with findings of O'Connor and Kuo (2009) who found that visual perturbations in ML plane affected step-length variability in AP plane. In anticipation of the shortened first step after lateral perturbation, the COP under the stance leg (LS0) is pushed actively toward the forefoot (Fig. 6) as can be concluded from comparison with the COP trajectory of unperturbed walking (foot locations were in both cases identical). The described control of COP increased the COP-COM signal over the gait cycle causing COM to decelerate in the sagittal direction. This mechanism was more pronounced after an outward perturbation where the first and the second steps (RS1, LS1) were significantly shortened. The described control of COP in the sagittal direction must be performed otherwise after the next step (RS1), which is significantly shortened, COP would be much closer to COM resulting in a COP-COM value inadequate to decelerate COM movement sufficiently; this would set conditions in the sagittal plane as if subject was pushed forward. Our results confirm the proposition of Hof (2008) that modulation of COP along the foot may play significant role in adjusting the COP – COM relationship needed for balancing in the sagittal plane.

Results also show that an additional mechanism that helps in bringing the swinging leg to the next stance location, particularly in the case of outward perturbation, is the rotation of the pelvis

around the vertical axis resulting in the so called “pelvic step” (Liang et al., 2014).

Considering the limited capacity of neurologically impaired subjects, especially stroke survivors, to modulate COP under parietic extremity (Chisholm et al., 2015) as well as to move parietic extremity to the next stance location, lateral perturbations may induce not only changes to COM movement in the frontal plane but also “self-induced” perturbations in the sagittal plane. The findings of this study provide an additional explanation of why perturbations in the lateral directions are so challenging and frequently lead to falls in neurologically impaired as well as elderly individuals (Stolze et al., 2004).

4.2. Methodological considerations

In this study we limited our analysis mainly to COP-COM relationship. Previous studies suggested that humans during walking respond to lateral perturbations by adjusting lateral placement of the foot according to predictions of an inverted pendulum model (Hof, 2008; O'Connor and Kuo, 2009; Hof et al., 2010). While healthy individuals responding to tripping employ other mechanisms that also contribute to the modulation of the horizontal ground reaction force by rotation of body segments - most notably trunk (Wang et al., 2012) - this seems not to be the case when per-

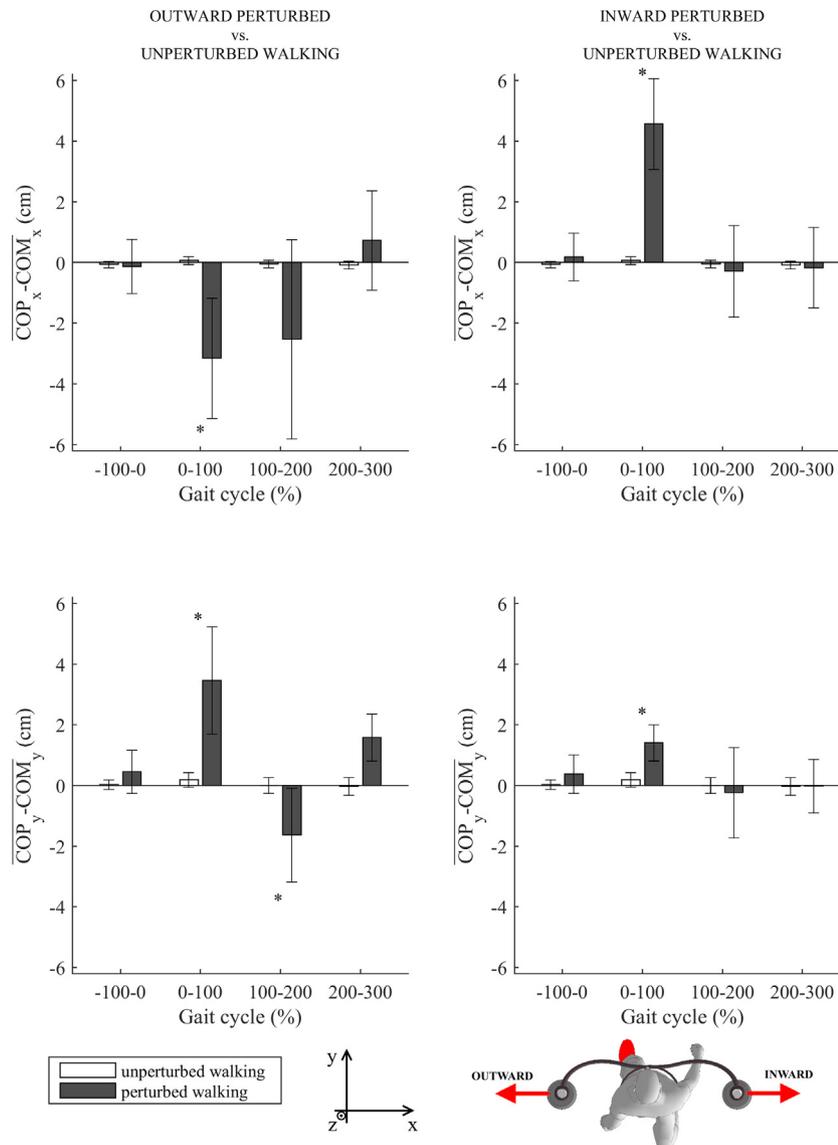


Fig. 7. Group mean values and standard deviations of $\overline{COP-COM}$ for each of the four consecutive gait cycles are shown. Left side – data for the outward perturbed walking condition along with data for the unperturbed walking condition, right side – data for the inward perturbed walking condition along with data for the unperturbed walking condition. Upper row – ML direction (x axis), lower row – AP direction (y axis). The asterisk (*) denotes values of post-perturbation gait cycles that are significantly different from the values of pre-perturbation gait cycle as determined by post hoc analysis. Complete post hoc analysis is given in [Table 1](#).

turbations are delivered proximally. Here, appropriate COP modulation is predominantly utilised, as COP-COM correlates very well with the acceleration of COM in the transverse plane (Hof, 2007; Vlutters et al., 2016). This was the rationale for looking into $\overline{COP-COM}$ over a gait cycle as this variable should, according to the inverted pendulum model, correspond to the displacement of COM at the end of the gait cycle. In unperturbed walking conditions $\overline{COP-COM}$ is close to zero indicating that the COM position should be the same at the beginning and the end of a gait cycle, which is confirmed by the results. In the perturbed experimental conditions COM was accelerated sideways after the push. Thus, $\overline{COP-COM}$ in ML direction should be increased in the first gait cycle following perturbation in the direction of perturbation action to contain perturbation-induced movement of COM. On the other hand, $\overline{COP-COM}$ in the AP direction in the first gait cycle should be positive to match the observed displacement of COM backward.

Comparison of MLD and MSD displacements after lateral perturbations (Fig. 5) and corresponding $\overline{COP-COM}$ values in ML and AP directions (Fig. 7) following perturbation showed that this was the case.

In Hof et al. (2010) trunk angles were monitored in perturbed walking where similar magnitude of perturbation pulses were used, as in our study showing negligible trunk inclinations. This is also consistent with the nature of the push, which is directed close to COM thus mainly affecting the linear momentum. Humans are during walking very much concerned with minimizing fluctuations of the overall angular momentum in all three planes (Herr and Popovic, 2008). Thus, when counteracting perturbations it seems a natural choice to employ strategies that change angular momentum as little as possible.

Analysis of the imposed perturbation energy has shown that perturbations acted purely in the lateral direction; therefore the

Table 1

F-test values and *p* values from one way repeated measures ANOVA and Bonferroni adjusted post hoc pairwise analysis on step length, step width, step time and COP-COM (AP and ML directions). Statistics was done separately for each experimental condition (unperturbed walking, outward perturbed walking and inward perturbed walking).

		Unperturbed walking		Outward perturbed walking	Inward perturbed walking
Step length	Within - subjects effect	F-test	0.162	18.332	45.439
		p-value	0.767	0.000*	0.000*
	Pairwise comparison - Step-Step (<i>p</i> -value)		/	1–2 (0.0064*), 1–3 (0.030*), 2–4 (0.009*), 2–5 (0.022*), 2–6 (0.006*), 3–4 (0.039*), 3–5 (0.014*)	1–2 (0.002*), 2–3 (0.001*), 2–4 (0.002*), 2–5 (0.002*), 2–6 (0.001*)
Step width	Within - subjects effect	F-test	1.577	55.059	39.734
		p-value	0.197	0.000*	0.000*
	Pairwise comparison - Step-Step (<i>p</i> -value)		/	1–2 (0.000*), 1–4 (0.001*), 2–3 (0.007*), 2–4 (0.000*), 2–5 (0.003*), 2–6 (0.000*), 3–4 (0.011*), 3–6 (0.008*), 4–5 (0.001*)	1–2 (0.000*), 1–3 (0.048*), 2–4 (0.001*), 2–5 (0.000*), 2–6 (0.000*), 3–4 (0.000*), 3–6 (0.025*)
Step time	Within - subjects effect	F-test	0.486	1.751	35.557
		p-value	0.784	0.154	0.000*
	Pairwise comparison - Step-Step (<i>p</i> -value)		/	/	1–2 (0.002*), 1–3 (0.009*), 2–3 (0.048*), 2–4 (0.002*), 2–5 (0.005*), 2–6 (0.002*), 3–4 (0.047*), 3–5 (0.039*), 3–6 (0.005*)
<u>Cop_y-Com_x</u>	Within - subjects effect	F-test	1.507	6.304	27.148
		p-value	0.257	0.036*	0.001*
	Pairwise comparison – Gait cycle- Gait cycle (<i>p</i> -value)		/	1–2 (0.016*), 2–4 (0.032*)	1–2 (0.000*), 2–3 (0.017*), 2–4 (0.000*)
<u>Cop_y-Com_y</u>	Within - subjects effect	F-test	0.855	16.702	8.587
		p-value	0.482	0.003*	0.001*
	Pairwise comparison - Gait cycle-Gait cycle (<i>p</i> -value)		/	1–2 (0.035*), 1–3 (0.037*), 2–3 (0.031*), 3–4 (0.015*)	1–2 (0.010*), 2–3 (0.074*), 2–4 (0.017*)

* *p* < 0.05.

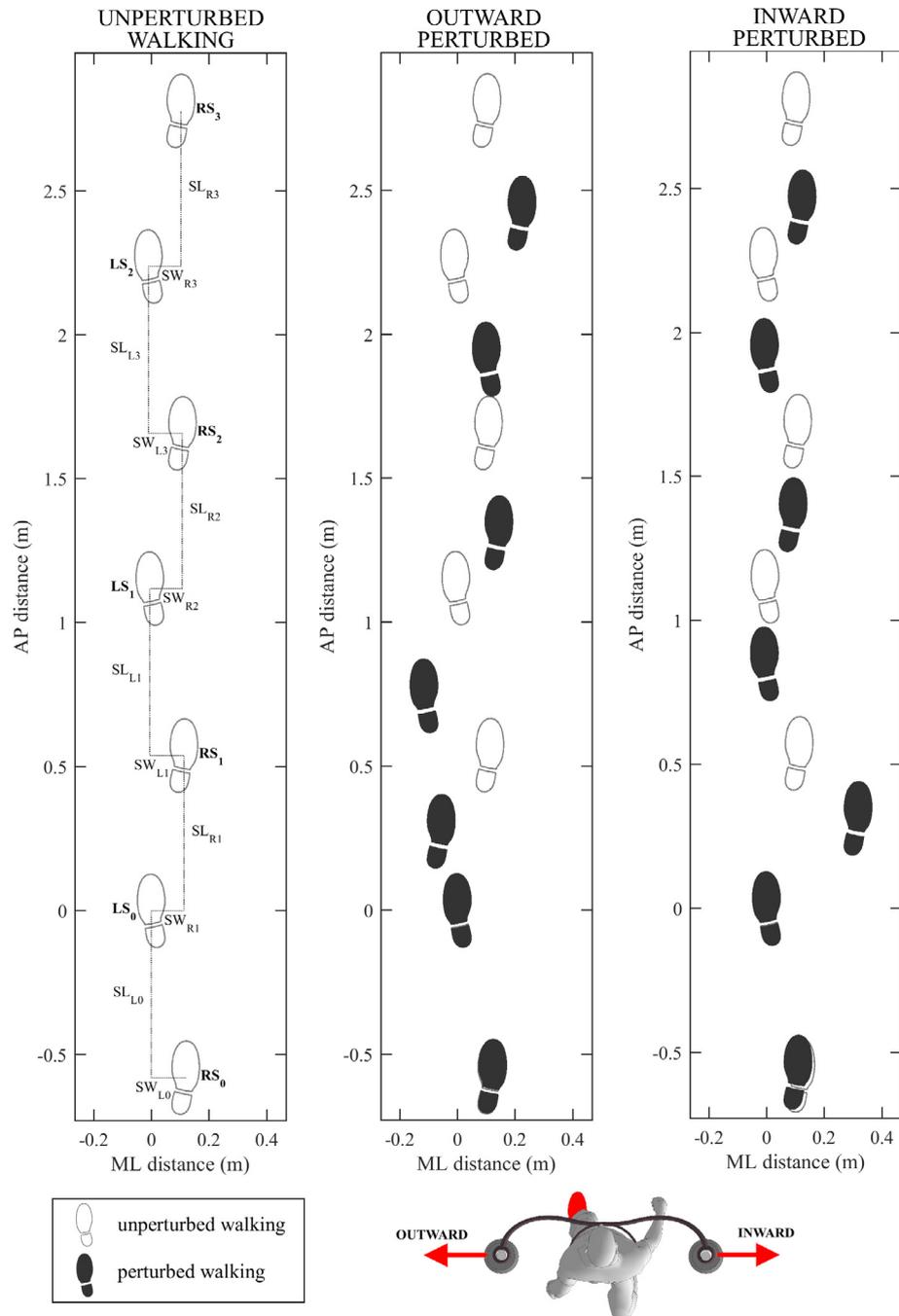


Fig. 8. Footprints illustrating stepping for all three experimental conditions in a representative subject. Average of five trials are shown for each experimental condition. Left side – unperturbed walking (each consecutive left and right stance phases are denoted along with the indication and labeling of consecutive step lengths and step widths), middle – outward perturbed walking together with unperturbed footprints, right – inward perturbed walking together with unperturbed footprints. One step prior (from RS₀ to LS₀) and five steps after perturbation commencement (from LS₀ through RS₃) are shown.

observed balancing responses in the sagittal plane are fully attributable to human activity and not to the interaction with the apparatus. The energy exchange between the BAR and the walking subjects outside the perturbation interval was rather small.

4.3. Limitations of the study

This study was limited in terms of the explored perturbation space. Perturbation timing was limited to the heel strike. The responses would be different if perturbations were delivered at

other time instants (Hof et al., 2005). Also, a single speed of walking was investigated. It is clear that balancing responses differ considerably with speed of walking. Finally, the magnitude of perturbation was selected to produce substantial stepping responses but was still limited to values that did not evoke substantial rotational movement of body segments. Varying the magnitude of perturbations would produce different results. However, even with these limitations the findings of this study provide important new insight into complexity of balancing during walking following lateral perturbations.

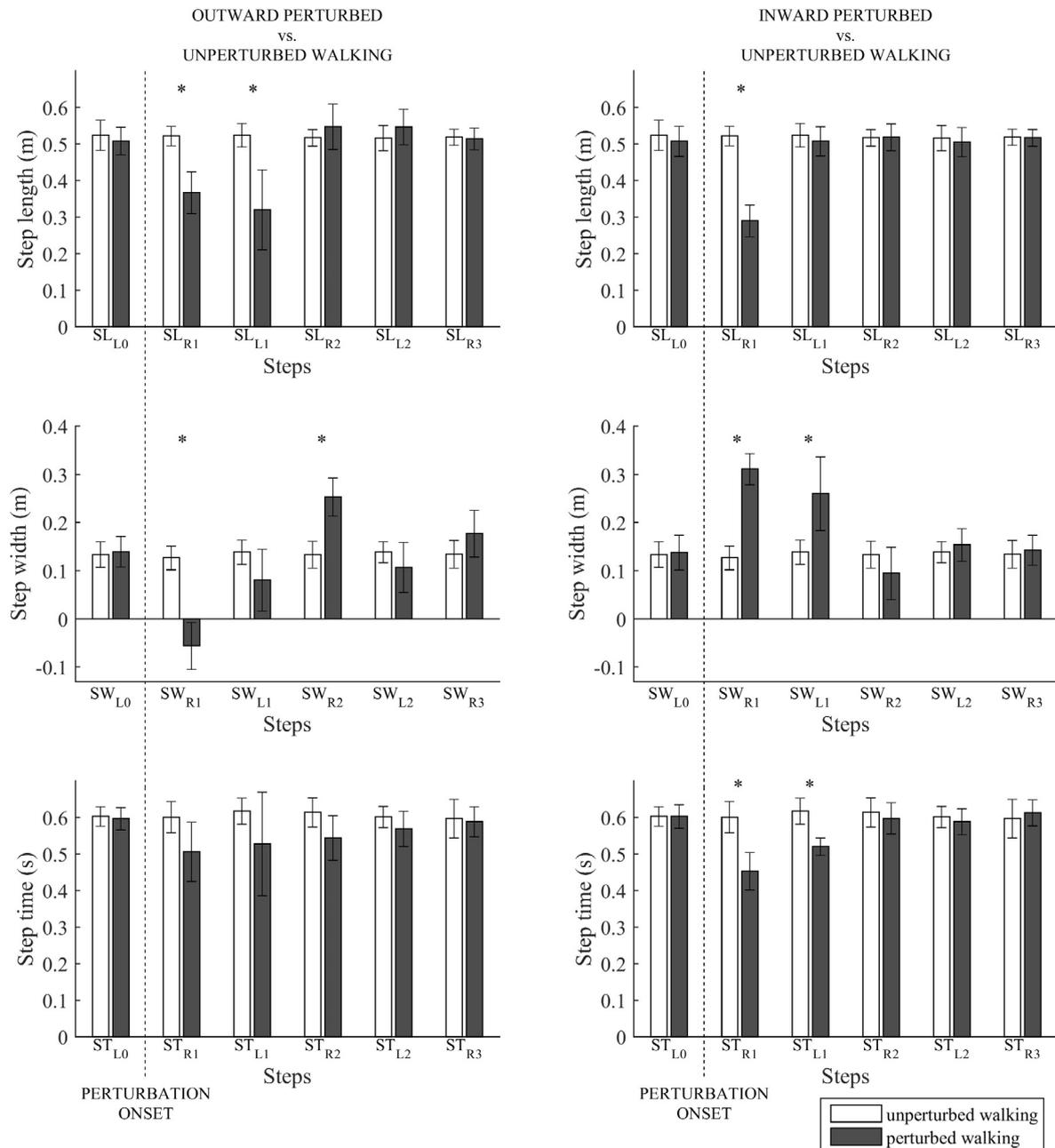


Fig. 9. Group mean values and standard deviations of step lengths, step widths and step times comparing perturbed (left side – outward perturbed walking condition, right – inward perturbed walking condition) and unperturbed walking condition for one step prior and five consecutive steps following perturbation commencement are shown. The asterisk (*) denotes values of post-perturbation gait cycles that are significantly different from the values of pre-perturbation gait cycle as determined by post hoc analysis. Complete post hoc analysis is given in Table 1.

Conflict of interest statement

The authors declare no conflict of interest.

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