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Small, movement dependent perturbations substantially alter postural control strategy in healthy young adults

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ABSTRACT

Postural control is commonly investigated by observing responses to perturbations. We developed a perturbation paradigm mimicking self-generated errors in weight shifting, which are a common cause of falling among older adults. Our aim was to determine the effects of this small, but complex, perturbation on postural sway of healthy young adults and evaluate the role of vision and cognition during movement dependent perturbations. Fifteen participants stood hip-width apart with their eyes open, closed and while performing two different cognitive tasks. Participants were continuously perturbed by medial-lateral (ML) support surface translations corresponding to, and hence doubling, their own center of mass sway. We analyzed the standard deviation (SD), root mean square (RMS), range, and mean power frequency (MPF) of center of pressure displacements. ML postural sway increased due to the perturbation (SD $p \leq .001$, range $p < .001$, RMS $p \leq .001$, MPF $p < .001$). Cognitive load increased the ML sway range ($p = .048$). Lack of vision increased ML MPF ($p = .001$) and anterior-posterior (AP) range ($p < .001$), SD ($p < .001$), and RMS ($p = .001$). Significant interaction of vision with the perturbation was found for the ML range ($p = .045$) and AP SD ($p = .018$). The perturbation specifically affected ML postural sway. Increased MPF is indicative of a postural control strategy change, which was insufficient for fully controlling the increased sway. Despite being small, this type of perturbation appears to be challenging for young adults.

1. Introduction

How humans maintain an upright posture, i.e., prevent falling, is one of the key questions of human movement research. Falls are an important health and societal problem, occurring in one third of older adults each year (Milat et al., 2011), often with severe and even fatal consequences (Rubenstein, 2006). The main approach to studying postural control in the laboratory is using perturbations that mimic circumstances of falling, such as obstacles that appear during gait (Potocanac et al., 2014a), unexpected trips (Potocanac et al., 2014b), slips (Pai et al., 2010), or displacements of the standing surface that mimic balance loss (Jilk et al., 2014). However, none of these perturbations are representative of incorrect shifts of bodyweight that occur in the absence of any external perturbation, the most common cause of falling in older adults living in residential care (Robinovitch et al., 2013).

Shifts of bodyweight are self-generated and develop dynamically over the course of a movement, hence mimicking them in the laboratory requires a novel approach. To do so, we developed a novel support surface perturbation paradigm, which is self-generated and movement dependent (Potocanac et al., 2017). This perturbation doubles one's self-generated bodyweight shifts in real time, resulting in continuous perturbations of varying amplitude. Such perturbations should interfere with both voluntary movements and postural control. Voluntary whole-body movements call for large bodyweight shifts and would result in large perturbations under this paradigm. In contrast, postural control during quiet stance would lead to small perturbations. The primary goal of this study is to evaluate the effects of such small, movement dependent perturbations on postural control of healthy young adults.

Vision and cognition are known to be involved in postural control during quiet stance. Vision is one of the vital sensory systems for postural control, which can compensate for the absence of other sensory information (Mergner, 2010). Deprivation of visual

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information increases body sway in healthy young adults (Benjuya et al., 2004) and these effects are exacerbated by more demanding postural tasks, e.g., narrower stance (Benjuya et al., 2004). Additionally, postural control requires cognitive resources (Dault et al., 2001), but the underlying mechanism remains unclear, as both increased/deteriorated (Pellecchia, 2003; Woollacott and Velde, 2008) and decreased/improved (Riley et al., 2010; Swan et al., 2007) postural sway were reported when quiet stance was performed concurrently with a cognitive task. A U-shaped relation between posture and cognition was proposed, whereas undemanding cognitive tasks result in improved postural control, while demanding tasks lead to its deterioration (Huxhold et al., 2006). When both tasks are sufficiently challenging, postural control deteriorates due to limited cognitive resources. On the other hand, an easy cognitive task might divert attention from the highly automated postural control task, thereby improving its efficacy (Huxhold et al., 2006). This results in increased frequency of postural sway, accompanied by decreased postural sway range and variability (Carpenter et al., 2001; Richer et al., 2017). Therefore, postural control can be challenged by various cognitive tasks and by withdrawal of visual information, which typically results in a change in the postural control strategy. We will evaluate how visual information and cognitive resources contribute to postural control which is already challenged by our novel perturbation paradigm inducing small, movement dependent translations of the support surface.

The aims of this paper are multifold. Firstly, we will determine the effects of movement dependent perturbations of support surface on postural sway of healthy young adults. We hypothesize that movement dependent perturbations would lead to increased sway frequency and decreased sway variability and range, which is a common finding in studies utilizing simpler support surface perturbations (e.g., single translations). However, due to the perturbation complexity, we expect these effects to be stronger compared to previous literature. Secondly, we will evaluate the role of visual information in postural control during movement dependent perturbations by comparing postural sway with eyes open and closed. We hypothesize that lack of visual information would exacerbate the perturbation effects. Finally, we will investigate the contribution of cognition to postural control during movement dependent perturbations by comparing postural sway when participants are or are not performing a concurrent cognitive task. We hypothesize that adding a cognitive task would exacerbate the perturbation effects.

2. Material and methods

2.1. Participants

Fifteen healthy young adults (mean ± standard deviation: age 24.1 ± 3.3 years, height 174.1 ± 7.6 cm, mass 70.2 ± 9.6 kg, 8 females) participated in this study after signing an informed consent from. The study was approved by National Medical Ethics Committee of the Republic of Slovenia (No. 112/06/13).

2.2. Experimental protocol

Participants performed four tasks: they stood in hip-wide stance with (1) eyes open (EO), (2) eyes closed (EC), (3) eyes open while performing a concurrent cognitive task requiring working memory (MEMORY), and (4) eyes open while performing a concurrent cognitive task requiring response inhibition (INHIBITION). In the memory task, participants had to remember common nouns presented for 4 s on a display positioned 1.4 m in front of them. Participants were instructed to remember as many nouns as possible and recollection was tested immediately after the task ended. In the inhibition task, participants performed an auditory Stroop task (Potocanac et al., 2015) by listening to the words ‘high’ and ‘low’ delivered in a low- or a high-pitched voice through headphones. They were required to respond by verbalizing the pitch used and their performance was tracked by the experimenter. Each task was performed with and without a postural perturbation, resulting in a total of eight experimental conditions. Experimental conditions were performed in random order and each lasted 240 s with a 2 to 3 min break in between.

We perturbed the participants using continuous, self-generated perturbations, which amplified their own postural sway by moving the support surface in the medial–lateral (ML) direction and in the transverse plane (Potocanac et al., 2017). More specifically, we translated the support surface by an amount corresponding to center of mass (COM) displacement relative to the fixed support surface, but in the opposite direction. Support surface perturbations were delivered using a system consisting of a robotic Stewart platform paired with real-time computing (Potocanac et al., 2017). The average delay of the perturbation was 154 ms (Potocanac et al., 2017), which is short enough for participants to perceive the perturbation as self-generated (Blakemore and Frith, 2003). Participants stood still on top of the robotic platform, with their arms relaxed beside the body and their feet hip-width apart. Position of the feet was marked on the floor and kept constant for the whole experiment.

Kinematics of the COM and the support surface were recorded at 100 Hz using a 2×3 Optotrak camera array (Northern Digital Inc., Waterloo, Ont., Canada), from markers positioned on the participant at the level of the fifth lumbar vertebra as an approximation of COM and on the robotic platform, respectively. The COM data was used as input for delivering the movement dependent, real-time perturbation. Center of pressure (COP) displacement was recorded using two Kistler force platforms (Kistler Instrumente AG, Winterthur, Switzerland) and analyzed as an outcome variable. We truncated the first 45 s of the data to avoid learning effects and filtered the data using a second-order, zero-lag, low pass Butterworth filter with a cut-off frequency of 5 Hz. This cut-off frequency was selected to facilitate comparison to previous literature (Benjuya et al., 2004; Carpenter et al., 2010; Dault et al., 2001; Zaback et al., 2016) and we visually verified that the filtered waveforms accurately represented the raw data. Finally, we calculated the standard deviation (SD), range, root mean square (RMS), and mean power frequency (MPF) of the COP movement in the subsequent 60 s for the anterior-posterior (AP) and ML directions separately. We analyzed 60 s instead of the remaining 195 s of the data to avoid fatigue effects.

2.3. Statistical analysis

Two separate two-way repeated measures analyses of variance (rANOVAs) with LSD posthoc tests were performed for each outcome measure (SD, range, RMS, MPF). The first rANOVA analyzed the role of vision, with factors vision (EO, EC) x perturbation (on, off). The second rANOVA analyzed cognitive load, with factors cognitive task (EO, MEMORY, INHIBITION) x perturbation (on, off). Prior to statistical analysis, all data were Box-Cox transformed to conform to normality and sphericity assumptions for the rANOVA. In the Results section (Table 1) data are presented in the original form for interpretation. All analyses were performed using MATLAB 2015 (Mathworks, Natick, USA) and SPSS. Statistics 21 (IBM, Chicago, IL, USA), with statistical significance set at $\alpha = 0.05$.

3. Results

Full data and p-values are shown in Tables 1 and 2, respectively.

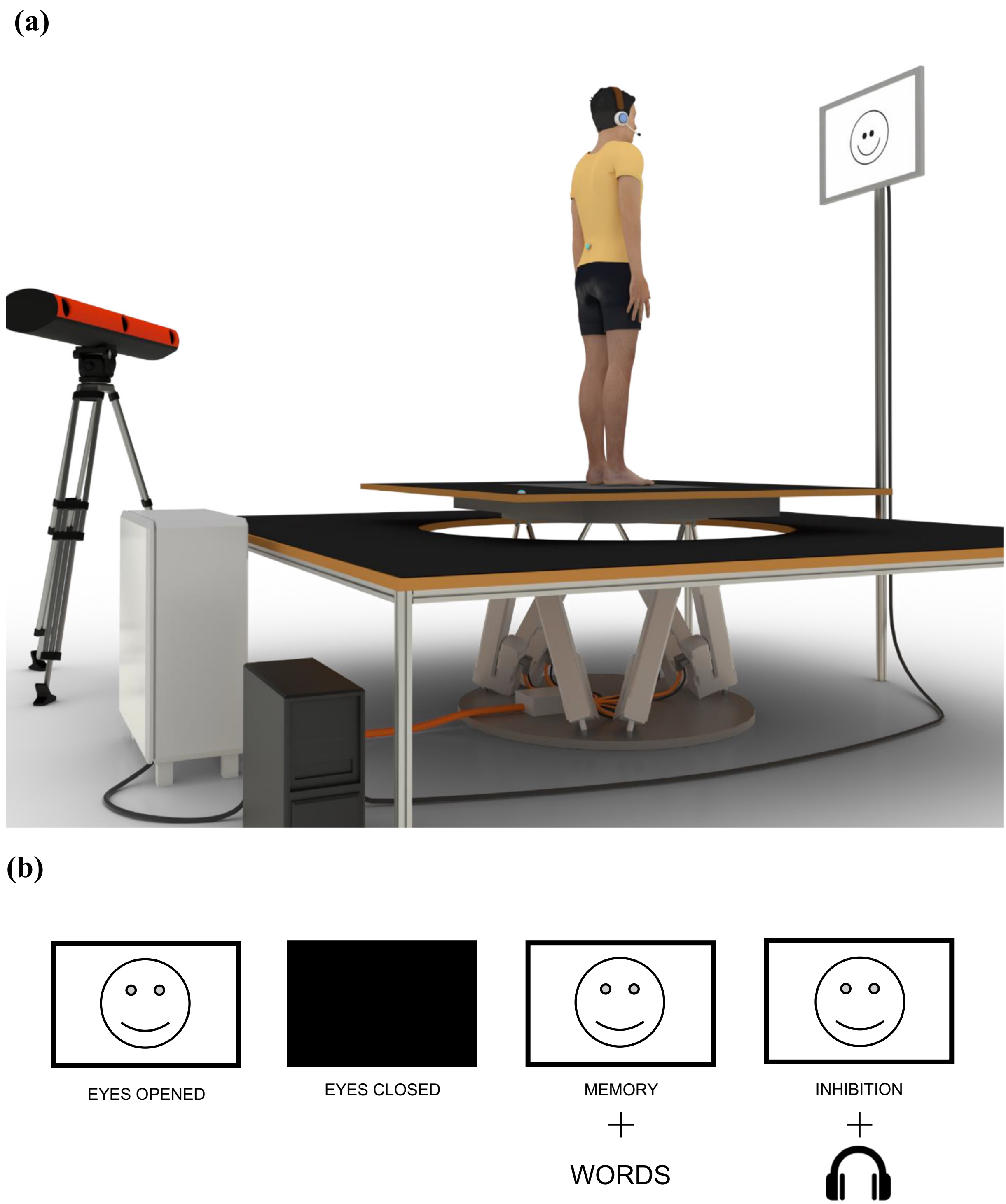

Fig. 1. Experimental set up and conditions. (a) Participants stood still on the platform, with their feet hip-width apart, in four conditions (b): with the eyes open, closed, and with the eyes open concurrently performing one working memory cognitive task and one response inhibition (audio Stroop) cognitive task. Each of these conditions was performed with and without the perturbation.

3.1. Standard deviation

SD of the COP was significantly increased by the perturbation (single task (ST): $F(1, 14) = 20.32$, $p = .001$, $\eta_p^2 = 0.529$ and dual task (DT): $F(1, 14) = 21.20$, $p < .001$, $\eta_p^2 = 0.602$), but only in the ML direction. In the AP direction, we found a significantly smaller SD when vision was available ($F(1, 14) = 34.34$, $p < .001$, $\eta_p^2 = 0.710$), compared to vision elimination. In the ML direction, we found no statistically significant effects of vision nor cognitive task. We found an interaction effect of vision with the perturbation in the AP direction ($F(1, 14) = 7.15$, $p = .018$ $\eta_p^2 = 0.338$), indicating the effect of vision elimination was exacerbated by the perturbation. In contrast, this interaction effect was not found in the ML direction. We found no interaction of cognitive tasks with the perturbation.

Table 1
Measures of center of pressure sway (mean ± standard deviation) in four conditions; eyes opened, eyes closed, memory cognitive task and response inhibition cognitive task, for medial-lateral and anterior-posterior direction separately, with and without the perturbation.

			EO	EC	Memory	Inhibition
Platform off	ML	SD (mm)	1.64 ± 0.69	1.67 ± 1.12	2.22 ± 1.28	2.67 ± 1.98
		Range (mm)	10.00 ± 5.35	12.45 ± 15.04	18.94 ± 26.80	16.71 ± 13.06
		RMS (mm)	1.90 ± 0.65	2.10 ± 1.40	2.38 ± 1.22	2.81 ± 1.95
		MPF (Hz)	0.21 ± 0.12	0.29 ± 0.15	0.31 ± 0.35	0.25 ± 0.13
	AP	SD (mm)	3.58 ± 1.13	4.10 ± 1.39	3.66 ± 1.32	3.61 ± 1.35
		Range (mm)	21.66 ± 15.01	24.80 ± 14.88	22.12 ± 11.35	23.45 ± 12.29
		RMS (mm)	3.80 ± 1.14	4.73 ± 1.65	3.88 ± 1.27	3.91 ± 1.58
		MPF (Hz)	0.18 ± 0.08	0.20 ± 0.09	0.21 ± 0.11	0.22 ± 0.07
Platform on	ML	SD (mm)	2.41 ± 0.90	3.77 ± 2.10	4.02 ± 3.00	3.96 ± 3.71
		Range (mm)	14.85 ± 6.93	30.79 ± 29.77	23.93 ± 17.25	27.42 ± 26.44
		RMS (mm)	2.70 ± 1.09	3.94 ± 2.01	4.23 ± 2.96	4.21 ± 3.67
		MPF (Hz)	0.52 ± 0.22	0.75 ± 0.25	0.46 ± 0.22	0.52 ± 0.15
	AP	SD (mm)	3.46 ± 1.05	4.56 ± 1.77	3.76 ± 1.91	3.36 ± 0.77
		Range (mm)	19.05 ± 7.16	27.43 ± 14.02	20.95 ± 7.99	19.37 ± 6.73
		RMS (mm)	3.83 ± 1.10	5.04 ± 2.23	3.99 ± 2.15	3.61 ± 0.84
		MPF (Hz)	0.16 ± 0.08	0.20 ± 0.09	0.18 ± 0.06	0.19 ± 0.09

EO = eyes opened, EC = eyes closed, ML = medial–lateral, AP = anterior-posterior, SD = standard deviation, RMS = root mean square, MPF = mean power frequency.

Table 2
P-values for main effect of perturbation, vision, cognitive tasks and interaction of perturbation with vision or cognitive task. Significant p-values (p <.05) are shown in bold.

		rANOVA vision × perturbation			**rANOVA cognitive task × perturbation**		
		Vision	Perturbation	Interaction	Cognitive task	Perturbation	Interaction
ML	SD (mm)	.216	**<.001**	.060	.055	**<.001**	.598
	Range (mm)	.115	**<.001**	**.045**	**.048**	**<.001**	.865
	RMS (mm)	.636	**.005**	.066	.115	**.001**	.641
	MPF (Hz)	**.001**	**<.001**	.148	.845	**<.001**	.220
AP	SD (mm)	**<.001**	.055	**.018**	.901	.901	.920
	Range (mm)	**<.001**	.823	.579	.759	.692	.775
	RMS (mm)	**.001**	.919	.990	.952	.944	.976
	MPF (Hz)	.115	.874	.375	.087	.079	.930

ML = medial-lateral, AP = anterior-posterior, SD = standard deviation, RMS = root mean square, MPF = mean power frequency.

3.2. Root mean square

RMS of the COP was significantly increased by the perturbation (ST: F (1, 14) = 11.18, p = .005, η_p^2 = 0.444 and DT: F (1, 14) = 18.46, p = .001, η_p^2 = 0.569), but only in the ML direction. In the AP direction, we found significantly smaller RMS (F (1, 14) = 17.52, p = .001, η_p^2 = 0.556) when vision was available. We found no significant effect of the cognitive task nor interactions of cognitive task or vision with the perturbation.

3.3. Range

Range of COP movement was significantly increased by the perturbation, but only in the ML direction (ST: F (1, 14) = 20.53, p < .001, η_p^2 = 0.595 and DT: F (1, 14) = 25.69, p < .001, η_p^2 = 0.647). In the AP direction, we found a significantly smaller range of COP motion when vision was available (F (1, 14) = 30.22, p < .001, η_p^2 = 0.683). In the ML direction, we found a significant effect of the cognitive task (F (2, 28) = 3.38, p = .048, η_p^2 = 0.195), but the post-hoc comparisons were not significant. Additionally, we found an interaction of vision with the perturbation (F (1, 14) = 4.87, p = .045, η_p^2 = 0.258) in ML direction indicating an exacerbated effect of the perturbation during the EC condition. We found no other significant interactions.

3.4. Mean power frequency

MPF was significantly increased by the perturbation, but only in the ML direction (ST: F (1, 14) = 81.48, p < .001, η_p^2 = 0.853 and DT: F (1, 14) = 42.63, p < .001, η_p^2 = 0.753. In the ML direction, we found a significantly higher MPF when vision was not available (F (1, 14) = 16.63, p = .001, η_p^2 = 0.543). We found no interactions of cognitive task or vision with the perturbation.

4. Discussion

The primary aim of this study was to investigate the effects of a novel continuous movement dependent support surface perturbation on postural sway of healthy young adults in single and dual task situations. Since our perturbation paradigm doubled self-generated bodyweight shifts and our participants were asked to stand still, the support surface perturbations delivered were very small (7.32 ± 3.95 mm). Yet, such small ML perturbations led to increased frequency, variability and range of ML COP sway in healthy young adults, under both single and dual task conditions. However, this is in contrast with several previous studies, in which challenging conditions typically lead to increased sway frequency, but decreased variability and range (Carpenter et al., 2001; Richer et al., 2017). We expected similar results, given the predictability of our perturbation. Alternatively, one could argue that increased frequency, range and variability of COP displacement are not a surprise, since the perturbation amplified the participants' COM sway. Indeed, our results are similar to those of a previous study comparing participants standing on seesaw-shaped support surfaces (Dault et al., 2001). In this study, the shape of the support surface induced movement dependent perturbations, and the authors found increased sway frequency and amplitude (Dault et al., 2001). Hence, this type of perturbation appears to be especially challenging; increased sway frequency indicates that participants altered their postural control strategy (Carpenter et al., 2001;

Stins et al., 2011), but this was insufficient to cope with the perturbation, resulting in increased range and variability of postural sway. The reasons for this insufficient response should be investigated further, since this study cannot answer whether young adults were unable or unwilling to completely account for the perturbation. It might be that the perturbation was not threatening enough to elicit a strong response. We believe this is not the case, based on previous literature showing lower values of MPF during challenging tasks (Carpenter et al., 2001; Polskaia & Lajoie, 2016). Alternatively, it is possible that increased sway served an exploratory role. Namely, these findings appear to be in line with a recent proposal for exploratory role of postural sway (Carpenter et al., 2010). This theory postulates that postural sway might be a part of a perception–action strategy enabling the participant to elucidate his interaction with the environment (Carpenter et al., 2010). In our experiment, the environment moves in a complex, movement dependent manner, possibly requiring increased sway to explore the perturbation, i.e., the interaction between the participant and the environment.

Our second aim was to evaluate the role of visual information in postural control during movement dependent perturbations. We expected the lack of vision to exacerbate the effects of the perturbation, which was only partially the case. We found a significant interaction between vision and perturbation for the COP sway range in the ML direction and SD in AP direction, indicating that participants swayed more when perturbed with eyes closed compared to eyes open. Two possible explanations for the lack of a more profound effect exist. It is possible that a ceiling effect occurred, since the perturbation already significantly affected postural sway. We measured MPF values of COP sway which were higher than previously reported in a range of tasks. For example, we found MPF of 0.52 ± 0.22 Hz when the perturbation and vision were present, while previous work reported maximal MPF values of ~0.45 Hz when participants performed a challenging cognitive task in a feet-together stance (Richer et al., 2017). Typically, MPF values of 0.23–0.33 Hz are reported during quiet stance performed with and without various cognitive tasks, at various levels of threat, and with varying availability of sensory information (Carpenter et al., 2001; Polskaia and Lajoie, 2016; Zaback et al., 2016). Alternatively, it is possible that participants relied less on visual information due to the type of the perturbation. Namely, the perturbation doubled participants' own postural sway. This predictive information could have been incorporated in an internal model of movement (Wolpert et al., 1998), thus reducing reliance on vision. Furthermore, the perturbation was continuous, possibly leading to anticipatory strategies for improving balance, such as muscle co-contraction (Carpenter et al., 2001). It was previously found that relying on co-contraction around the ankle under challenging postural conditions reduces detrimental effects of vision loss in older adults (Benjuya et al., 2004). Unfortunately, we did not measure muscle activity and cannot confirm this hypothesis.

Overall, we found significant main effects of vision on sway variability and range in the AP direction and on sway frequency in the ML direction. This confirms the importance of visual information for postural control, reported by many previous studies (Albertsen et al., 2017; Sarabon et al., 2013). When vision was removed, we found higher sway frequency in the ML direction, which indicates a change in balance control strategy (Carpenter et al., 2001; Stins et al., 2011). In contrast, in the AP direction, we found increased sway variability and range, but not frequency. Such discrepancies of results in the AP and ML direction are common in previous studies (Carpenter et al., 2001) and support the idea of independent control of AP and ML balance in side-by-side stance position (Winter et al., 1996).

Thirdly, we investigated the contribution of cognition to postural control during movement dependent perturbations by comparing postural sway between a single-task and two dual task conditions. When participants performed the cognitive tasks, the ML sway range increased, in line with literature showing increased body sway when participants perform easy postural tasks concurrently with cognitive tasks (Pellecchia, 2003; Woollacott and Velde, 2008). However, in contrast to our hypothesis and previous literature, we found no significant interaction between the perturbation and cognitive load. Previously, interactions between cognitive load and postural load imposed by different stance types (Albertsen et al., 2017) and between reaction time and perturbation of quiet stance (Redfern et al., 2002) were reported. The lack of interaction in our study might indicate that this type of movement dependent perturbation does not require cognitive resources, as it can be automatically controlled. Such an explanation is conceivable, since the perturbation amplified participant's own postural sway, which is considered to be automatically controlled in healthy young adults (McNevin et al., 2003). Alternatively, previous studies pairing support surface perturbations with cognitive tasks indicate that cognitive resources might be needed in a specific phase of the recovery response (Maki and McIlroy, 2007). Our cognitive tasks were intermittent and the perturbation was triggered by the participants own movement, hence it is possible that participants acted strategically so that cognitive resources were available for the crucial moments of the perturbation. Unfortunately, we did not measure muscle activity and future work will have to address this open question. Finally, this lack of effect might be due to the fact that both the postural and cognitive tasks were relatively easy for healthy, young adults, as evident by excellent performance on the cognitive tasks (mean accuracy on the response inhibition task was 99% and on the memory task the mean accuracy was 84%).

The main limitation of this study was the lack of muscle activity recordings. Due to this, we are unable to answer some open questions with regard to potential changes in the postural control strategy, which should be addressed in the future. Secondly, we are unable to separate the anticipatory and corrective components of postural control. Studies using discrete perturbations can typically define a perturbation phase and a correction phase, which provide valuable information for understanding reactive postural control. In our experiment, the perturbation was continuous and movement dependent, resulting in intermingled perturbation and correction phases. Thirdly, it seems that the cognitive tasks and wide stance posture might have been too easy to elicit significant interactions with the perturbation of the support surface. Standing still with feet hip width apart is a very stable position and future work should incorporate more demanding stance poses, such as feet together, semi-tandem or tandem stance. This would allow us to further define the role of cognition in postural control under conditions of continuous, movement dependent perturbation. However, this stance pose was selected so that we could accurately describe the impact of the perturbation on postural control under the simplest of conditions, which will serve as a starting point for using this perturbation paradigm in studies investigating voluntary movements which start from wide stance, such as stepping. Additionally, this work will provide a basis for similar investigation in older adults, who might not be able to sustain more challenging postures. Studies comparing different populations using this kind of perturbation should consider data normalization with respect to differences in anthropometric characteristics of the participants. One possibility is normalizing the data regarding to the inverted pendulum model and moment of inertia. We investigated a homogeneous group of participants using within participant analysis, hence data normalization would not alter our results.

In conclusion, self-generated, continuous, movement dependent perturbations of the support surface caused a direction-specific response in healthy young adults. This response consisted of increased mean power frequency, variability and range of ML COP sway, indicating a change in postural control strategy, which was insufficient for controlling the increased postural sway. Such prevailing changes in postural sway are striking given the fact that the perturbations delivered were of small amplitude and participants were healthy young adults standing in a stable, wide stance position. The underlying cause of these insufficient responses requires further investigation and might be due to an exploratory role of increased sway. Because sideways falls in older adults, which are frequently self-generated (Robinovitch et al., 2013), often cause hip injuries (Yang et al., 2016) and increase impact on shoulder and elbow (Crenshaw et al., 2017), this type of perturbation provides a method for fall prevention research.

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