INTERACTION EFFECTS OF POSTURE AND UNEVEN GROUND ON ABLE-BODIED WALKING KINETICS

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Interactions between trunk orientation and gait kinetics are proposed to be inevitable for maintaining dynamic balance, and these interactions are unknown for walking on uneven ground. The purpose of this study was to investigate the interaction effects of posture (regular erect, 30°, 50° and 70° trunk flexion) and step category (unperturbed, perturbation, pre- and post-perturbation) on able-bodied walking kinetics. Statistical analysis revealed interactions posture×step: with increased trunk flexion, walking on uneven ground exhibited less changes in GRF kinetic parameters relative to upright walking. Pre-adaptations were more pronounced in the approach step to the drop in regular erect gait. It seems that in trunk-flexed gaits trunk is used in a compensatory way during the step-down to accommodate changes in ground level. In conclusion, exploitation of this mechanism resembles the ability of small birds in adjusting their zig-zag-like configured legs to cope with changes in ground level.

KEY WORDS: Trunk orientation, ground reaction force, step, locomotion

INTRODUCTION: On the one hand, the negotiation of changes in ground level raises challenges to the human locomotor system and requires continuous adaptations to potential perturbations. On the other hand, the dynamics of bipedal locomotion is proposed to be influenced by the orientation of the trunk owing to its significant effect on the position of the center of mass (CoM). The stabilization of a heavy trunk (50 % of total human body mass) is therefore an important functional task in human locomotion. While healthy humans can adapt to frequently faced uneven terrain, patients with impaired postural control are at increased risk of falling even during level walking. Impaired postural control caused by a flexed trunk during walking has been reported to be a major risk factor for falls and new fractures in the elderly (de Groot et al., 2014). On contrary, a forwardly bent trunk induces a gravitational moment that can be utilized to generate greater forward propulsion through the hip (Leroux et al., 2002) which in turn facilitates walking uphill/climbing stairs or to accelerate. At the same time, because the trunk is heavy, a forward bent trunk allows vertical alteration of CoM height (Aminiaghdam et al., 2017, Saha et al., 2008) when changing the hip angle. For example, when approaching the drop during walking, an upward rotation of the trunk during the step down would increase the distance between CoM and foot and thus limit changes in CoM height which in turn would likely lead to less changes in kinetic behaviour.
When human walkers encounter a drop, they modulate gait kinetics proportional with the drop height not only in the perturbed step (drop), but also in the step approach to the drop (Muller et al., 2014). Furthermore, to maintain a dynamic balance during trunk-flexed gait (up to 50° flexion), able-bodied participants were found to adjust gait loading forces by using compensatory kinematic mechanisms in lower limbs. This indicates that interactions between trunk orientation and gait kinetics are inevitable for maintaining dynamic balance, and these interactions are largely unknown for walking on uneven ground. Hence, the aim of this study was to investigate the kinetics of able-bodied gait during the stance phase as a function of trunk posture and step type. We hypothesize that gait kinetic characteristics vary proportionally with an increase of the sagittal trunk flexion and the impact of an altered trunk orientation on them are step-dependent with a more pronounced effects in perturbation step and adaptive strategies in pre- and post-perturbation steps that are different from those of level walking.

METHODS: Kinematic (8 Qualisys cameras, 240Hz) and kinetic data (3 Kistler force plates, 1000 Hz) of 12 able-bodied adults (6 m, 6 f: 26±3.35 years (mean±s.d.); 169.75±7.41 cm height; 65.08±8.07 kg mass) were recorded while walking at self-selected normal speed across two experimental ground conditions involving a level walkway and a walkway with a 10 cm drop with regular erect trunk alignment (RE), with 30° (TF1), 50° (TF2) and maximal trunk flexion (TF3). Trunk angles were compared visually with adjustable-height cardboard templates (drawn with angles displaying target trunk flexion angles) by a second examiner prior to performing of each trial and during gait along the walkway for TF1 and TF2. For TF3, there was no comparison. Trunk angle was defined by the angle sustained by the line connecting the L5-S1 junction and the seventh cervical spinous process and the vertical. From 8 successful trials per subject, first ($VGRF_{1P}$) and second peak ($VGRF_{2P}$) of vertical GRF, loading rate (LR) and unloading rate (UR), dimensionless vertical impulse (VIMP), braking (BIMP) and propulsive (PIMP) impulses were analysed. Impulses were normalized to the product of body weight and the square root of the quotient of leg length and gravity. A 13-body segment model was defined by 21 markers placed on the following bony landmarks: fifth metatarsal heads, lateral malleoli, lateral epicondyles of femurs, greater trochanters, anterior superior iliac spines, posterior superior iliac spines, L5–S1 junction, lateral humeral epicondyles, wrists, acromioclavicular joints, seventh cervical spinous process and middle of the forehead. Kinetic and kinematic data of all successful trials were analysed using custom written Matlab (Mathworks Inc., MA, USA) code. The raw coordinate data were filtered using a fourth-order low-pass, zero-lag Butterworth filter with 12 Hz cutoff frequency. For normally-distributed data, two-way repeated measures ANOVAs were implemented with SPSS using two within-subjects factors: (1) posture (RE, TF1, TF2 and TF3), and (2) step category (unperturbed step 'L'; pre-perturbation 'U-1', perturbation 'U0' and post-perturbation 'U+1' steps during uneven walking) with a statistical significance level of 0.05. Post hoc comparisons were performed using Bonferroni corrections.

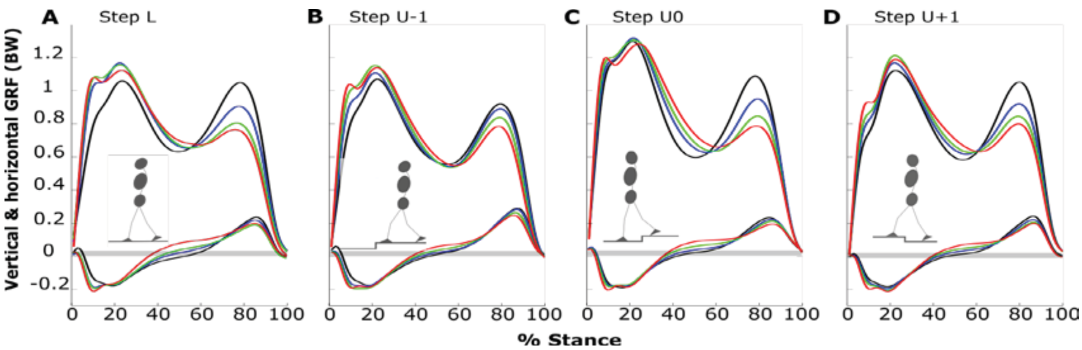

Figure 1: Vertical and horizontal ground reaction forces (GRF) waveforms for different walking conditions. Black, blue, green and red curves represent RE, TF1, TF2 and TF3 gaits, respectively.

RESULTS: The significant posture×step interactions, indicating step-dependent effects of the posture, were detected for the second peak of the vertical GRF ($VGRF_{2P}$) and propulsive impulse (PIMP). Post-hoc tests revealed that the RE gait was associated with a decreased $VGRF_{2P}$ and an increased PIMP in the pre-perturbation step compared with the step 'L', an increased $VGRF_{2P}$ and decreased PIMP in both perturbation and post-perturbation steps, respectively, relative to the pre-perturbation step. While trunk-flexed gaits demonstrated no change across steps in uneven ground relative to the step 'L', trunk-flexed gaits
TF1 gait led to only a significant decrease of the magnitude of the $VGRF_{2P}$ across steps 'L', 'U0' and 'U-1' as compared to the RE gait. TF2 gait was associated with a significant decrease of the magnitude of the $VGRF_{2P}$ across all step types relative to the RE gait. While no step-specific effects of the TF3 gait was found, the magnitude of the $VGRF_{2P}$ decreased across all step types relative to the RE gait and in steps 'U0' and 'U+1' relative to the TF1 gait.
For the main effect of the posture, as compared with the RE gait, the $VGRF_{1P}$ and LR increased, while UR (except for TF1) and VIMP significantly decreased in the gaits with trunk-flexed posture. By contrast, increased sagittal trunk flexion did not lead to a change in BIMP across gaits with bent posture relative to the RE gait. For the main effect of the step, in the pre-perturbation step (U-1) only VIMP increased relative to the unperturbed step (L). In

the perturbation step (U0), $VGRF_{1P}$, UR and VIMP increased in comparison to the step 'L'. As compared with the step 'L', the post-perturbation step (U+1) was associated with a higher UR and a greater VIMP.

Table 1

Means and standard deviations of kinetic parameters. In case of the interaction effect, significant differences from RE, TF1 and TF2 across each step are indicated with 'a', 'b', and 'c', respectively (p<0.05). Accordingly, shaded values indicate the significant difference from the unperturbed step 'L', bold values from the pre-perturbation step 'U-1' and underlined values from the perturbation step 'U0' (p<0.05) for each walking posture (N=12).

		Posture				p-value/F-value		
	Step	RE	TF1	TF2	TF3	Posture	Step	Posture×Step
$VGRF_{1P}$ (BW)	L	1.19 (0.08)	1.33 (0.12)	1.38 (0.13)	1.38 (0.14)	0.00/17.1	0.00/52.1	0.50/0.76
	U-1	1.24 (0.08)	1.34 (0.11)	1.40 (0.14)	1.40 (0.14)			
	U0	1.53 (0.13)	1.63 (0.17)	1.66 (0.20)	1.72 (0.30)			
	U+1	1.25 (0.08)	1.36 (0.12)	1.40 (0.14)	1.41 (0.16)			
$VGRF_{2P}$ (BW)	L	1.15 (0.06)	0.96 (0.10)[a]	0.89 (0.10)[a]	0.87 (0.07)[a]	0.00/86.6	0.19/1.65	0.00/8.97
	U-1	1.06 (0.07)	0.96 (0.11)	0.93 (0.13)[a]	0.90 (0.10)[a]			
	U0	**1.19 (0.10)**	1.01 (0.09)[a]	0.92 (0.12)[a]	0.86 (0.11)[a,b]			
	U+1	**1.20 (0.07)**	1.00 (0.08)[a]	0.93 (0.10)[a]	0.89 (0.09)[a,b]			
LR (BW/s)	L	10.6 (1.70)	12.8 (1.91)	13.5 (1.90)	12.9 (1.72)	0.00/9.19	0.13/2.11	0.07/2.37
	U-1	12.3 (1.46)	14.8 (2.11)	14.4 (3.25)	12.6 (1.80)			
	U0	11.5 (1.43)	13.5 (2.19)	12.7 (2.63)	11.1 (1.60)			
	U+1	12.0 (2.89)	14.7 (2.70)	14.0 (3.34)	13.2 (2.45)			
UR (BW/s)	L	9.21 (1.25)	7.87 (1.02)	6.94 (1.16)	6.60 (1.17)	0.00/22.1	0.00/6.06	0.06/3.11
	U-1	8.89 (1.10)	8.47 (1.44)	7.95 (1.39)	7.65 (1.14)			
	U0	9.90 (0.98)	9.11 (2.61)	7.94 (1.72)	7.20 (0.83)			
	U+1	10.0 (1.21)	8.97 (2.67)	8.04 (1.68)	7.51 (0.98)			
VIMP	L	1.84 (0.12)	1.75 (0.15)	1.70 (0.14)	1.70 (0.13)	0.00/23.0	0.00/20.9	0.10/2.04
	U-1	1.89 (0.13)	1.84 (0.15)	1.80 (0.13)	1.74 (0.15)			
	U0	1.96 (0.11)	1.88 (0.16)	1.84 (0.15)	1.80 (0.19)			
	U+1	2.01 (0.12)	1.91 (0.14)	1.87 (0.13)	1.82 (0.16)			
BIMP	L	-0.10(0.02)	-0.10 (0.03)	-0.09 (0.03)	-0.09 (0.03)	0.55/0.71	0.06/3.33	0.07/2.28
	U-1	-0.11(0.03)	-0.11 (0.04)	-0.12 (0.04)	-0.11 (0.03)			
	U0	-0.12(0.02)	-0.11 (0.02)	-0.10 (0.02)	-0.11 (0.02)			
	U+1	-0.11(0.02)	-0.11 (0.02)	-0.11 (0.02)	-0.11 (0.02)			
PIMP	L	0.13 (0.01)	0.12 (0.01)	0.12 (0.02)	0.13 (0.02)	0.30/1.26	0.00/8.13	0.00/6.91
	U-1	0.16 (0.02)	0.15 (0.02)	0.14 (0.02)	0.14 (0.02)			
	U0	**0.11 (0.02)**	0.12 (0.02)	0.12 (0.02)	0.13 (0.02)			
	U+1	**0.13 (0.02)**	0.13 (0.02)	0.13 (0.02)	0.13 (0.03)			

DISCUSSION: In this study, the adaptive kinetic behavior of able-bodied walking while negotiating uneven ground with altered trunk orientations was investigated. The posture×step interaction effects revealed that trunk-flexed gaits led to only a reduction in the $VGRF_{2P}$ across the steps in uneven ground compared with RE gait. Furthermore, the dynamics of such gaits remain almost unchanged across uneven ground.

Owing to an earlier toe-off at a steeper effective leg (connecting hip to center of pressure) angle, a trunk-flexed gait in human is associated with a more flexed leg joints leading to a significant decrease of the effective leg length at toe-off (Aminiaghdam et al., 2017). In fact, such a kinematic behavior may lead to an insufficient push-off. Furthermore, a combination of a longer propulsive phase and a lesser magnitude of the propulsive force might be a possible cause for observing insignificant differences between trunk-flexed gaits and normal walking in propulsive impulse. Despite a significant impact of trunk-flexed gaits in right skewing the GRF profile (i.e. higher peaks at the beginning and lower ones at the end of the stance phase proportional with sagittal trunk orientation), GRF pattern tended to remain consistent across steps in trunk-flexed gaits. Such a right-skewed profile of vertical GRF implies higher weight acceptance loads (loading rate), a lower push-off (unloading rate) and an overall loading (impulse). Furthermore, the kinetic and kinematic adjustments while crossing uneven ground depends on whether the drop is visible or camouflaged, i.e. the lesser the visual perception of the perturbation, the more pronounced compensations (Muller, et al. 2014). In an unexpectedly lowered step, due to a mismatch between the produced and required muscle force at the moment of impact, the initial impact peak increases (van der Linden, et al. 2009). In this study participants were young healthy volunteers who were successful in accommodating uneven ground with a various degree of the trunk flexion. The quality of their performance has likely been guided by a visual perception of the perturbation so that they could adjust their locomotor behavior using feed-forward strategies.

In addition, the main effect of the posture revealed that loading occurred at higher rates in response to a forward lean of the trunk. This leads to a faster deceleration of the CoM during weight acceptance and suggests a swift transition of the body weight from the contralateral limb to the support limb. Owing to a significant decrease of the vertical GRF at the end of the stance phase and likely a longer duration of the swing phase, participants exhibited a slower unloading rate in trunk-flexed gaits compared with RE gait.

In the trunk-flexed gaits the vertical impulse diminished due to a decreased magnitude of the vertical GRF and a shorter contact time. Walking with a forwardly bent posture alters the relationship between the CoM trajectory and moving base of support, suggesting that the control of balance when CoM is shifted forward requires a higher cadence to support and transfer body weight to the opposite leg which in turn leads to a decreased vertical impulse per step (Aminiaghdam et al., 2017). In the perturbation step (drop), participants experienced ~16% larger magnitude of the first peak of the vertical GRF, ~9% increase of the vertical impulse and ~7% higher unloading rate relative to the step 'L'. Stepping down is probably associated with higher landing velocities of the CoM which possibly leads to a larger first peak of the vertical GRF. Human walkers negotiate visible and camouflaged drops in ground level using the same motor strategy (Muller et al., 2014). Furthermore, since a higher unloading rate indicates a faster intra-limb transition of the body weight, considering a shorter swing phase of the contralateral limb due to an earlier landing on an elevated surface (post-perturbation step), the shift of the body weight during pre-swing phase likely occurs rapidly.

While flexed trunk in elderly patients has been identified as a risk factor for fall during walking (de Groot et al., 2014), in able-bodied trunk-flexed gaits, the trunk may be used in a compensatory way during the step-down to accommodate changes in ground level by adjusting its angle leading to less variations in CoM height. Exploitation of this mechanism would resemble the ability of small birds in adjusting their zig-zag-like configured legs to cope with large ground level perturbations.

CONCLUSION: While increased sagittal trunk flexion leads to significant changes in human GRF profile, negotiation of changes in ground level with trunk-flexed gaits as opposed to regular upright walking is associated with GRF parameters that are more consistent. In the approach step to the drop, walking with regular upright trunk requires modulation of the GRF. In contrast, in trunk-flexed gaits the upper body seems to be transformed into an active component of the human locomotor system by adjusting its angle during the step-down. This compensatory mechanism helps to accommodate changes in ground level leading to less variation in CoM height.

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