

RESEARCH ARTICLE

The influence of sagittal trunk lean on uneven running mechanics

Soran AminiAghdam^{1,*}, Reinhard Blickhan² and Kiros Karamanidis¹

ABSTRACT

The role of trunk orientation during uneven running is not well understood. This study compared the running mechanics during the approach step to and the step down for a 10 cm expected drop, positioned halfway through a 15 m runway, with that of the level step in 12 participants at a speed of 3.5 m s^{-1} while maintaining self-selected ($17.7 \pm 4.2 \text{ deg}$; mean \pm s.d.), posterior ($1.8 \pm 7.4 \text{ deg}$) and anterior ($26.6 \pm 5.6 \text{ deg}$) trunk leans from the vertical. Our findings reveal that the global (i.e. the spring-mass model dynamics and centre-of-mass height) and local (i.e. knee and ankle kinematics and kinetics) biomechanical adjustments during uneven running are specific to the step nature and trunk posture. Unlike the anterior-leaning posture, running with a posterior trunk lean is characterized by increases in leg angle, leg compression, knee flexion angle and moment, resulting in a stiffer knee and a more compliant spring-leg compared with the self-selected condition. In the approach step versus the level step, reductions in leg length and stiffness through the ankle stiffness yield lower leg force and centre-of-mass position. Contrariwise, significant increases in leg length, angle and force, and ankle moment, reflect in a higher centre-of-mass position during the step down. Plus, ankle stiffness significantly decreases, owing to a substantially increased leg compression. Overall, the step down appears to be dominated by centre-of-mass height changes, regardless of having a trunk lean. Observed adjustments during uneven running can be attributed to anticipation of changes to running posture and height. These findings highlight the role of trunk posture in human perturbed locomotion relevant for the design and development of exoskeleton or humanoid bipedal robots.

KEY WORDS: Trunk posture, Perturbed locomotion, Spring-mass model, Leg stiffness, Joint stiffness

INTRODUCTION

Running is a low cost and easy accessibility form of physical activity that enhances health and increases longevity (Lee et al., 2017). The popularity of running continues to grow worldwide. In 2019 alone, millions of recreational participants around the world covered 1.3 billion total miles with an average distance of 4.1 miles per run (<https://2019.strava.com/community/en-us/>). Running outdoors in a natural environment often entails a frequent negotiation of the terrain irregularities such as variations in ground compliance, slipperiness or substrate height. In running over changing surfaces, human runners appear to use spring-mass dynamics to help passively stabilize their

locomotion (Blickhan, 1989; McMahon and Cheng, 1990). The spring-mass model (Blickhan, 1989) is composed of a mass-less spring and the body (represented by a point mass), and is simply described by: leg stiffness (k_{leg}), leg orientation (θ_{TD}) and leg length (l_{TD}) at touchdown (TD) (Blickhan, 1989; McMahon and Cheng, 1990). Albeit the model does not account for the trunk movements, this is a simplistic approach to describe the basic mechanics of human locomotion.


The mechanics of the body's interaction with the ground is influenced by k_{leg} , which is indicative of the average stiffness of the overall musculoskeletal system during the ground contact phase. For running on uneven ground, θ_{TD} and the k_{leg} appear to be adjusted relative to the nature of changes in the substrate height. This involves a flatter θ_{TD} and a more compliant leg when stepping up, but a steeper θ_{TD} and a stiffer leg when stepping down. For the latter situation, runners reduce k_{leg} in the preceding step, possibly to lower the centre of mass (CoM) in preparation for the safe negotiation of expected changes in substrate height, e.g. a drop (Ernst et al., 2014). When traversing an obstacle, k_{leg} decreases in the penultimate step and increases in the final step prior to the barrier in order to accelerate the CoM upward and take-off velocity (Mauroy et al., 2014). Therefore, the fine adjustments of the leg properties not only in the perturbed contact but also in the preceding contacts (Mauroy et al., 2014; Müller and Blickhan, 2010; Müller et al., 2012a, 2010) potentially enhance stability in response to ground perturbations.

The global k_{leg} is determined by the combination of the local stiffness of the joints and the leg geometry at touchdown (Farley et al., 1998; Farley and Morgenroth, 1999; Günther and Blickhan, 2002). Adjustments in both joint stiffness and k_{leg} during the stance phase seem to be governed by initial conditions such as l_{TD} , θ_{TD} and landing velocity (Blickhan et al., 2007; Farley et al., 1998; Farley and Morgenroth, 1999; Grimmer et al., 2008; Mauroy et al., 2014), which are previously geared by a swing-leg retraction throughout the second half of the swing phase prior to landing (Seyfarth et al., 2003). Therefore, muscle pre-activation plays a major role in adjusting the leg posture in preparation for traversing, for example, an obstacle. While adjustments in leg mechanics have been widely proven as key to the accommodation of uneven ground, in this context, little is known about the role of trunk posture during uneven running. Given the ~50% contribution of the trunk segment to total body mass (De Leva, 1996), mechanical demands of the lower limbs can be potentially influenced by even slight alterations in trunk kinematics. Moreover, analysis of an altered mechanical behavior of the lower limbs evoked by trunk posture modifications against an expected substrate height change (i.e. induced deviations from the cyclic motion) could help the understanding of human locomotion.

There are a limited number of studies that have addressed the influence of an anterior trunk lean on knee joint energetics (Arendse et al., 2004; Teng and Powers, 2015), knee muscle activity (Teng and Powers, 2016), patellofemoral joint stress (dos Santos et al., 2019; Teng and Powers, 2014) and impact loading (Huang et al., 2019) during running over uniform ground surfaces. For running on an uneven terrain, one study showed that runners

¹Sport and Exercise Science Research Centre, School of Applied Sciences, London South Bank University, London SE1 0AA, UK. ²Department of Motion Science, Institute of Sport Sciences, Friedrich Schiller University Jena, Seidelstraße 20, 07740 Jena, Germany.

*Author for correspondence (aminiags@lsbu.ac.uk)

 S.A., 0000-0001-9310-6768

List of symbols and abbreviations

ATL	anterior trunk lean
BW	body weight
CoM _{TD} , CoM _{TO}	normalized centre of mass at touchdown and at toe-off
F_z	peak normalized vertical ground reaction force
GRF	ground reaction force
k_{ankle}	normalized ankle stiffness
k_{knee}	normalized knee stiffness
k_{leg}	dimensionless leg stiffness
$l_{\text{TD}}, l_{\text{TO}}$	normalized leg length at touchdown and at toe-off
M_{ankle}	peak normalized ankle extension moment
M_{knee}	peak normalized knee extension moment
PTL	posterior trunk lean
STL	self-selected trunk lean
step _→	control step
step _→	approach step
step _↓	step down
t	time
ΔL_{max}	dimensionless maximum leg compression
$\theta_{\text{leg,TD}}, \theta_{\text{leg,TO}}$	leg orientation at touchdown and at toe-off

adopt a more crouched posture and larger k_{leg} (Voloshina and Ferris, 2015), possibly to facilitate adaptations to an unfamiliar environment. As recently shown by a study on recreational runners (Shih et al., 2019), modifications in the trunk orientation are achievable with simple postural instructions following 1 month of training. However, it is unknown how the modification of sagittal plane trunk posture can potentially impact lower limb operation during uneven running. In walking with increasing trunk flexion angles against changes in substrate height, we previously demonstrated that the trunk plays a functional role through extending its angle during stepping down, which is necessary to control the angular momentum of the whole body (AminiAghdam and Blickhan, 2018; Aminiaghdam et al., 2017a). This contributed to an observed elevated CoM trajectory during the step down, presumably to ease the drop negotiation.

An overall objective of our study was to provide insight into the contribution of sagittal plane trunk lean to running mechanics against an expected change in substrate height. Anticipation of changes to running pattern can facilitate stability and manoeuvrability (Dhawale et al., 2019; Müller et al., 2015; Qiao and Jindrich, 2012). We expected runners to exhibit step-to-step modulations in local and global running mechanics to account for expected changes in the substrate height. We hypothesized that runners would exhibit strategies in the global and local running mechanics that are task specific, owing to the anticipatory nature of changes to posture and substrate height. In particular, we expected running with a posterior versus anterior trunk lean to be associated with greater compensatory kinematic adjustments, as a result of a further shift of the ground reaction force (GRF) line of action from the knee joint axis of rotation, resulting in augmented changes in the magnitude of lower limb local and global kinetic parameters.

MATERIALS AND METHODS

Participants

A convenience sample of 12 (six females, six males) volunteer recreational runners (mean±s.d. age 28.5±5.7 years, body mass 65.5±8.6 kg, height 168.9±6.4 cm) gave written, informed consent to participate in the study. The experimental protocol was approved by the local Ethics Committee of Friedrich-Schiller-University Jena and conducted according to the Declaration of Helsinki.

Experimental design and protocol

Data were collected on a 15 m long instrumented track, with two consecutive force plates embedded halfway along (1000 Hz; 9281B, 9287BA, Kistler, Winterthur, Switzerland) and 12 infrared system cameras (250 Hz; MCU1000, Qualisys, Gothenburg, Sweden). Force plates were set at a distance of one step from each other, allowing step lengths ranging from 1.40 to 2.30 m. Kinematics and GRF data were synchronized using a Kistler external trigger and BioWare data acquisition software (Kistler). A 12 body segment model was defined using 19 reflective markers. Joint coordinate standards of the International Society of Biomechanics (Wu et al., 2002, 2005) were applied. Trunk angle was defined by the angle sustained by the line connecting the midpoint between the L5–S1 junction (L5) and the seventh cervical spinous process (C7) with respect to the vertical. Following running with self-selected trunk lean (STL; mean±s.d. 17.7±4.2 deg), participants were instructed to run with anterior trunk lean (ATL; 26.6±5.6 deg) and posterior trunk lean (PTL; 1.8±7.4 deg) within a range in which they felt comfortable when running across even/uneven runways (Fig. 1). Mean trunk flexion angle was calculated as the average sagittal plane trunk posture during the stance phase of running over the level step across all trials and all participants. The order of the ATL and PTL conditions was randomized for each participant. After running on the even, uniform runway, the variable-height force plate at the site of the second contact was visibly lowered by 10 cm and participants ran along the uneven runway (Fig. 1). Practice trials were permitted to allow participants to become familiar with the running velocity and with the desired trunk postures. The participants accomplished 10 valid trials per condition in which each force plate was fully struck with a single foot, whilst the first force plate was set to be hit by the left foot.

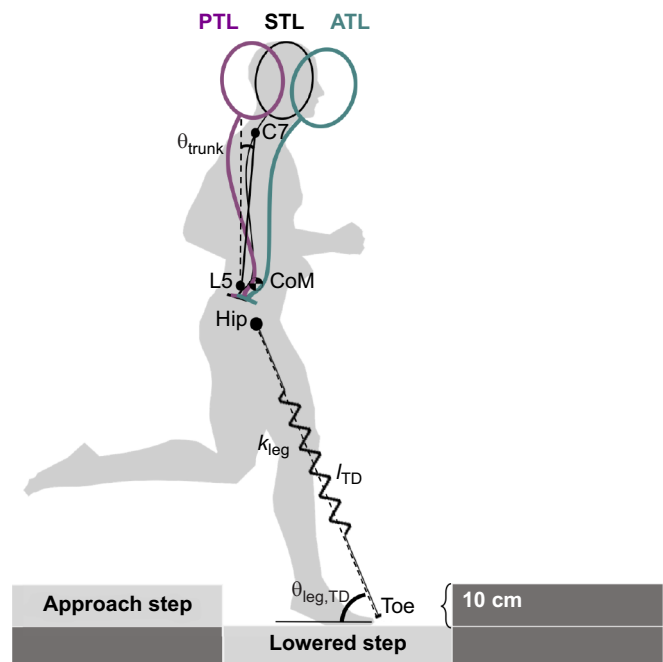


Fig. 1. Schematic illustration of drop negotiation during running with altered trunk postures. The figure depicts the definition of the trunk angle (θ_{trunk}) and the spring–mass model properties as used in this study. A visible drop was created after the approach step (step_→) by lowering the second force plate (step down; step_↓) by 10 cm. STL, self-selected trunk lean; ATL, anterior trunk lean; PTL, posterior trunk lean; CoM, centre of mass; k_{leg} , leg stiffness; l_{TD} , leg length at touchdown; $\theta_{\text{leg,TD}}$, leg angle at touchdown.

Parameters of interest

The ensemble average of the following parameters in the sagittal plane during the stance phase across the level, unperturbed step (step), approach step (step) and step down (step) were determined: (1) vertical position of the CoM relative to the ground, determined by the body segmental analysis method relative to the laboratory coordinate system (De Leva, 1996; Gard et al., 2004); (2) leg length, defined as the length between the hip and the ball of the foot marker (Fig. 1); (3) leg angle, the angle between the leg and the ground (Fig. 1), calculated with respect to the negative x -axis; (4) knee and ankle joint angles; (5) peak vertical GRF (F_z); (6) maximum leg compression (ΔL_{\max}) by subtracting the minimum leg length between TD and toe-off (TO) from L_{TD} ; (7) leg stiffness (k_{leg}) as the ratio between the F_z and ΔL_{\max} ; (8) net knee and ankle joint moments calculated using a rigid linked segment model, anthropomorphic data and inverse dynamics analysis (Zatsiorsky, 1983); and (9) knee (k_{knee}) and ankle (k_{ankle}) joint stiffness calculated from the ratio of the change in net muscle moment to joint angular displacement between TD and the instant when the joints reached maximal flexion. A vertical GRF threshold of $0.03 \times$ body weight was used to determine the instants of TD and TO at each contact (Aminiaghdam et al., 2017b). Leg length, ΔL_{\max} and vertical position of the CoM were normalized to the distance between the greater trochanter marker and the lateral malleoli marker at the instant of TD (l_0). F_z was normalized to the participant's body weight (N/BW). Each joint moment (N m kg⁻¹) and joint stiffness (N m deg⁻¹ kg⁻¹) were normalized to the subject's body mass.

Data processing

For data analysis, we chose all trials that were distributed in a narrow range of each participant's running steady-state velocity (3.5 m s⁻¹). From the calculated mean horizontal velocity of the L5 marker for each of two force plates, we discarded those trials that contained two values differing by more than 5%. Kinetic and kinematic data of all successful trials were analysed using custom written Matlab (Mathworks Inc., Natick, MA, USA) code. The raw coordinate data were filtered using a fourth-order low-pass, zero-lag Butterworth filter with 12 Hz cutoff frequency (Aminiaghdam et al., 2017b). Following confirmation that the data were normally distributed, a two-way repeated measurements ANOVA was used to examine interactions between posture (STL, ATL and PTL) and step (step, step and step) on the parameters of interest (outlined earlier) in SPSS (v. 21.0, IBM® Co.). In the case of a significant interaction, simple main effects were used to compare between-posture changes across each step as well as between-step changes for each individual posture using one-way ANOVA and *post hoc* comparisons with Bonferroni adjustments for multiple comparisons. In the case of a non-significant interaction, the main effects of posture and step were evaluated on each dependent variable of interest. Where Mauchly's test indicated a violation of sphericity, P -values and degrees of freedom were corrected using the Greenhouse–Geisser correction factor. Results are expressed as means \pm s.d. over all participants and parameters. The statistical significance level of all tests was set to $P=0.05$.

RESULTS

The data analysed included 720 trials with a total of 2160 step cycles. Participants were successful in maintaining their stability (no falls) on every trial while running across the level and uneven track. Table 1 summarizes the kinematic and kinetics parameters of uneven running with altered sagittal plane trunk orientations along with the main effects of step and posture and their interaction effects. Fig. 2 illustrates ensemble-averaged global and local sagittal

plane kinematic waveforms, and Fig. 3 illustrates ensemble-averaged joint moment and vertical GRF waveform of running with three trunk orientations across step, step and step.

Two-way repeated measures ANOVA indicated step-specific effects of posture on $\theta_{ankle,TD}$, ΔL_{\max} , $\theta_{ankle,TO}$, $\theta_{leg,TO}$ and $t_{contact}$ (Table 1). *Post hoc* comparisons (simple main effects) revealed significant changes in some kinematic parameters during the step down (step) as compared with those during the level step (step) and the approach step (step), with no between-posture changes across each step (Table 1). In step, the ankle angle demonstrated substantial increases in plantarflexion at TD for all running conditions. Likewise, ΔL_{\max} in step significantly increased by $\sim 35\%$, $\sim 70\%$ and 75% during ATL, STL and PTL postures, respectively, compared with those of step and step (Table 1). The ankle angle at TO of the step increased by $\sim 45\%$ and $\sim 60\%$ during STL and ATL conditions, respectively, compared with that of step (Table 1). Furthermore, the leg became more vertical across all running conditions, as demonstrated by significant decreases in leg angle (~ 4.5 deg during the STL and ATL conditions, and ~ 7 deg during the PTL condition), presumably in preparation for accelerating the body CoM for surmounting the drop. The contact time significantly decreased in step compared with step, irrespective to the trunk orientation (Table 1).

Analyses considering the main effect of posture revealed significant alterations in running mechanics due to changes in the sagittal trunk lean angle. Looking at the global leg parameters, k_{leg} was significantly reduced during the PTL condition by 15% compared with the STL condition, and by $\sim 12\%$ compared with the ATL condition (Fig. 4A), whilst F_z remained unchanged. However, k_{leg} did not significantly differ during ATL running compared with the STL condition (Fig. 4A). In addition, PTL running was characterized by a slightly more vertical leg at TD as compared with that in the ATL condition (Fig. 4D). As for the local leg parameters, the knee flexion angle at TD was slightly greater in both PTL and ATL conditions when compared with the STL condition (Fig. 4E). The peak knee extension moment (M_{knee}) was significantly lower by $\sim 7\%$ during the ATL condition, and was significantly higher by $\sim 15\%$ during the PTL condition when compared with the STL condition (Fig. 4C). As compared with the STL condition, k_{knee} was significantly greater by $\sim 13\%$ during the PTL condition, whereas it remained unchanged during the ATL condition (Fig. 4B). At TO, the knee flexion angle was greater by $\sim 11\%$ during the PTL condition when compared with the STL condition (Fig. 4F).

Furthermore, analyses considering the main effect of step revealed significant changes in kinetic and kinematic parameters of running during step and step when compared with step. In step, runners reduced their k_{leg} by $\sim 24\%$ (Fig. 5G), resulting in a $\sim 13\%$ decrease in the magnitude of F_z (Fig. 5H). At the ankle level, k_{ankle} (Fig. 5I) and M_{ankle} (Fig. 5J) were reduced by $\sim 24\%$ and 12% , respectively, in step. At the end of step, runners lowered their CoM by $\sim 6\%$ (Fig. 5B) using an increased knee flexion (Fig. 5F). This was associated with a $\sim 2\%$ significant decrease in leg length (Fig. 5D). At TD of step, landing involved an increased leg length (Fig. 5C) and a more extended knee joint angle (Fig. 5E), leading to a $\sim 5\%$ higher vertical position of the CoM (Fig. 5A). As compared with values for step, M_{ankle} increased by $\sim 40\%$ (Fig. 5J), while k_{ankle} demonstrated a $\sim 36\%$ decrease (Fig. 5I) in step. k_{leg} was reduced (Fig. 5G), despite a $\sim 27\%$ increase in F_z (Fig. 5H); however, the difference did not reach significance. This was mainly due to a much greater increase in ΔL_{\max} (Table 1) across all running conditions. At the end of the stance phase, the vertical position of the CoM was enhanced by $\sim 4\%$ (Fig. 5B), possibly in preparation to surmount the drop.

Table 1. Kinematics and kinetics of uneven running with altered sagittal plane trunk orientations

Variable	Reference mean	Step	Reference mean – trunk lean mean			<i>P</i> -value/ <i>F</i> -value Partial η^2		
			Self-selected	Anterior	Posterior	Step	Posture	Interaction
CoM _{TD} (<i>l</i> ₀)	0.99±0.04	$\overline{\text{step}}$		0.02±0.04	0.02±0.06	0.001/7.42	0.26/1.41	0.21/1.53
		step	0.05±0.09	0.05±0.07	0.03±0.08	0.41	0.11	0.12
		step↓	−0.04±0.04	−0.03±0.03	−0.02±0.05			
CoM _{TO} (<i>l</i> ₀)	1.01±0.04	$\overline{\text{step}}$		0.02±0.04	0.02±0.06	0.001/9.61	0.18/1.83	0.32/1.21
		step	0.08±0.09	0.09±0.07	0.06±0.07	0.46	0.14	0.09
		step↓	−0.04±0.05	−0.02±0.03	−0.02±0.05			
<i>t</i> _{TD} (<i>l</i> ₀)	0.93±0.02	$\overline{\text{step}}$		0.01±0.02	0.01±0.02	0.001/25.7	0.65/0.22	0.27/1.36
		step	0.01±0.01	0.01±0.02	0.01±0.05	0.71	0.02	0.11
		step↓	−0.06±0.03	−0.05±0.03	−0.06±0.06			
<i>t</i> _{TO} (<i>l</i> ₀)	1.01±0.02	$\overline{\text{step}}$		0.01±0.02	−0.01±0.01	0.001/6.68	0.53/0.43	0.39/0.88
		step	0.01±0.01	0.02±0.02	0.03±0.05	0.37	0.03	0.07
		step↓	−0.01±0.01	−0.01±0.02	0.01±0.05			
$\theta_{\text{leg,TD}}$ (deg)	65.9±2.45	$\overline{\text{step}}$		0.37±3.43	−0.97±3.51	0.001/8.74	0.001/9.22	0.35/1.13
		step	0.05±2.94	0.88±3.95	−1.71±3.61	0.44	0.45	0.09
		step↓	−2.65±1.47	−2.13±1.83	−3.85±1.66			
$\theta_{\text{leg,TO}}$ (deg)	115±2.99	$\overline{\text{step}}$		−1.11±2.91	−2.51±2.96	0.001/97.7	0.001/8.06	0.001/12.3
		step	−3.51±2.53	−3.97±2.61	−5.11±2.66	0.89	0.42	0.52
		step↓	4.56±2.61 ^{a,b}	3.31±2.46 ^{a,b}	4.51±2.24 ^{a,b}			
$\theta_{\text{knee,TD}}$ (deg)	24.1±4.88	$\overline{\text{step}}$		−2.03±3.74	−4.81±5.21	0.001/15.5	0.001/34.1	0.06/3.11
		step	−1.41±4.62	−2.27±4.83	−5.69±6.06	0.58	0.75	0.22
		step↓	3.91±3.97	1.83±5.11	2.63±4.21			
$\theta_{\text{knee,TO}}$ (deg)	17.9±5.32	$\overline{\text{step}}$		0.26±5.98	−1.22±5.08	0.001/62.7	0.001/5.85	0.14/1.81
		step	−7.75±4.01	−9.16±5.61	−10.5±4.03	0.85	0.34	0.14
		step↓	2.11±4.66	0.88±5.69	−0.64±4.84			
$\theta_{\text{ankle,TD}}$ (deg)	4.71±7.42	$\overline{\text{step}}$		0.11±7.12	1.56±7.21	0.001/84.5	0.03/4.05	0.001/7.72
		step	1.65±5.41	0.87±6.54	1.21±7.56	0.88	0.26	0.41
		step↓	−24.9±9.1 ^{a,b}	−23.5±11.7 ^{a,b}	−31.9±5.7 ^{a,b}			
$\theta_{\text{ankle,TO}}$ (deg)	30.7±6.19	$\overline{\text{step}}$		0.74±6.51	−2.18±7.39	0.001/21.1	0.002/12.1	0.001/4.62
		step	7.07±4.26	9.06±4.92	3.24±7.02	0.65	0.52	0.29
		step↓	−3.77±6.37 ^b	−3.71±6.54 ^b	−5.07±6.91			
<i>F_z</i> (BW)	2.61±0.22	$\overline{\text{step}}$		0.06±0.29	0.08±0.31	0.001/266	0.18/1.85	0.32/1.21
		step	0.35±0.21	0.39±0.21	0.39±0.26	0.96	0.14	0.09
		step↓	−0.68±0.29	−0.62±0.29	−0.64±0.32			
<i>k</i> _{leg} (BW/ <i>l</i> ₀)	36.9±10.8	$\overline{\text{step}}$		2.07±11.8	5.34±11.1	0.04/4.74	0.001/17.6	0.15/1.77
		step	9.06±9.03	10.9±7.47	12.3±9.44	0.31	0.61	0.13
		step↓	8.62±9.68	8.19±9.42	14.1±4.41			
ΔL_{max} (<i>l</i> ₀)	0.07±0.01	$\overline{\text{step}}$		−0.01±0.01	−0.01±0.01	0.001/22.4	0.001/28.6	0.01/4.01
		step	−0.01±0.02	−0.01±0.02	−0.02±0.02	0.67	0.72	0.26
		step↓	−0.05±0.02 ^{a,b}	−0.04±0.02 ^{a,b}	−0.07±0.02 ^{a,b}			
<i>M</i> _{knee} (N m kg ^{−1})	2.18±0.52	$\overline{\text{step}}$		0.16±0.51	−0.31±0.49	0.07/2.97	0.001/35.3	0.71/0.38
		step	0.13±0.44	0.22±0.48	−0.23±0.41	0.21	0.76	0.03
		step↓	−0.12±0.33	0.03±0.41	−0.42±0.44			
<i>k</i> _{knee} (N m deg ^{−1} kg ^{−1})	0.11±0.02	$\overline{\text{step}}$		−0.01±0.03	−0.01±0.03	0.12/2.25	0.001/6.51	0.07/2.29
		step	0.01±0.02	0.01±0.03	−0.01±0.02	0.17	0.37	0.17
		step↓	−0.01±0.02	−0.01±0.02	−0.01±0.02			
<i>M</i> _{ankle} (N m kg ^{−1})	3.26±0.52	$\overline{\text{step}}$		0.05±0.57	−0.01±0.58	0.001/73.1	0.09/2.63	0.46/0.79
		step	0.41±0.42	0.46±0.43	0.33±0.45	0.86	0.19	0.06
		step↓	−1.33±0.85	−1.32±0.86	−1.34±0.89			
<i>k</i> _{ankle} (N m deg ^{−1} kg ^{−1})	0.22±0.11	$\overline{\text{step}}$		0.01±0.11	0.01±0.11	0.04/4.56	0.04/3.73	0.16/1.95
		step	0.05±0.05	0.06±0.05	0.05±0.07	0.29	0.25	0.15
		step↓	0.07±0.11	0.07±0.08	0.11±0.04			
<i>t</i> _{contact} (s)	0.23±0.01	$\overline{\text{step}}$		0.01±0.01	−0.02±0.02	0.001/51.6	0.001/9.33	0.01/5.11
		step	−0.01±0.02	−0.01±0.02	−0.03±0.02	0.82	0.45	0.31
		step↓	0.02±0.01 ^b	0.02±0.01 ^b	0.01±0.02 ^{a,b}			

Data are means±s.d. The last three columns outline the *P*-values, *F*-values and effect size (partial eta-squared) pertaining to the main effects of step and posture as well as the step×posture interaction. In the case of an interaction effect, significant differences (simple main effect) from the control step ($\overline{\text{step}}$) and step down (step↓) across each running posture are indicated by 'a' and 'b', respectively (*P*<0.05). The leg length (*l*), maximal leg compression (ΔL_{max}) and centre of mass (CoM) were normalized to the distance between the greater trochanter marker and the lateral malleoli marker at the instant of touchdown (*l*₀). step, approach step; *k*, stiffness; θ , angle; *M*, peak net joint moment; *F_z*, peak vertical ground reaction force; *t*, time; TD, touchdown; TO, toe-off.

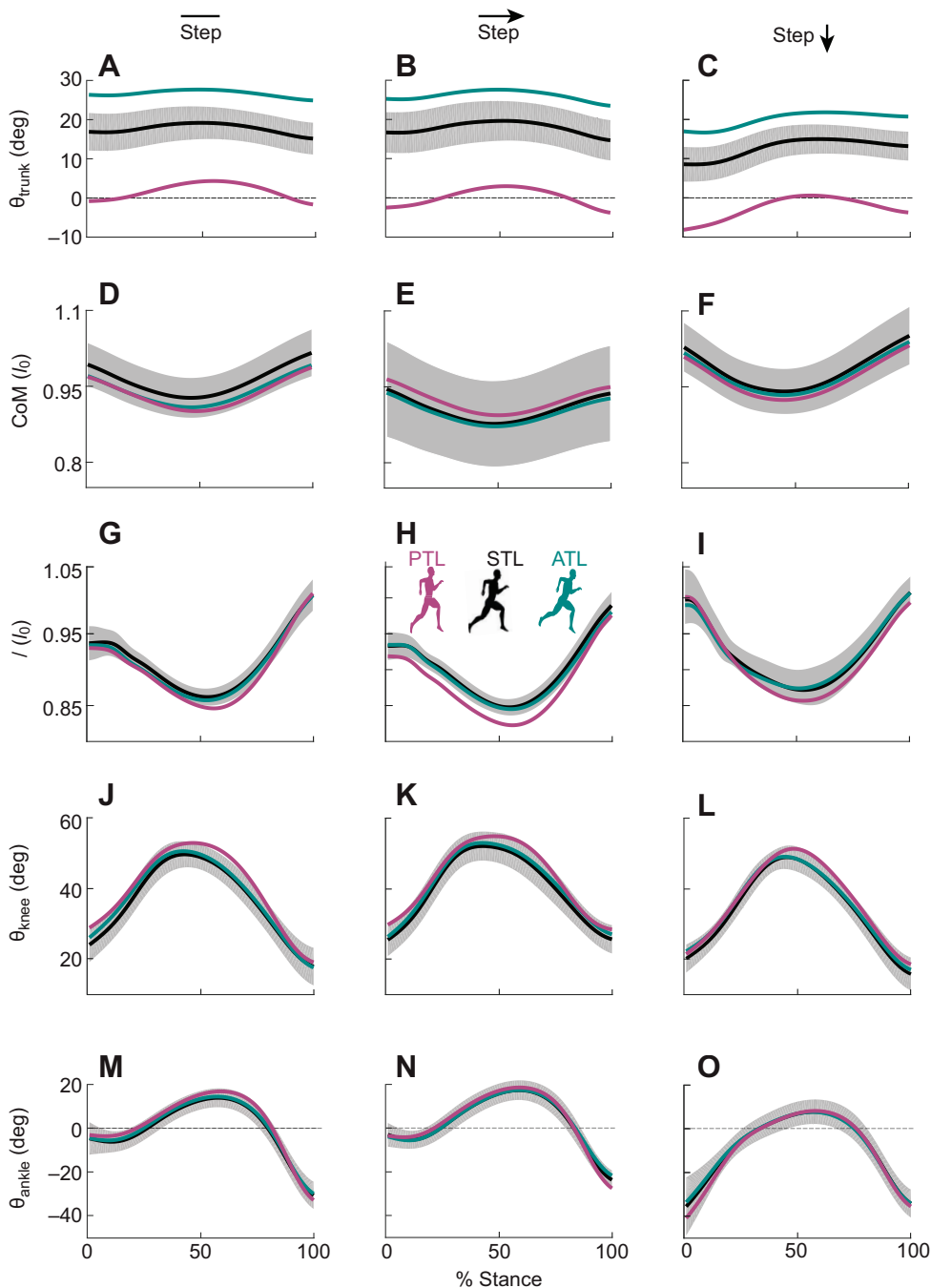


Fig. 2. Global and local sagittal plane kinematic waveforms. Shown are ensemble-averaged trunk angular displacement (A–C), normalized body CoM (D–F), normalized leg length (G–I), knee angular displacement (J–L) and ankle angular displacement (M–O) for STL, ATL and PTL running conditions across the stance phase of step (control step), step and step↓ ($N=12$). The contact time was normalized to 100%. The grey shaded area represents the corresponding s.d. for the STL condition.

DISCUSSION

An overall objective of our study was providing insight into how changing the sagittal plane trunk orientation influences running mechanics against an expected substrate change. It was postulated that runners would exhibit strategies in global and local running mechanics that are task specific, facilitated by the anticipatory nature of changes to posture and substrate height. This was confirmed by the findings of the present study, as running with a posterior versus anterior trunk lean was associated with greater compensatory kinematic adjustments, resulting in augmented changes in the magnitude of lower limb local and global kinetic parameters. However, runners exhibited step-to-step modulations in local and global running mechanics to account for expected change in the substrate height. Interestingly, the control of body

motion for the accommodation of expected substrate height changes appears not to be affected by trunk orientation in running humans.

Posture-related adjustments in running mechanics

When considering the main effect of posture (averaging across the steps), we observed posture-specific adjustments in locomotor behaviour. These results are consistent with those of our previous studies analysing trunk-flexed walking (with the same experimental setup), which revealed that leaning the trunk forward up to 30 deg does not lead to between-step changes in the kinematics (Aminiaghdam et al., 2017a) and kinetics of the braking phase of stance (Aminiaghdam and Rode, 2017) in able-bodied gait. Running with ATL is characterized with a slightly more crouched

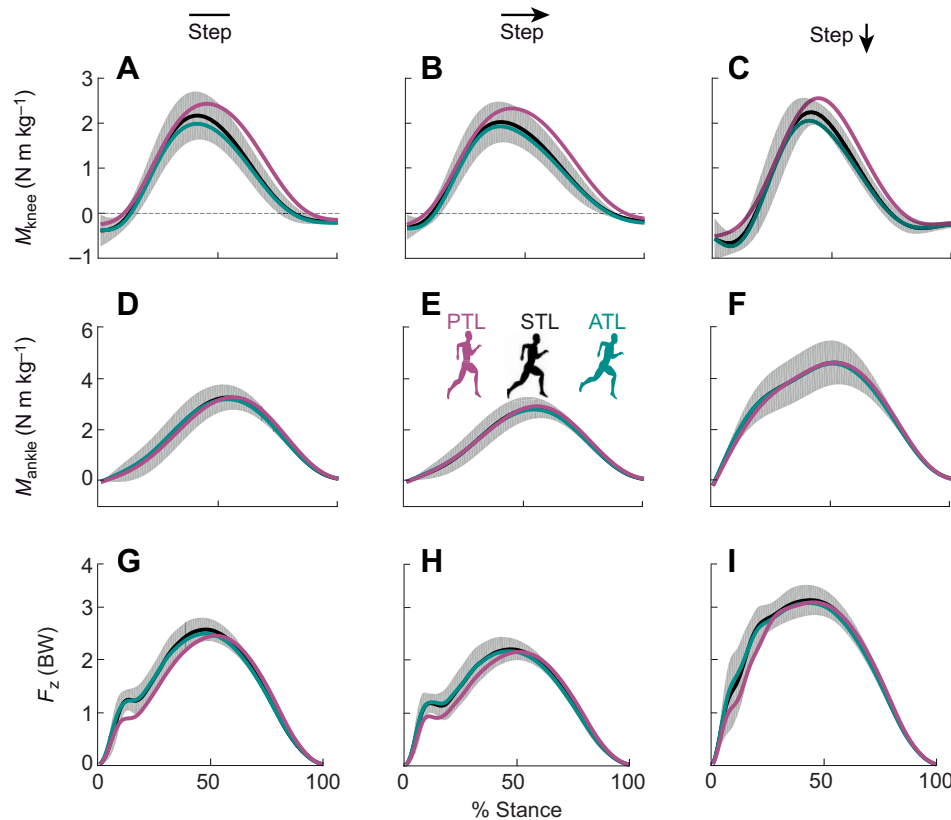


Fig. 3. Joint moment and vertical ground reaction force (GRF) waveforms. Shown are ensemble-averaged normalized knee extensor moments (A–C), ankle extensor moments (D–F) and vertical GRF (F_z) normalized to body weight (BW) (G–I) for STL, ATL and PTL running conditions across the stance phase of step, step and step (N=12). The contact time was normalized to 100%. The grey shaded area represents the corresponding s.d. for the STL condition.

configuration than that of the STL condition, induced possibly by a forward shift of the CoM. Further, along with an increased knee flexion at TD (Fig. 4E), the knee moment reduced significantly by ~7% (Fig. 4C) when compared with the STL condition. This finding is in agreement with a previous study by Teng and Powers (2014) showing a ~7% decrease in knee extensor moment at the time of peak patellofemoral joint stress during level running when increasing sagittal trunk lean by ~7 deg from a self-selected trunk posture (Teng and Powers, 2014). By contrast, the study by dos Santos et al. (2019) revealed no significant decrease in the peak knee extensor moment (1.55 ± 0.32 versus 1.62 ± 0.36 N m kg⁻¹) when runners increase the trunk inclination angle (by ~9 deg) from

their self-selected trunk posture. Moreover, an anterior lean of the trunk does not induce significant changes in joint stiffness across the knee and ankle joints and thus the leg stiffness tends to remain virtually the same as for STL running (Fig. 4A, Table 1). Likewise, the ankle moment appears not to be influenced by alterations in trunk orientation. These results match those observed in an earlier study (Teng and Powers, 2014) that an increase of the sagittal plane trunk inclination redistributes the energetics across the hip and knee without imposing biomechanical demands on the ankle plantar-flexor muscles during level running.

In contrast, a posterior lean of the trunk yields increases in knee flexion at TD (3.4 deg) and TO (2.2 deg) as well as in the knee

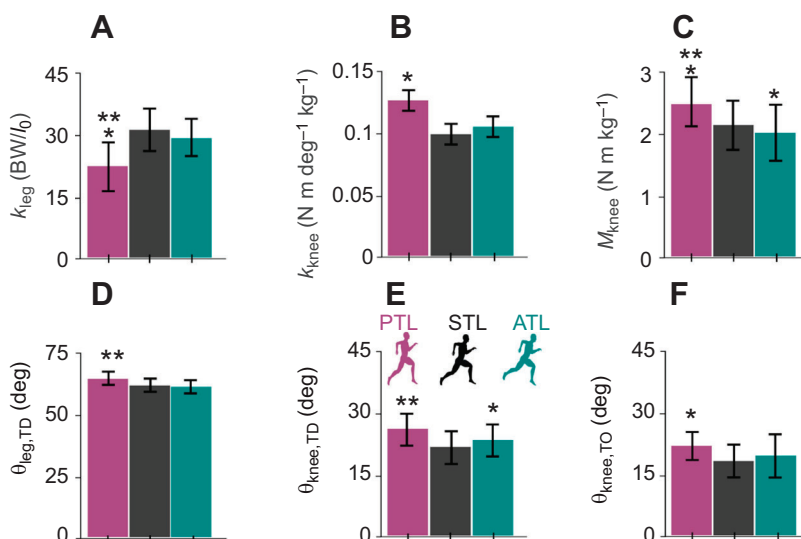


Fig. 4. Main effects of posture. The main effects of posture (mean±s.d.) on variables for which two-way repeated-measurement ANOVA revealed no step×posture interaction (N=12). Leg (A) and knee (B) stiffness (k), peak knee extension moment (M_{knee} ; C), leg (D) and knee (E) angle (θ) at touchdown, and knee angle at toe-off (F). Significant differences from STL and ATL are indicated by * and **, respectively ($P < 0.05$; Bonferroni *post hoc* test). Error bars indicate s.d.

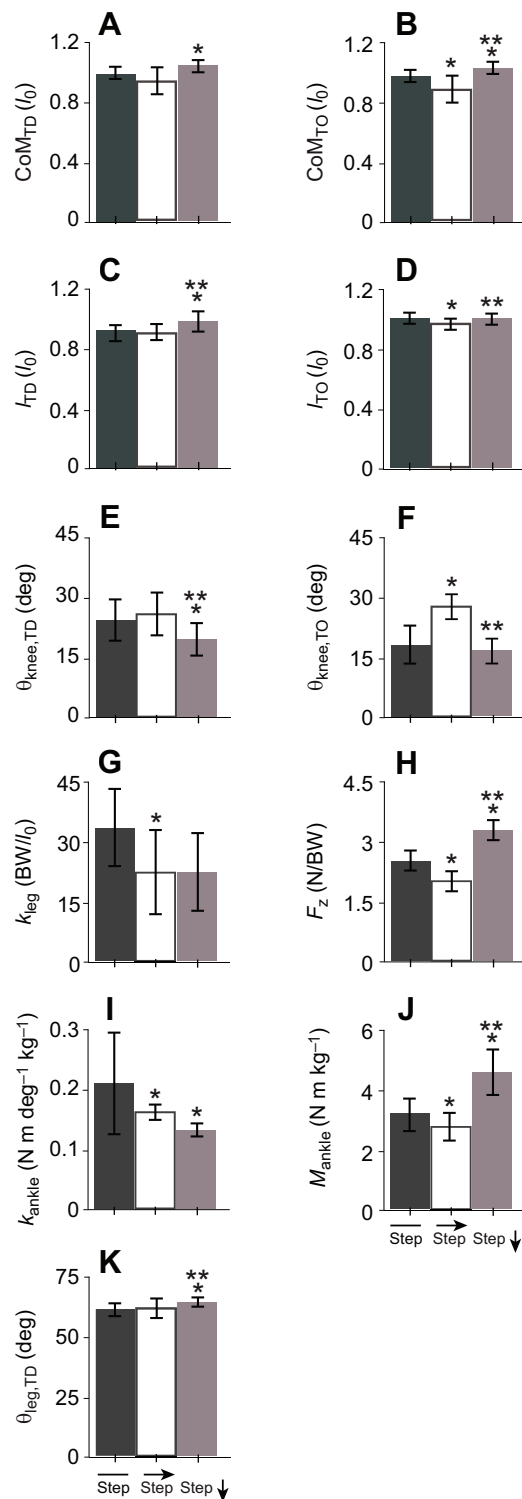


Fig. 5. Main effects of step. The main effects of step (mean \pm s.d.) on variables for which two-way repeated-measurement ANOVA revealed no step \times posture interaction ($N=12$). CoM at touchdown (A) and toe-off (B) normalized to leg length (l_0), normalized leg length at touchdown (C) and toe-off (D), knee angle at touchdown (E) and toe-off (F), leg stiffness normalized to body weight (G), peak vertical GRF normalized to body weight (H), ankle stiffness (I), peak normalized ankle extension moment (J) and leg angle at touchdown (K). Significant differences from step and step are indicated by * and **, respectively ($P < 0.05$; Bonferroni *post hoc* test). Error bars indicate s.d.

moment ($\sim 15\%$) when compared with STL running (Fig. 4). An increased extension moment implies a higher muscle activation of the knee extensors, leading to a higher k_{knee} (Fig. 4B). Teng and Powers (2014) reported a $\sim 5\%$ increase in the peak knee extensor moment when running with a more extended sagittal trunk lean (Teng and Powers, 2014). A lower extent of the decrease in the extension moment compared with that of our study can be explained by the fact that the degree of posterior lean in our study was profoundly greater. This, in turn, induces more compensatory kinematic adaptations in leg posture. In another similar study, Teng and Powers (2016) demonstrated that weak hip-extensor muscles prompt the adoption of a more upright trunk posture during running (Teng and Powers, 2016). Such postural adaptation appears to reduce the demand on the hip extensors, but at the cost of imposing higher demands on the knee extensors. Moving the trunk backwards shifts the line of action of the GRF away from the axis of rotation at the knee joint, and therefore the larger moment arms result in greater net external moments. This, in turn, increases the angular displacement at the knee joint, and subsequently the compression of the spring-leg. Runners demonstrate adaptations in local stiffness by decreasing k_{ankle} by $\sim 10\%$ and increasing k_{knee} by $\sim 13\%$ (Fig. 4B). The local regulation of the joint stiffness coupled with the combination of increased leg compression and decreased F_z yields a reduction in k_{leg} for the PTL condition (Fig. 4B). Consequently, the sagittal plane trunk posture has an influence on biomechanical demands (global and local) of the lower limb during running, but this is independent of changes in substrate height.

Step-related adjustments in running mechanics

It appears that the anticipation of substrate height change governs the control scheme of running. Runners exhibit substantial adjustments in body mechanics, namely feed-forward strategies to accommodate substrate height changes while adopting various trunk orientations. This includes changes in the spring-mass properties (i.e. l_{TD} , $\theta_{leg,TD}$ and k_{leg}) and local mechanics across the knee and ankle joints. The properties of the spring-mass model are modulated with an attenuation bias in the preceding contact when coping with drops and downward steps, and with an augmentation bias in the lowered level (Müller et al., 2012a). These adjustments are utilized to smooth the CoM kinematics (Blickhan et al., 2007). In our study, analysis of the main effect of step revealed the global and local adjustments in the locomotor behaviour specific to the step nature during uneven running. In the approach step, runners demonstrated a $\sim 24\%$ reduction in leg stiffness in preparation to step down when compared with the control step (Fig. 5G). Likewise, ankle stiffness was reduced by $\sim 24\%$ (Fig. 5I). As a result, global leg stiffness appears to be mainly adjusted locally at the level of the ankle joint. These modulations in local and global stiffness gave rise to changes in the magnitude of the leg force, which decreased by $\sim 13\%$ (Fig. 5H). Given the joint stiffness also depends on the activation level of the muscles acting on the joint (Müller et al., 2010, 2012b), an attenuation of global leg stiffness may be due to decreased muscle activation, and a more flexed leg configuration. Here, runners demonstrated a 12% reduction in the ankle moment (Fig. 5J). However, as they approached the end of the stance, knee flexion (Figs 2K and 5F) increased so that the leg length (Figs 2H and 5D) decreased by $\sim 2\%$ and subsequently the CoM was lowered by $\sim 6\%$ at TO (Figs 2E and 5B). As for obstacle negotiation, k_{leg} decreased in the penultimate step and then increased in the final step prior to a 0.65 m-high barrier

in an attempt to enhance the movement of the CoM while leaping the obstacle (Mauroy et al., 2013). It seems that the runners resort to spring-leg stiffness attenuation at the obstacle or upward step, but also in the preparatory step(s) prior to a drop or downward step (Müller et al., 2012a). The findings observed in our study mirror those of previous studies that have reported decreases in the stiffness of the ankle joint proportional to the elevation of the next step during running (Grimmer et al., 2008; Mauroy et al., 2013, 2014; Müller et al., 2012a, 2010). Müller et al. (2012a, 2010) suggested that the modulation of the leg stiffness is actively achieved.

In the step down (drop), the perturbed leg lands in a more vertical orientation with more extended knee and ankle joint positions, resulting in an elongated leg (Figs 2I and 5C). Similarly, Müller et al. (2012a) demonstrated increases in leg angle and leg length when accommodating expected/unexpected drops and expected steps during running. However, they noted that the negotiation of the unexpected versus expected perturbations is associated with greater leg adjustments at TD, due possibly to the lack of feed-forward control of motion prior to the perturbation. It seems that the modulation of the TD leg angle plays a crucial role in the dynamics of stance following the perturbation. Such mechanical behaviour appears to also be exploited by small birds as Daley and Biewener (2006) noted that the extended limb posture at TD of an unexpected drop accounts for 80% of their stance limb loading. Furthermore, given a relatively straight architecture of the human leg, the longer leg at TD is achieved mainly by the ankle plantar flexion (Müller and Blickhan, 2010; Müller et al., 2012a) and slightly by the knee extension. The coupling of these adjustments in the perturbed leg with a posterior motion of the trunk across all running conditions (Fig. 2C) slightly enhances the vertical position of the CoM at TD (Fig. 2F). Seethapathi and Srinivasan (2019) showed that runners adjust leg force over the whole stance and foot placement in a manner to recover the CoM trajectory back to steady state as they run. They further demonstrated the application of these control strategies for locomotion in simulation, showing the robustness of a simple biped against larger discrete perturbations or constant noise-like perturbations. Here, the peak leg force increased by 27% in the perturbed step but decreased by 13% in the preparatory step when compared with the level step. In agreement with Seethapathi and Srinivasan (2019), running humans exhibit adjustments in leg loading and CoM motion in the approach step and the step down (Figs 2D–F and 5A,B) to facilitate the traverse of expected substrate height changes.

In locomotion, the foot segment enables forward propulsion by serving as the link between the ground and the kinetic chain of the lower limb and trunk. Compared with other lower limb moments and forces, the primary contributor to the vertical GRF is the ankle moment (Chen, 2006). The ankle extensor moment exhibits a ~40% increase (Figs 3F and 5J), due potentially to a higher activation of the plantar-flexor muscles, which stems particularly from an increased pre-activation of the m. gastrocnemius medialis (Müller et al., 2015). Our findings accord with those of previous studies in terms of the modulation of leg stiffness (Müller and Blickhan, 2010; Müller et al., 2012a), which have shown that stepping down off an elevation (e.g. single drop of different elevations) is not associated with changes in leg stiffness. This can be due to the extensive leg compression displayed across all running conditions. However, the duration of the ground contact during the lowered (perturbed) step is significantly shorter than during the level steps, irrespective to the running postures (Table 1). Towards the end of the drop, the CoM rises (by ~4%) through greater plantar

flexion and a more vertical leg orientation (regardless of trunk posture), possibly in preparation to surmount the drop (Figs 2F and 5B). The step down therefore appears to be dominated by CoM height changes, regardless of trunk lean.

When interpreting the findings of this study, several limitations need to be acknowledged. First, the runway utilized in the present study may not fully resemble running outdoors, which represents a wide range of expected/unexpected and/or varied magnitudes of surface perturbations, and thus possibly elicits different locomotor outputs. Second, the sample size of the study might be relatively small to reach a convincing conclusion. Third, our results also do not exclude the possibility of being influenced by using an overly conservative Bonferroni correction for multiple comparisons which preserves type I error of the global null hypothesis. Overall, the step-to-step adjustments in various running mechanical parameters exhibit no dependency on sagittal trunk orientation. It is most likely that the observed adjustments are influenced by feed-forward control as both postural and environmental changes were imposed experimentally in an anticipatory fashion. The reliance of the stability of the sagittal-plane dynamics of running on anticipatory strategies when traversing rough terrains (i.e. slope and height variations) has also been validated by a simulation study (Dhawale et al., 2019). In agreement with a previous study (Qiao and Jindrich, 2012), our findings demonstrate that human locomotion employs task-level strategies to account for either or both external and internal perturbations. Characterizing the mechanical behaviour governed by feed-forward control strategies could help better understanding of sensory-motor mechanisms underlying the stabilization of human perturbed locomotion. By illustrating adjustments in global leg-spring stiffness specific to alterations in posture and ground configurations, our findings further support the notion of dependency of the stability of spring-mass running through the global leg stiffness on local joint elasticity, and on leg geometry at TD (Arampatzis et al., 1999; Farley et al., 1998; Farley and Morgenroth, 1999; Günther and Blickhan, 2002; Mauroy et al., 2014). These findings may improve understanding of the role of posture in human perturbed locomotion relevant for the design and development of exoskeleton or humanoid bipedal robots.

Competing interests

The authors declare no competing or financial interests.

Author contributions

Conceptualization: S.A., R.B.; Methodology: S.A., R.B.; Software: S.A., R.B.; Validation: S.A., R.B., K.K.; Formal analysis: S.A.; Investigation: S.A.; Resources: S.A., R.B., K.K.; Data curation: S.A.; Writing - original draft: S.A.; Writing - review & editing: S.A., R.B., K.K.; Visualization: S.A.; Supervision: S.A., R.B., K.K.; Project administration: S.A., R.B., K.K.

Funding

This research received no specific grant from any funding agency in the public, commercial or not-for-profit sectors.

References

- Aminiaghdam, S. and Rode, C. (2017). Effects of altered sagittal trunk orientation on kinetic pattern in able-bodied walking on uneven ground. *Biol. Open* **6**, 1000–1007. doi:10.1242/bio.025239
- Aminiaghdam, S. and Blickhan, R. (2018). The effects of an expected twofold perturbation on able-bodied gait: Trunk flexion and uneven ground surface. *Gait Posture* **61**, 431–438. doi:10.1016/j.gaitpost.2018.02.013
- Aminiaghdam, S., Blickhan, R., Müller, R. and Rode, C. (2017a). Posture alteration as a measure to accommodate uneven ground in able-bodied gait. *PLoS ONE* **12**, e0190135. doi:10.1371/journal.pone.0190135
- Aminiaghdam, S., Rode, C., Müller, R. and Blickhan, R. (2017b). Increasing trunk flexion transforms human leg function into that of birds despite different leg morphology. *J. Exp. Biol.* **220**, 478–486. doi:10.1242/jeb.148312

- Arampatzis, A., Brüggemann, G.-P. and Metzler, V. (1999). The effect of speed on leg stiffness and joint kinetics in human running. *J. Biomech.* **32**, 1349–1353. doi:10.1016/S0021-9290(99)00133-5
- Arendse, R. E., Noakes, T. D., Azevedo, L. B., Romanov, N., Schwellnus, M. P. and Fletcher, G. (2004). Reduced eccentric loading of the knee with the pose running method. *Med. Sci. Sports Exerc.* **36**, 272–277. doi:10.1249/01.MSS.0000113684.61351.B0
- Blickhan, R. (1989). The spring-mass model for running and hopping. *J. Biomech.* **22**, 1217–1227. doi:10.1016/0021-9290(89)90224-8
- Blickhan, R., Seyfarth, A., Geyer, H., Grimmer, S., Wagner, H. and Günther, M. (2007). Intelligence by mechanics. *Phil. Trans. R. Soc. A Math. Phys. Eng. Sci.* **365**, 199–220. doi:10.1098/rsta.2006.1911
- Chen, G. (2006). Induced acceleration contributions to locomotion dynamics are not physically well defined. *Gait Posture* **23**, 37–44. doi:10.1016/j.gaitpost.2004.11.016
- Daley, M. A. and Biewener, A. A. (2006). Running over rough terrain reveals limb control for intrinsic stability. *Proc. Natl Acad. Sci. USA* **103**, 15681–15686. doi:10.1073/pnas.0601473103
- De Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J. Biomech.* **29**, 1223–1230. doi:10.1016/0021-9290(95)00178-6
- Dhawal, N., Mandre, S. and Venkadesan, M. (2019). Dynamics and stability of running on rough terrains. *R. Soc. Open Sci.* **6**, 181729. doi:10.1098/rsos.181729
- dos Santos, A. F., Nakagawa, T. H., Serrão, F. V. and Ferber, R. (2019). Patellofemoral joint stress measured across three different running techniques. *Gait Posture* **68**, 37–43. doi:10.1016/j.gaitpost.2018.11.002
- Ernst, M., Götz, M., Müller, R. and Blickhan, R. (2014). Vertical adaptation of the center of mass in human running on uneven ground. *Hum. Mov. Sci.* **38**, 293–304. doi:10.1016/j.humov.2014.05.012
- Farley, C. T., Houdijk, H. H., Van Strien, C. and Louie, M. (1998). Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. *J. Appl. Physiol.* **85**, 1044–1055. doi:10.1152/jappl.1998.85.3.1044
- Farley, C. T. and Morgenroth, D. C. (1999). Leg stiffness primarily depends on ankle stiffness during human hopping. *J. Biomech.* **32**, 267–273. doi:10.1016/S0021-9290(98)00170-5
- Gard, S. A., Miff, S. C. and Kuo, A. D. (2004). Comparison of kinematic and kinetic methods for computing the vertical motion of the body center of mass during walking. *Hum. Mov. Sci.* **22**, 597–610. doi:10.1016/j.humov.2003.11.002
- Grimmer, S., Ernst, M., Günther, M. and Blickhan, R. (2008). Running on uneven ground: leg adjustment to vertical steps and self-stability. *J. Exp. Biol.* **211**, 2989–3000. doi:10.1242/jeb.014357
- Günther, M. and Blickhan, R. (2002). Joint stiffness of the ankle and the knee in running. *J. Biomech.* **35**, 1459–1474. doi:10.1016/S0021-9290(02)00183-5
- Huang, Y., Xia, H., Chen, G., Cheng, S., Cheung, R. T. H. and Shull, P. B. (2019). Foot strike pattern, step rate, and trunk posture combined gait modifications to reduce impact loading during running. *J. Biomech.* **86**, 102–109. doi:10.1016/j.jbiomech.2019.01.058
- Lee, D.-c., Brellenthin, A. G., Thompson, P. D., Sui, X., Lee, I.-M. and Lavie, C. J. (2017). Running as a key lifestyle medicine for longevity. *Prog. Cardiovasc. Dis.* **60**, 45–55. doi:10.1016/j.pcad.2017.03.005
- Mauroy, G., Schepens, B. and Willems, P. A. (2013). The mechanics of running while approaching and jumping over an obstacle. *Eur. J. Appl. Physiol.* **113**, 1043–1057. doi:10.1007/s00421-012-2519-1
- Mauroy, G., Schepens, B. and Willems, P. (2014). Leg stiffness and joint stiffness while running to and jumping over an obstacle. *J. Biomech.* **47**, 526–535. doi:10.1016/j.jbiomech.2013.10.039
- McMahon, T. A. and Cheng, G. C. (1990). The mechanics of running: how does stiffness couple with speed? *J. Biomech.* **23**, 65–78. doi:10.1016/0021-9290(90)90042-2
- Müller, R. and Blickhan, R. (2010). Running on uneven ground: leg adjustments to altered ground level. *Hum. Mov. Sci.* **29**, 578–589. doi:10.1016/j.humov.2010.04.007
- Müller, R., Grimmer, S. and Blickhan, R. (2010). Running on uneven ground: leg adjustments by muscle pre-activation control. *Hum. Mov. Sci.* **29**, 299–310. doi:10.1016/j.humov.2010.01.003
- Müller, R., Ernst, M. and Blickhan, R. (2012a). Leg adjustments during running across visible and camouflaged incidental changes in ground level. *J. Exp. Biol.* **215**, 3072–3079. doi:10.1242/jeb.072314
- Müller, R., Siebert, T. and Blickhan, R. (2012b). Muscle preactivation control: simulation of ankle joint adjustments at touchdown during running on uneven ground. *J. Appl. Biomech.* **28**, 718–725. doi:10.1123/jab.28.6.718
- Müller, R., Häufle, D. F. B. and Blickhan, R. (2015). Preparing the leg for ground contact in running: the contribution of feed-forward and visual feedback. *J. Exp. Biol.* **218**, 451–457. doi:10.1242/jeb.113688
- Qiao, M. and Jindrich, D. L. (2012). Task-level strategies for human sagittal-plane running maneuvers are consistent with robotic control policies. *PLoS ONE* **7**, e51888. doi:10.1371/journal.pone.0051888
- Seethapathi, N. and Srinivasan, M. (2019). Step-to-step variations in human running reveal how humans run without falling. *ELife* **8**, e38371. doi:10.7554/eLife.38371
- Seyfarth, A., Geyer, H. and Herr, H. (2003). Swing-leg retraction: a simple control model for stable running. *J. Exp. Biol.* **206**, 2547–2555. doi:10.1242/jeb.00463
- Shih, H.-T., Teng, H.-L., Gray, C., Poggemiller, M., Tracy, I. and Lee, S.-P. (2019). Four weeks of training with simple postural instructions changes trunk posture and foot strike pattern in recreational runners. *Phys. Ther. Sport* **35**, 89–96. doi:10.1016/j.ptsp.2018.11.005
- Teng, H.-L. and Powers, C. M. (2014). Sagittal plane trunk posture influences patellofemoral joint stress during running. *J. Orthop. Sports Phys. Ther.* **44**, 785–792. doi:10.2519/jospt.2014.5249
- Teng, H.-L. and Powers, C. M. (2015). Influence of trunk posture on lower extremity energetics during running. *Med. Sci. Sports Exerc.* **47**, 625–630. doi:10.1249/MSS.0000000000000436
- Teng, H.-L. and Powers, C. M. (2016). Hip-extensor strength, trunk posture, and use of the knee-extensor muscles during running. *J. Athl. Train.* **51**, 519–524. doi:10.4085/1062-6050-51.8.05
- Voloshina, A. S. and Ferris, D. P. (2015). Biomechanics and energetics of running on uneven terrain. *J. Exp. Biol.* **218**, 711–719. doi:10.1242/jeb.106518
- Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, D. D., Cristofolini, L. and Witte, H. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *J. Biomech.* **35**, 543–548. doi:10.1016/S0021-9290(01)00222-6
- Wu, G., van der Helm, F. C., Veeger, H. D., Makhssous, M., Van Roy, P., Anglin, C., Nagels, J., Karduna, A. R., McQuade, K. and Wang, X. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. *J. Biomech.* **38**, 981–992. doi:10.1016/j.jbiomech.2004.05.042
- Zatsiorsky, V. (1983). The mass and inertia characteristics of the main segments of the human body. *Biomechanics* **4B**, 1152–1159.