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Soran AminiAghdam, Kiros Karamanidis & Christian Rode

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Uneven running: How does trunk-leaning affect the lower-limb joint mechanics and energetics?

Soran AminiAghdam [a,b], Kiros Karamanidis [a] and Christian Rode [c]

[a]Sport and Exercise Science Research Centre, School of Applied Sciences, London South Bank University, London, UK; [b]Science of Motion, Institute of Sports Science, Friedrich Schiller University Jena, Jena, Germany; [c]Institute of Sports Science, University of Rostock, Rostock, Germany

ABSTRACT

This study aimed to investigate the role of trunk posture in running locomotion. Twelve recreational runners ran in the laboratory across even and uneven ground surface (expected 10 cm drop-step) with three trunk-lean angles from the vertical (self-selected, ∼15°; anterior, ∼25°; posterior, ∼0°) while 3D kinematic and kinetic data were collected using a 3D motion-capture-system and two embedded force-plates. Two-way repeated measures ANOVAs ($\alpha = 0.05$) compared lower-limb joint mechanics (angles, moments, energy absorption and generation) and ground-reaction-force parameters (braking and propulsive impulse) between *Step* (level and drop) and *Posture* conditions. The *Step*-by-*Posture* interaction revealed decreased hip energy generation, and greater peak knee extension moment in the drop-step during running with posterior versus anterior trunk-lean. Furthermore, energy absorption across hip and ankle nearly doubled in the drop-step across all running conditions. The *Step* main effect revealed that the knee and ankle energy absorption, ankle energy generation, ground-reaction-force, and braking impulse significantly increased in the drop-step. The *Posture* main effect revealed that, compared with a self-selected trunk-lean, the knee's energy absorption/generation, ankle's energy generation and the braking impulse were either retained or attenuated when leaning the trunk anteriorly. The opposite effects occurred with a posterior trunk-lean. In conclusion, while the pronounced mechanical ankle stress in drop-steps is marginally affected by posture, changing the trunk-lean reorganizes the load distribution across the knee and hip joints. Leaning the trunk anteriorly in running shifts loading from the knee to the hip not only in level running but also when coping with ground-level changes.

Highlights

- Changing the trunk-lean when running reorganizes the load distribution across the knee and hip joints.
- Leaning the trunk anteriorly from a habitual trunk posture during running attenuates the mechanical stress on the knee, while the opposite effect occurs with a posterior trunk-lean, irrespective to the ground surface uniformity.
- The effect of posture on pronounced mechanical ankle stress in small perturbation height during running is marginal.
- Leaning the trunk anteriorly shifts loading from the knee to the hip not only in level running but also when coping with small perturbation height.

KEYWORDS
Biomechanics; locomotion; posture; knee; perturbation; injuries

Introduction

The popularity of running as a low-cost and easily accessible form of physical activity with potential health benefits continues to grow worldwide. However, a downside is the accompanying high number of injuries (Van Gent et al., 2007). The knee joint is the most vulnerable site for running-related injuries. For instance, increased compressive forces acting on the patellofemoral joint is among the most common mechanisms of patellofemoral pain in runners (Powers, Witvrouw, Davis, & Crossley, 2017), constituting nearly 50% of all knee injuries (Fernández-López & Rojano-Ortega, 2020; Taunton et al., 2002; Van Gent et al., 2007). Therefore, it is imperative to gain insights into the running-related biomechanical risk factors underlying knee injuries.

Recent studies propose the efficacy of manipulating the running gait pattern to systematically redistribute lower-limb joint loading with implications for reducing

CONTACT Soran AminiAghdam soran942@gmail.com; aminiags@lsbu.ac.uk School of Applied Sciences, London South Bank University, 103 Borough Rd, London SE1 0AA, UK

the mechanical stress on the knee joint. For instance, transitioning the running landing technique from rearfoot to forefoot strike pattern shifts the mechanical loading from the knee to the ankle (Arendse et al., 2004; Bonacci et al., 2013; Goss & Gross, 2013; Williams, Green, & Wurzinger, 2012). Increasing the running step-rate results in reduced energy absorption by the hip and knee (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011), and decreasing the step length leads to lower patellofemoral joint stress (Willson, Ratcliff, Meardon, & Willy, 2015; Willson, Sharpee, Meardon, & Kernozek, 2014). Alternatively, an anterior trunk-lean during level running is proposed to manage pain and/or prevent injuries at the knee via shifting mechanical stress to the hip (Arendse et al., 2004; dos Santos, Nakagawa, Serrão, & Ferber, 2019; Goss & Gross, 2013; Huang et al., 2019; Teng & Powers, 2014, 2015, 2016). However, there is a need for running posture and techniques suggestions considering changes in ground surface conditions.

Today, recreational running devotees use city parks, roads, and mountain trails, which contain substantial changes in surface height and compliance. Stepping into a hole or taking a downward step prolongs flight time and thus increases vertical kinetic energy. Controlling the increased centre of mass's energy gained in the drop-step results in a higher mechanical loading of the tri-articulate muscle-skeletal system. Few studies have explored its biomechanical response to (un)expected changes to the running or hopping surface height. The anticipation of 5–10 cm drop or curb while running enables feed-forward control strategies such as lowering the centre of mass in the preceding step or decreasing the vertical ground-reaction-force (GRF) through leg stiffness adjustment during the drop-step (Müller & Blickhan, 2010; Müller, Ernst, & Blickhan, 2012). In unexpected drops, mechanical and within-step feedback driven adjustments can occur that may stem from, e.g. mechanical muscle properties (Tomalka, Rode, Schumacher, & Siebert, 2017). For instance, successful negotiation of lower than 10 cm drops during hopping tasks predominantly relies on increased ankle energy absorption (Dick, Punith, & Sawicki, 2019). In contrast, mechanical loading shifts proximally to the knee and hip for larger drops (20 cm). Thus, the redistribution pattern of lower-limb joints' loading is influenced by the magnitude of surface drops. However, none of these studies investigated the influence of posture on the mechanical behaviour of lower-limb joints. In a recent study (AminiAghdam, Blickhan, & Karamanidis, 2021), we demonstrated that the global (i.e. the spring-mass model dynamics and centre of mass height) and local (i.e. kinematics and kinetics across the knee and ankle) mechanical adjustments during uneven running are specific to the step nature and trunk posture.

Considering that the upper-body represents nearly two-thirds of the total body mass (Zatsiorsky & Zaciorskij, 2002), alterations in the trunk orientation influence the lower-limb mechanical demands of locomotion due to changing orientation (Aminiaghdam, Rode, Muller, & Blickhan, 2017; Müller, Rode, Aminiaghdam, Vielemeyer, & Blickhan, 2017) and position (Sanno, Willwacher, Epro, & Brüggemann, 2018) of the GRF vector in relation to the joints. An anterior trunk-lean during level running decreases knee stress and thus knee pain through reductions in knee extension moment (Arendse et al., 2004; Teng & Powers, 2014), energy absorption (Arendse et al., 2004; Teng & Powers, 2015, 2016) and patellofemoral stress (Teng & Powers, 2014). In contrast, the opposite changes occur when running with a more upright trunk. On the other hand, the negotiation of drop-steps during running involves a more vertical leg orientation and less flexed leg joints in early and late stance (AminiAghdam et al., 2021; Müller & Blickhan, 2010), changing the joint kinetics and energetics. Anterior or posterior trunk-lean are expected to influence joint kinetics and energetics different from habitual trunk orientation when negotiating changes in the ground level. For example, hip extension moments might increase more in anterior trunk-lean in the drop-steps than in habitual trunk orientation because of gravity's increased lever arm supporting upper body forward rotation. Exploring joint-level mechanics of running under a perturbation paradigm such as uneven ground with altered trunk postures would provide insight into mechanical adaptation mechanisms relevant for understanding lower-limb injury risk factors.

This study's primary aim was to evaluate the interaction of changes to the trunk posture and surface height on lower-limb joint (hip, knee, and ankle) mechanics during running. Extrapolating from research on the influence of trunk-lean on level running (dos Santos et al., 2019; Teng et al., 2020; Teng & Powers, 2014, 2016), we specifically hypothesized that leaning the trunk anteriorly (ATL) from the self-selected trunk-lean (STL) would shift mechanical demands from the knee to the hip, and vice versa when leaning the trunk posteriorly (PTL) in the drop-step. Given that the ankle is highly load-sensitive (Daley, Felix, & Biewener, 2007) and plays a predominant role in rebounding the whole-body mass in hopping (Dick et al., 2019), we further hypothesized an increased mechanical contribution of the ankle in response to the drop-step across all running conditions.

Materials and methods

A convenience sample of twelve (six females) volunteer recreational runners (mean ± standard deviation (SD); 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 conducted according to the Declaration of Helsinki.

Data were collected using a twelve-camera motion capture system (250 Hz; MCU1000, Qualisys, Gothenburg, Sweden) and two consecutive force plates (1000 Hz; 9281B, 9287BA, Kistler, Winterthur, Switzerland) embedded halfway along a 15 m long instrumented track. Force plate arrangement allowed step lengths ranging from 1.40 to 2.30 m. We synchronized kinematics and GRF data using an external trigger and BioWare data acquisition software (Kistler Instrument AG, Switzerland). Applying joint coordinate standards of the International Society of Biomechanics (Wu et al., 2002), a twelve-body segment model was defined using nineteen reflective markers (spherical retro-reflective surface, 14 mm). The markers were placed on the following bony landmarks: fifth metatarsal heads, lateral malleolis (ankle), lateral epicondyles of femurs (knee), greater trochanters (hip), anterior superior iliac spines, L5–S1 junction, lateral humeral epicondyles, wrists, acromioclavicular joints, seventh cervical spinous process (C7) and middle of the forehead. The trunk angle was defined by the angle sustained by the line connecting the midpoint between the L5–S1 junction and the C7 w.r.t. the vertical. Mean trunk flexion angle was the average sagittal plane trunk-lean angle during the stance phase of level running. Following running with self-selected trunk-lean, participants were instructed to run with anterior and posterior trunk-leans within a range in which they felt comfortable during running across even and uneven instrumented tracks (supplemental material). After running on an even track, participants faced an expected drop-step (10 cm, at the second, height-adjustable force plate) halfway through the running track. The order of ATL or PTL running was randomized for each participant. Practice trials were permitted to allow participants to become familiar with the running velocity and desired postures. The participants accomplished ten valid runs per condition (wearing their personal shoes) in which they fully struck each force plate with a single foot while hitting the first force plate with the left foot.

We determined the ensemble average of the following variables in the sagittal plane during the stance phase of the right foot (second contact) across the level-step (L0) and the drop-step (D10): (1) lower-limb joint flexion angles calculated as the motion of the distal segment relative to the proximal reference; (2) vertical GRF; (3) braking impulse and propulsive impulse integrating the posterior and anterior GRF over time, respectively; (4) net lower-limb joint moments calculated by inverse dynamics using the GRF, the centre of pressure, a rigid linked segment model, and anthropomorphic data (Zatsiorsky, 1996); (5) net joint power calculated as the dot product of the joint's angular velocity and moment; (6) lower-limb joint energy absorption and generation integrating the negative and positive portions of the power-time curve, respectively. We determined peak values of the variables mentioned in (1), (2), and (4) for statistical evaluation. A vertical GRF threshold of 0.03 body weight defined the instants of foot-touchdown and toe-off at each contact (Aminiaghdam et al., 2017).

For data analysis, we chose all trials completed at a speed of 3.5 m s^{-1} and discarded trials that differed by more than 5% in speed from step to step (calculated from the mean horizontal velocity of the L5 marker for the gait cycle across the force plates). Kinetic and kinematic data of all successful trials were analyzed 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 cut-off frequency (Aminiaghdam et al., 2017). Two-way repeated measuresANOVAs were conducted to examine interaction effects between *Step* (L0 and D10) and *Posture* (STL, ATL, and PTL) on the variables of interest (Table 1) in SPSS (ver 21.0, IBM® Co., USA) with a statistical significance level of 0.05. In the case of a significant interaction, we reported the simple main effects to determine the mean difference for dependent variables of interest between *Posture* conditions at each *Step* level using one-way ANOVA, as well as between *Step* conditions for each *Posture* level using paired *t*-test. To account for multiple comparisons, we used *post-hoc* analysis with conservative Dunn-Bonferroni adjustments. In the case of a non-significant interaction, we evaluated the *Posture* and *Step's* main effects on each dependent variable. Results were expressed as means ± SD over all participants and variables.

Results

The analyzed data includes 720 trials (step cycles). The mean ± SD trunk flexion angles under the STL, ATL, and PTL running conditions were 15.6° ± 4.2°, 24.9° ± 5.7°, and −0.6° ± 6.8°, respectively. Figure 1 illustrates the ensemble-averaged sagittal plane lower-limb joint angles, moments, and power waveforms during the

Table 1. Kinematics, kinetics, and energetics of lower-limb joints during the stance phase of level-step and drop-step of running for the posterior, self-selected and anterior trunk-leans.

		Trunk-lean postures			*p*-value|*F*-value ES		
	Step	PTL	STL	ATL	Step	Posture	Interaction
Peak joint flexion angle (deg.)							
Hip_θ	L0	13.9 ± 4.42[a,b]	27.9 ± 6.52	38.6 ± 8.61[a]	<0.001|43.4	<0.01|96.1	0.04|7.28
	D10	12.1 ± 4.99[a,b]	**22.8 ± 5.99**	**29.9 ± 8.03**	0.79	0.89	0.39
$Knee_\theta$	L0	53.3 ± 3.86	50.1 ± 4.31	51.8 ± 5.23	0.02|7.35	<0.01|13.3	0.27|1.35
	D10	51.89 ± 4.31	49.1 ± 3.49	49.3 ± 3.28	0.41	0.54	0.11
$Ankle_\theta$	L0	28.5 ± 5.41	25.6 ± 5.71	25.8 ± 5.98	0.004|13.1	0.02|12.1	0.09|8.78
	D10	**23.05 ± 6.44**	**22.71 ± 5.54**	**22.61 ± 5.54**	0.54	0.52	0.44
Ground reaction force							
F_Z (N/BW)	L0	2.51 ± 0.28	2.62 ± 0.19	2.57 ± 0.24	<0.001|291	0.05|3.43	0.09|2.59
	D10	3.18 ± 0.31	3.22 ± 0.26	3.14 ± 0.23	0.96	0.23	0.19
I_{br} (N.s/BW)	L0	−0.10 ± 0.02	−0.09 ± 0.02	−0.07 ± 0.02	<0.001|36.2	<0.01|38.7	0.85|0.15
	D10	−0.14 ± 0.02	−0.12 ± 0.02	−0.11 ± 0.01	0.76	0.77	0.01
I_{pro} (N.s/BW)	L0	0.10 ± 0.02	0.10 ± 0.02	0.11 ± 0.02	0.62|0.17	<0.01|10.9	0.42|0.88
	D10	0.09 ± 0.02	0.10 ± 0.02	0.11 ± 0.02	0.01	0.51	0.07
C_t (s)	L0	0.25 ± 0.02[a,b]	0.23 ± 0.01	0.23 ± 0.01	<0.001|27.3	0.01|9.98	0.02|8.01
	D10	**0.22 ± 0.02**	**0.21 ± 0.01**	**0.21 ± 0.01**	0.71	0.47	0.42
Peak joint extension moment (Nm/kg)							
Hip_M	L0	0.85 ± 0.41	1.89 ± 0.43	2.27 ± 0.47	0.004|13.4	<0.01|51.1	0.86|0.11
	D10	1.43 ± 0.82	2.51 ± 0.63	2.88 ± 0.84	0.55	0.82	0.01
$Knee_M$	L0	2.48 ± 0.37[b]	2.06 ± 0.51	1.84 ± 0.56	0.04|4.98	<0.01|31.3	0.04|3.47
	D10	2.61 ± 0.44[b]	**2.31 ± 0.33**	2.15 ± 0.41	0.31	0.74	0.24
$Ankle_M$	L0	3.61 ± 0.78	3.64 ± 0.71	3.64 ± 0.72	<0.001|53.6	0.91|0.08	0.71|0.34
	D10	4.61 ± 0.89	4.61 ± 0.85	4.58 ± 0.86	0.83	0.01	0.03
Joint energy absorption (J/kg)							
Hip_{abs}	L0	−0.32 ± 0.14[a,b]	−0.14 ± 0.09	−0.13 ± 0.11	0.15|2.39	0.01|6.95	0.03|7.91
	D10	−0.28 ± 0.13	**−0.27 ± 0.18**	**−0.22 ± 0.12**	0.17	0.38	0.41
$Knee_{abs}$	L0	−0.52 ± 0.11	−0.43 ± 0.14	−0.34 ± 0.13	0.03|6.04	<0.01|18.5	0.31|1.22
	D10	−0.61 ± 0.17	−0.52 ± 0.12	−0.46 ± 0.15	0.35	0.62	0.11
$Ankle_{abs}$	L0	−0.98 ± 0.37	−0.96 ± 0.37	−1.01 ± 0.39	<0.001|47.4	0.02|4.34	0.01|4.77
	D10	**−2.09 ± 0.49**	**−1.93 ± 0.55**	**−1.91 ± 0.64**	0.81	0.28	0.31
Joint energy generation (J/kg)							
Hip_{gen}	L0	0.13 ± 0.11[a,b]	0.37 ± 0.21	0.51 ± 0.21	0.62|0.25	<0.01|35.3	0.03|7.81
	D10	0.18 ± 0.13[b]	0.33 ± 0.18	0.42 ± 0.19	0.02	0.76	0.41
$Knee_{gen}$	L0	0.39 ± 0.08	0.25 ± 0.06	0.25 ± 0.07	0.11|3.12	<0.01|26.6	0.09|2.64
	D10	0.37 ± 0.08	0.31 ± 0.05	0.31 ± 0.08	0.22	0.71	0.19
$Ankle_{gen}$	L0	0.97 ± 0.17	0.87 ± 0.16	0.86 ± 0.19	<0.001|30.5	<0.01|14.7	0.21|1.67
	D10	1.12 ± 0.21	1.06 ± 0.22	1.06 ± 0.25	0.73	0.57	0.13

The information presented in the last three columns outline the *p*-values, *F*-value, and effect size (ES, partial eta-squared) pertaining to the main and interaction effects of *Step* and *Posture*, respectively. In the case of an interaction effect: "a" = sig. different from STL; "b" = sig. different from ATL; bold value = sig. different from the level step ($p < 0.05$). ES, effect size; PTL, posterior trunk-lean; θ, angle; *M*, moment; *abs*, absorption; gen, generation; F_Z, peak vertical ground-reaction-force; I_{br}, braking impulse; I_{pro}, propulsive impulse.

stance phase of various running conditions in the level-step and drop-step.

An interaction effect of *Posture* and *Step* on the peak hip flexion angle (Hip_θ) revealed a significantly lower angle during PTL running than STL and ATL conditions in both steps (Table 1). Furthermore, the Hip_θ was significantly lowerin the drop-step during ATL and STL running conditions compared with the level-step. There was a significant main effect of *Posture* on the peak knee flexion angle (Figure 2A), showing a greater value during PTL versus STL running. An interaction effect on the peak ankle dorsiflexion angle revealed a reduced angle in the drop-step, regardless of the trunk posture (Table 1).

There were no interaction effects on F_Z, braking (I_{br}) and propulsive (I_{pro}) impulses (Table 1), but significant main effects of *Step* and *Posture*, indicating that variations in the GRF-related parameters due to the drop-step were not trunk posture-dependent. The F_Z and I_{br} (Figure 2B and C) were significantly greater in the drop-step versus the level-step. Compared with STL running, there was greater I_{br} (Figure 2C) and smaller I_{pro} (Figure 2D) during PTL running and vice versa during ATL running. There was no interaction effect on the peak hip extension moment (Hip_M) (Table 1).

The main effects of *Step* and *Posture* revealed significantly higher Hip_M in the drop-step versus the level-step (Figure 2E) and significantly lower and higher Hip_M during PTL or ATL running, respectively, compared with STL running (Figure 2E). The interaction effect on the peak knee extension moment ($Knee_M$) revealed significantly higher values during PTL running in both steps compared with the ATL condition. Furthermore, there was a significantly higher $Knee_M$ in the drop-step versus the level-step during STL running only (Table 1). For the peak ankle extension moment, there was only

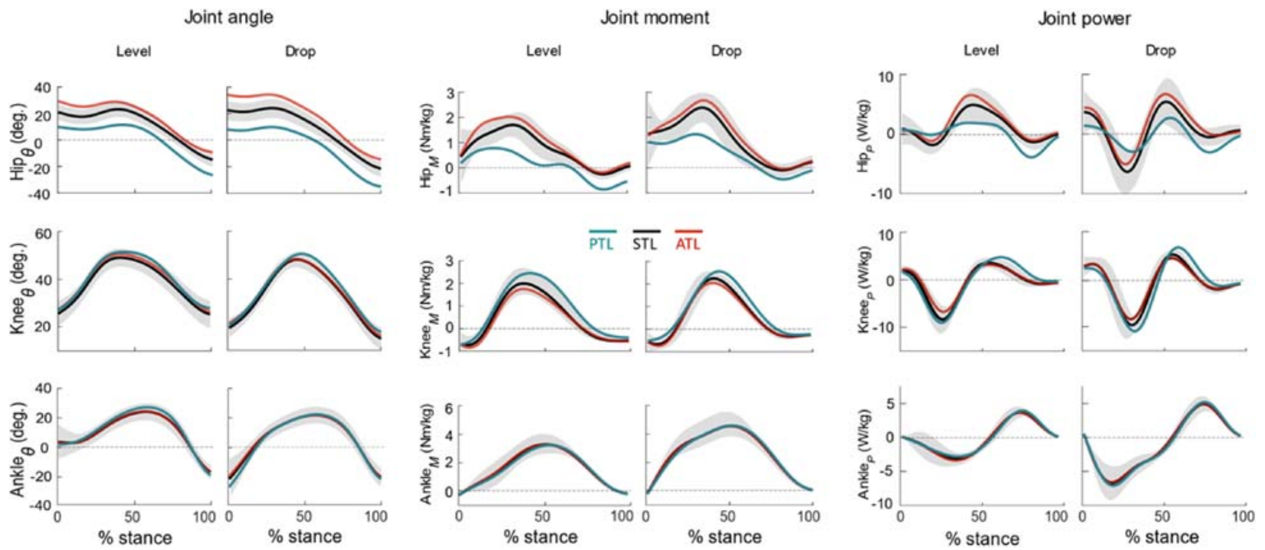

Figure 1. Lower limb joint mechanics. Ensemble-average sagittal plane hip (top row), knee (middle row) and ankle (bottom row) joint angular displacements, moments, and powers for self-selected (STL; black), anterior (ATL; red) and posterior (PTL; green) trunk-leans during the stance phase of running across the level-step (left) and the drop-step (right). The contact time is normalized to 100%. The grey shaded area represents the corresponding SD for the STL condition.

a significant main effect of *Step*, showing a higher value in the drop-step versus the level-step (Figure 2F).

There were interaction effects on the hip energy absorption and generation. During STL and ATL running, the hip energy absorption increased significantly in the drop-step versus the level-step (Table 1). Also, the hip energy generation was significantly higher during PTL running in both steps than other conditions (Table 1). For the knee energy absorption (Knee$_{abs}$), a significant main effect of *Step* revealed a higher value in the drop-step versus the level-step (Figure 2G). *Posture*'s significant main effect revealed a higher Knee$_{abs}$ during PTL running and lower Knee$_{abs}$ during ATL running than STL running (Table 1) and a higher knee energy generation during PTL versus ATL running (Figure 2H). The analysis also revealed an interaction effect on the ankle absorption, indicating a higher value in the drop-step, irrespective of the trunk posture (Table 1). *Step* and *Posture* had main effects on the ankle energy generation, which was higher in the drop-step and during PTL versus STL running (Figure 2I).

Discussion

This study sought to evaluate the lower-limb joint mechanics and GRF parameters in response to the interaction of changes to the trunk posture and the ground level surface during running. Our first hypothesis that leaning the trunk anteriorly from the self-selected trunk-lean would shift mechanical demands from the knee to the hip in the drop-step, and vice versa when leaning the trunk posteriorly, was partially supported. A posterior trunk-lean during running resulted in lower hip angle and energy generation and higher extension moments in the drop-step compared to the self-selected trunk-lean (Figure 1 and Table 1). Leaning the trunk anteriorly attenuated the mechanical loading at the ankle and knee, while opposing effects occurred when leaning the trunk posteriorly (*Posture* main effect). Our second hypothesis regarding an increased mechanical contribution of the ankle to rebound the whole-body mass in the drop-step irrespective of trunk postures was confirmed: the ankle energy absorption nearly doubled in the drop-step during all running conditions (Figure 1).

Changes to the trunk posture during running impact lower-limb joint mechanics. Teng and Powers (Teng & Powers, 2015) reported lower energy absorption (~23%) and generation (~13%) at the knee and higher energy absorption during level running for runners with a high versus low trunk flexion angle. In another study (Teng & Powers, 2014), they revealed that increasing the trunk flexion angle by ~7° (from 7.3° to 14.1°) results in a significantly lower knee extension moment (~7%) and thus a reduced patellofemoral joint stress during level running. In our study, knee joint energetics in the drop-step were not affected when altering trunk posture; however, compared with the level-step, the hip absorption nearly doubled in the drop-step during running with self-selected or anterior trunk-lean. High energy absorption at the hip might coincide with increased eccentric muscle work (due to

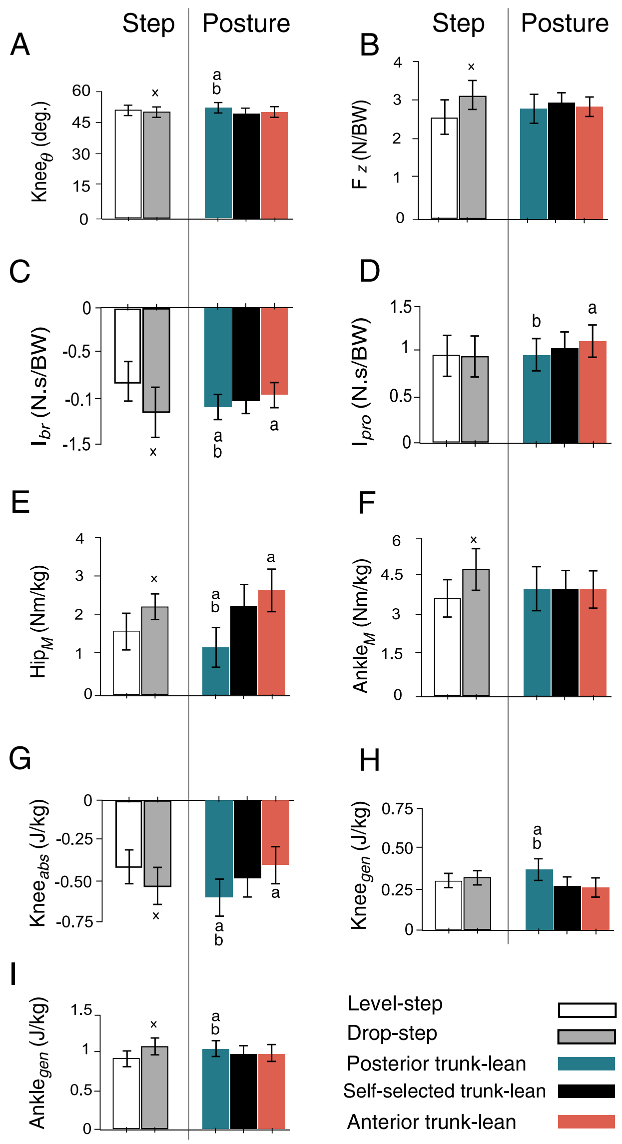

Figure 2. Main effects of *Step* and *Posture*. Shown (mean ± SD) are the main effects on variables that two-way repeated measures ANOVAs did not reveal *Step* × *Posture* interaction. The vertical grey line splits the bar graphs into main effects of *Step* (left) and *Posture* (right). Between-posture differences: "a" = sig. different from self-selected trunk-lean; "b" = sig. different from anterior trunk-lean; between-step differences: "×" = sig. different from the level-step ($p < 0.05$). Error bars denote standard deviation. θ, peak joint angle; *M*, peak net joint moment; *abs*, absorption; gen, generation; F_Z, peak vertical ground-reaction-force; I_{br}, braking impulse; I_{pro}, propulsive impulse.

the low tendon length to fascicle length ratio for the hip extensors, e.g. gluteus maximus). Moreover, posterior trunk-lean running was associated with lower peak hip extension moments (~48%), while leaning the trunk anteriorly led to higher values (~8%) compared with the habitual running posture (Figure 2E). As for the knee energetics, a posterior trunk-lean resulted in higher energy absorption (~20%) and generation (~35%), while leaning the trunk anteriorly led to lower energy absorption (~15%) when compared with the typical running posture (Figure 2G and H). Considering a small contribution (less than 15%) of the hip to sagittal plane total work during the stance phase of running (Jin & Hahn, 2018) and further, that runners habitually adopt different trunk-lean angles (Teng & Powers, 2015, 2016), a distal-to-proximal load shift by an anterior trunk-lean might be an effective technique to reduce pain or prevent injury in runners with knee complaints.

Downward steps are associated with a delayed onset of ground contact, requiring a more robust mechanical response of the limb. This response depends on the hole's visibility (Müller et al., 2012) and the magnitude of the drop (Dick et al., 2019). In line with previous findings on expected drops (Dick et al., 2019; Müller & Blickhan, 2010; Müller et al., 2012), runners exhibited slightly decreased ankle and knee flexion in the 10 cm drop-step. Likewise, anterior trunk-lean only changed the hip angle, not the knee and ankle angles; however, posterior trunk-lean increased the knee flexion. The drop-step was, unsurprisingly, associated with increased vertical force as well as shorter contact time irrespective of the trunk posture. Further, greater braking impulse, extension moment, and energy absorption at the knee and ankle reflect increased mechanical demands in the drop-step.

Interestingly, despite aiming at a steady running speed, leaning the trunk anteriorly decreased braking and increased propulsive impulse. In contrast, the posterior trunk-lean yielded the opposite effects (Figure 2C and D). The changed centre of mass position relative to the foot at touchdown induced by posture alterations might explain these effects. For example, an anterior trunk-lean reduces the distance between centre of mass and foot and thus a shorter braking time before reaching midstance, yielding a decreased braking impulse. Thus, in order to achieve a steady speed with the similar efficiency of running with a habitual posture, running with altered trunk-leans needs compensatory adaptations in motor patterns.

The redistribution patterns of lower-limb joint-level kinetics and energetics vary with alterations to the running technique or the ground level. For example, a recent study (dos Santos et al., 2019) analyzing various level running techniques reported a ~26% decrease in peak knee extension moment, but ~37% increase in peak plantarflexion moment when changing from rearfoot strike to forefoot landing, while adopting anterior trunk-lean resulted in ~5% decrease in peak knee extension moment and no change in peak plantarflexion moment. Thus, changing the foot-strike pattern could place greater mechanical demands on the ankle joint

(Bonacci et al., 2013; Daoud et al., 2012; dos Santos et al., 2019; Goss & Gross, 2012; Williams et al., 2012), and possibly a higher risk of running-related foot and ankle injuries. However, these studies have not investigated resulting mechanical requirements under unsteady locomotion. In response to a 10 cm surface-induced perturbation, our findings show a substantial contribution of the ankle to the whole-body rebound in the drop-step, including ∼30% increase in ankle plantarflexion moment, nearly doubling of energy absorption and ∼20% increase in energy generation irrespective of the trunk posture (Table 1). A similar ankle contribution has been reported when accommodating an unexpected drop (∼10% leg length) during hopping (Dick et al., 2019). In contrast, increasing the slope declination during treadmill running shifts mechanical demands from the ankle to the knee (Vernillo et al., 2017). Although the drop-step increases the mechanical demands across all lower-limb joints, the ankle accounts for the largest part of the excess mechanical load.

Several limitations need to be considered when interpreting the findings of this study. First, our experimental setup was restricted to a single, expected 10 cm drop-step. The lower-limb joint mechanics during running could manifest differently depending on the nature and/or the magnitude of the environmental perturbations. Second, only young, healthy runners were examined at a fixed running velocity. Therefore, caution should be taken when generalizing our results to symptomatic runners and/or to different running velocities. Third, we did not examine the impact of altered trunk postures on the pelvic tilt or the lumbar extension moments; therefore, our results do not exclude the possibility of an augmented or attenuated mechanical load on the low back region when changing posture. Fourth, our study did not consider the foot-strike patterns of runners or other technique modification such as changing the step-rate. Fifth, due to a small sample size, we opted for two-way repeated measures ANOVAs to compare dependent mechanical variables between *Step* and *Posture* conditions; however, recruiting a larger sample size and/or using other statistical methods such as multiple linear regression can provide further insight into changes in lower-limb loading pattern with increasing or decreasing trunk-lean angles. Future perturbation experiments addressing these limitations are needed to provide further insight into the biomechanics of running in real-world settings.

Conclusion

The trunk-lean adjustments appear to reorganize the load distribution patterns across the knee and hip joints with marginal effects on pronounced ankle loading in the drop-step. Our findings illustrate the influence of an anterior trunk-lean on knee load reduction through a distal-to-proximal redistribution of the joint moment and work not only during level running, as reported by previous research, but also when coping with an expected drop-step.

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ORCID

Soran AminiAghdam http://orcid.org/0000-0001-9310-6768
Kiros Karamanidis http://orcid.org/0000-0002-1839-6643
Christian Rode http://orcid.org/0000-0003-1938-2478

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