Title:

Leaning the trunk forward decreases patellofemoral joint loading during uneven running

ABSTRACT

While decline surfaces or a more upright trunk posture during running increase the patellofemoral joint (PFJ) contact force and stress, less is known about these kinetic parameters under simultaneous changes to the running posture and surface height. This study aimed to investigate the interaction between *Step* (10-cm drop-step and level step) and *Posture* (trunk angle from the vertical: self-selected, ~15°; backward, ~0°; forward, ~25°) on PFJ kinetics (primary outcomes) and knee kinematics and kinetics as well as hip and ankle kinetics (secondary outcomes) in twelve runners at 3.5 ms-1. Two-way repeated measures ANOVAs ($\alpha = 0.05$) revealed no step-related changes in peak PFJ kinetics across running postures; however, a decreased peak knee flexion angle and increased joint stiffness in the drop-step only during backward trunk-leaning. The *Step* main effect revealed significantly increased peak hip and ankle extension moments in the drop-step, signifying pronounced mechanical demands on these joints. The *Posture* main effect revealed significantly higher and lower PFJ kinetics during backward and forward trunk-leaning, respectively, when compared to the self-selected condition. Forward trunk-leaning yielded significantly lower peak knee extension moments and higher hip extension moments, whereas the opposite effects occurred with backward trunk-leaning. Overall, changes to the running posture, but not to the running surface height, influenced the PFJ kinetics. In line with the previously reported efficacy of forward trunk-leaning in mitigating PFJ stress while even or decline running, this technique, through a distal-to-proximal joint load redistribution, also appears effective during running on surfaces with height perturbations.

Key words: locomotion, posture, joint mechanics, knee, injury, uneven ground

INTRODUCTION

The popularity of running continues to increase, due to being a low cost and easily accessible form of physical activity with obvious health advantages (16). In 2019 alone, recreational runners globally ran over 2 billion kilometres with an average distance of 6.5 kilometres per run (29). The vast majority of running is undertaken outdoors (30), where natural terrain is rarely flat. This requires runners to frequently modulate their locomotor behavior for expected or unexpected alterations in surface compliance and/or surface height (20). Considering the potential impact of uneven surface on the dynamics of human locomotion (12), the analysis of running on uniform even surfaces (e.g. on a regular treadmill) may less reflect an accurate representation of habitual running experience in a real-world setting (13). Hence, the results from experimental analyses resembling uneven surface are more likely to directly translate to running in a field environment.

Despite the health benefits of running, there is a high prevalence of running-related lower-limb injuries (i.e., 19% to 79%) (40), mainly due to the repetitive nature of biomechanical loading during running (37). The most frequently reported running-related symptom is patellofemoral pain (PFP; 25% of all running injuries) (31). PFP can be aggravated by weight-bearing activities such as climbing stairs, squatting, jumping and running, which increase compressive forces acting on the patellofemoral joint (PFJ) (44). Elevated PFJ stress (PFJ_{stress}) is one of the most widely proposed mechanisms of PFP among runners (24). Considering the nature of repetitive, cyclic movements during running, any decrease in load magnitude borne by the knee joint during the stance phase may be significant in reducing the risk of overuse knee injuries. As a complex condition, PFP is linked with biomechanical changes local (tibiofemoral and patellofemoral), proximal (hip) and distal (ankle) to the knee joint (6).

Previous research has illustrated several modifications in posture and/or technique aimed at mitigating running related PFP. For instance, switching from rear- to mid- or fore-foot strike pattern induces a distal shift of mechanical loading from the knee to the ankle (11, 41). Increasing the running step rate diminishes the negative work performed at the hip and knee joints (14) or decreasing the step length can decrease PFJ_{stress} (42, 43). Similarly, postural modifications such as forward trunk-leaning appears effective in reducing PFJ_{stress} (10, 32, 33), since trunk orientation during walking (18) or running (1, 2, 32, 33) also significantly redistributes the mechanical demands of lower-limb joints. This is mainly due to changing the orientation (3) and position (28) of the ground reaction force vector relative to the lower-limb joints.

Forward trunk-leaning has already been shown to mitigate PFJ_{stress} during running on even (32, 33) or sloped (15) surfaces. During even running (32, 33), the reduction of PFJ_{stress} due to forward trunk-leaning occur without placing further mechanical stress on ankle plantar-flexors (32-34). This strategy differs to changing foot-strike pattern and step-rate in so much that the latter modifications shift biomechanical loading more distally (10, 19, 27), which might be associated with higher risks of ankle and foot injuries. The slope of running surface also influences PFJ_{stress}. Running on a decline treadmill elevates peak PFJ_{stress}, while incline running does not (15). Elevated joint stress during decline running is mainly attributed to less forward trunk-leaning. Considering outdoor running is performed on uneven ground, the PFJ kinetic profile may alter in response to surface height changes. This is mainly due to the fact that, compared to even running, the accommodation of downward steps during running is associated with more extended lower-limb joints and a more vertical leg orientation (1, 21), leading to an altered pattern of lower-limb joint kinetics.

One can argue that an altered trunk orientation conceivably influences PFJ lower-limb joint kinetics differently with changing surface height. Hence, assessing the role of trunk posture in reorganization of the lower-limb joint mechanics associated with PFJ_{stress} under perturbed locomotion may provide further insight into running-related patellofemoral pain and guide the development of preventive and/or conservative interventions effective to reduce PFJ load in the runners. In this framework, the aim of this study was primarily to determine the effects of sagittal plane trunk posture on PFJ kinetics during running across uneven ground surface, namely an expected 10 cm drop-step (DS10; Figure 1). We hypothesized that running with forward and backward trunk-leaning would decrease and increase PFJ kinetics (contact force and stress) compared to self-selected trunk posture, and the changes in PFJ kinetics would be greater during the DS10 compared to level, unperturbed step (LS).

METHODS

Experimental Approach to the Problem

We used a cross-sectional repeated measures study design to evaluate the interaction of changes to the running surface height (10 cm expected drop-step) and running posture (backward and forward trunk-leaning from a self-selected trunk posture) on kinetic behaviour of lower-limb joints, including hip, tibiofemoral (knee), patellofemoral and ankle. The drop-step was created by lowering the height-adjustable force plate by 10 cm embedded halfway along a 15 m long instrumented track.

Subjects

Twelve (six females) recreational runners (mean ± *SD*); age = 28.5 ± 5.7 years; body mass index = 22.4 ± 1.9 kg.m^{-2}; running distance = 15.6 ± 5.3 km.wk^{-1}), free of any current/previous history of lower-limb surgery/injury or low back pain for at least the previous 6 months voluntarily participated in the study. All participants self-reported that they had experience with running on uneven surfaces. A minimum sample size of eleven participants was determined from a priori power analysis using G*Power (Version 3.1, University of Dusseldorf, Germany) implementing an effect size of 0.33 and statistical power of 80% (α = 0.05). The experimental protocol was approved by the local Ethics Committee of the Friedrich Schiller University Jena (3532–08/12) and met all requirements for human experimentation according to the

Declaration of Helsinki. All participants were informed of the benefits and risks before signing the approved informed consent document.

Experimental design and procedures

Data were collected using a twelve-camera motion-capture system (250 Hz; MCU1000, Qualisys, Sweden) and two consecutive force plates [1000Hz; 9281B (0.4 × 0.6 m), 9287BA (0.6 × 0.9 m), Kistler, Switzerland] embedded halfway along a 15 m long instrumented track. The arrangement of the force plates allowed for step lengths ranging from 1.40 to 2.30 m. We synchronized kinematics and ground reaction force data using an external trigger and BioWare data acquisition software (Kistler Instrument AG, Switzerland). Applying joint coordinate standards of the International Society of Biomechanics (45), a twelve-body segment model was defined using nineteen reflective markers. The markers were placed on the following bony landmarks: fifth metatarsal heads, lateral malleolus, lateral epicondyles of femurs, greater trochanters, anterior superior iliac spines, L5–S1 junction (L5), lateral humeral epicondyles, wrists, acromioclavicular joints, seventh cervical spinous process (C7) and middle of the forehead. Trunk angle was defined by the angle sustained by the line connecting the L5 and C7 with respect to the vertical (1, 3) (Figure 1). Mean trunk angle was calculated as the average sagittal plane trunk posture during the stance phase of the level step. Following running with a self-selected trunk lean (TL_0), participants were instructed to run with anterior (TL^+) and posterior (TL^-) trunk leans within a range in which they felt comfortable when running across even or uneven tracks (1) (Figure 1). Following the completion of the running conditions on even, uniform track (level step; LS), the variable-height force plate at the site of the second contact (drop-step; DS10) was visibly lowered by 10 cm and participants ran along the uneven track. The order of the TL^+ and TL^- conditions was randomized for each participant while the order of the running tracks was fixed. Practice trials were permitted to allow participants to become familiar with the running velocity and with the desired trunk postures. Participants accomplished ten valid runs per condition in which they fully struck each force plate with a single foot in such a manner that the second force plate was always hit by the right (dominant) foot. The selected kinematic and kinetic variables were analyzed for the right limb only.

Data analysis and statistical analysis

For data analysis, we chose all trials completed at a speed of 3.5 $m \cdot s^{-1}$ (33) and discarded trials that differed by more than 5% in speed from step to step (calculated from mean horizontal velocity of the L5 marker for each of the two force plates). Kinetic and kinematic data of all successful trials were analyzed using custom written Matlab code (Mathworks Inc., MA, USA). The raw coordinate data were filtered using a fourth-order low-pass, zero-lag Butterworth filter with 12 Hz cutoff frequency (3). Sagittal plane knee joint kinematics (range of motion and peak flexion angle) were determined as the motion of the distal segment relative to the proximal reference. We calculated net lower-limb joint moments by inverse dynamics using the ground-reaction-force, the center of pressure, a rigid linked segment model, and anthropomorphic data (7). A vertical ground reaction force threshold of 3% body weight was used to determine the instants of foot-touchdown and toe-off at each contact (3). The knee joint stiffness was calculated from the ratio of the change in net muscle moment to joint angular displacement between foot-touchdown and the instant of peak maximal flexion. Net joint moments and knee joint stiffness were normalized to the participant's body mass.

Kinetic data related to the PFJ were computed from knee flexion angle (θ_k) and net extension moment (M_k) during the stance phase of running using a previously described biomechanical model (4, 10). The effective lever arm (L_Q) of the quadriceps muscle was calculated as a function of θ_k by use of a nonlinear equation (38):

$$L_Q = 0.00008(\theta_k)^3 - 0.013(\theta_k)^2 + 0.28(\theta_k) + 0.046 \quad (1)$$

The force of the quadriceps (F_Q) was calculated using the following formula:

$F_Q = M_k / L_Q$ (2)

The PFJ_{CF} was estimated as the product of the F_Q and a constant (C):

$PFJ_{CF} = F_Q \, . \, C$ (3)

The constant (C) was determined using a nonlinear equation (39):

$C\,(\theta_k) = (0.462 + 0.00147(\theta_k) - 0.0000384(\theta_k)^2) \,/\, (1 - 0.0162(\theta_k) + 0.000155(\theta_k)^2 - 0.000000698(\theta_k)^3)$ (4)

PFJ_{stress} (MPa) was estimated by dividing the PFJ_{CF} by the PFJ contact area (PFJ_{CA}). The PFJ_{CA} was determined as a function of θ_k (23):

$PFJ_{CA} = 0.00002(\theta_k)^4 - 0.0033(\theta_k)^3 + 0.1099(\theta_k)^2 + 3.5273(\theta_k) + 81.058$

Following the normality test by Shapiro-Wilk test, two-way repeated measurements ANOVAs with post hoc Bonferroni adjustments were used to examine the main and interaction effects of *Step* (LS and DS10) and *Posture* (TL^{-}, TL_0 and TL^{+}) on the primary outcome variables (PFJ_{CF} and PFJ_{stress}) and the secondary outcome variables, including peak flexion angle, range of motion, peak extension moment and stiffness at the knee and peak extension moments across the hip and ankle. Statistical significance level was set at $p < 0.05$ and the data were analyzed in SPSS software (ver 21.0, IBM® Co., USA). Results were expressed as mean ± *SD* over all participants and variables.

RESULTS

The analysed data included 720 step cycles (12 participants × 6 conditions × 10 trails). We conducted separate two-way repeated measurements ANOVA to compare variations in the mean trunk angles across two steps. No *Step*-by-*Posture* effects nor *Step* main effect were detected for the mean trunk angle

($F_{1.23,13.6}$ = 3.07, p = 0.09, d = 0.21; $F_{1,11}$ = 0.91, p = 0.36, d = 0.07, respectively); however, there was a significant *Posture* main effect ($F_{1.16,12.8}$ = 96.5, p < 0.001, d = 0.89). Post hoc comparisons revealed that the mean trunk angle was significantly higher and lower during the TL^{+} and TL^{-} conditions, when compared with the TL_0 (Figure 1).

Figure 2 presents PFJ_{stress}, PFJ_{CF} and knee kinematics during the stance phase of the running cycle in the three *Posture* conditions across the LS and DS10. No *Step*-by-*Posture* effects were detected for PFJ_{stress} ($F_{2,22}$ = 1.02, p = 0.37, d = 0.08) and PFJ_{CF} ($F_{2,22}$ = 1.11, p = 0.34, d = 0.09). There was a significant *Posture* main effect on PFJ_{stress} ($F_{1.35,14.8}$ = 43.9, p < 0.001, d = 0.81) and PFJ_{CF} (= 43.5, p < 0.001, d = 0.79). Post hoc comparisons revealed that PFJ_{stress} and PFJ_{CF} were significantly higher during TL^{-} running (p = 0.001 and p < 0.001, respectively), while significantly lower during TL^{+} running (p = 0.003 for both variables), compared with TL_0 running (Figure 3). No significant changes were observed for the PFJ_{CA} across running conditions ($F_{2,22}$ = 1.22, p = 0.31, d = 0.11). Significant *Step*-by-*Posture* effects were found for peak knee flexion angle ($F_{2,22}$ = 4.38, p = 0.02, d = 0.28) and stiffness ($F_{2,22}$ = 3.46, p = 0.04, d = 0.24). Post hoc comparison revealed a decreased peak knee flexion angle (p < 0.001) and increased joint stiffness (p = 0.005) in the DS10 versus the LS during TL^{-} running (Figure 4).

Figure 5 presents the lower-limb joint moments during the stance phase of the running cycle in the three *Posture* conditions across the LS and DS10. No *Step*-by-*Posture* effects were detected for hip ($F_{1.84,20.2}$ = 0.06, p = 0.92, d = 0.006), knee ($F_{2,22}$ = 1.78, p = 0.19, d = 0.14) and ankle ($F_{2,22}$ = 0.13, p = 0.87, d = 0.01) peak extension moments. There was a significant main effect of *Step* ($F_{1,11}$ = 13.7, p = 0.003, d = 0.55) and *Posture* ($F_{2,22}$ = 63.5, p < 0.001, d = 0.85) on hip extension moment. Post hoc comparisons revealed a higher hip extension moment in DS10 versus LS (p = 0.003); and significantly lower and higher moments during TL^{-} and TL^{+} running conditions, respectively, compared with TL_0 running (p < 0.001; Figure 3). There was a significant *Posture* main effect on knee extension moment ($F_{2,22}$ = 52.7, p < 0.001,

d = 0.82). Post hoc comparisons revealed significantly higher and lower values during TL^{-} and TL^{+} running conditions, respectively, compared with TL_0 running ($p < 0.001$; Figure 3). There was a significant *Step* main effect on peak ankle extension moment ($F_{1,11} = 58.8$, $p < 0.001$, $d = 0.86$). Post hoc comparisons revealed a significantly increased peak ankle extension moment in the DS10 ($p < 0.001$; Figure 3).

DISCUSSION

In this study, we primarily examined the influence of simultaneous changes to the sagittal trunk posture and surface height on PFJ kinetics (contact force and stress) during running. Our hypothesis that running with forward and backward trunk-leaning would decrease and increase PFJ kinetics, respectively, was confirmed. The kinetics of PFJ, along with that of local and proximal joints were influenced by altering the trunk-lean angle independent of the running surface change. Specifically, while the ankle moment remained relatively unchanged across the running postures, forward trunk-leaning resulted in lower peak PFJ kinetics and knee extension moment but a higher hip extension moment. The opposite effects occurred with backward trunk-leaning. Step-wise, the accommodation of the drop-step was associated with significantly higher knee range of motion and peak extension moments across the hip and ankle.

PFJ kinetics did not change in response to an expected 10 cm drop-step while adopting various running postures. This finding is not in accordance with the existing literature, which suggests a 37% increase in PFJ_{CF} during decline running (15). This dissimilarity in findings can be attributed to compensatory biomechanical adjustments local, proximal, and distal to the knee joint. Locally, the knee range of motion and peak extension moment under altered running postures were not influenced by surface height changes. However, the peak knee flexion angle decreased, and the joint stiffness increased when stepping down during backward trunk-leaning only. Moreover, no step-related changes in the peak extension moments proximal or distal to the knee joint were observed for altered running postures. The lower-limb kinematic and kinetic adjustments during uneven running is influenced by the expectedness (9) and the magnitude (21) of ground surface changes. The lack of interaction between changes to the running posture and surface height in our study may be due to the use of feed-forward control strategies facilitated by visual awareness of mechanical perturbation to runners' locomotion, leading to some anticipatory postural adjustments (8, 22, 25). Accommodating unexpected or larger downward steps is more likely to elicit altered, feedback based, coping strategies and thus a different contribution of the lower-limb joints to the PFJ's mechanics.

Stability of trunk in response to external perturbations such as ground surface changes is becoming more challenging, since it requires the motor control mechanism to govern the additional change within the system's energy (36). Due to a delayed onset of ground contact (22), downward steps are associated with increases in vertical kinetic energy. Restraining an increased center of mass's energy gained in the drop-steps results in a higher mechanical loading of the lower-limb tri-articulate musculoskeletal system. In the current study, the peak hip and ankle extension moments in the drop-step increased by ~35% and ~28% (*Step* main effect), respectively, while peak knee extension moment remained unchanged. This indicates that the step-down enforces high hip moments to control the forward momentum of the trunk. The increased ankle joint moment can be explained by the fact that the distal joints are the direct point of contact with the ground and are the first to negotiate mechanical changes in the ground surface. Therefore, ankle is highly load-sensitive, owing to higher, swift proprioceptive feedback gain and responsiveness to intrinsic mechanical effects (5). This presumably enables rapid, necessary adjustments for the stability of unsteady or perturbed running. Consequently, the accommodation of small, expected drops during running appears to increase the mechanical stress on proximal and distal joints to the PFJ, independent of trunk posture.

Modification of sagittal trunk lean during running was found to significantly influence the PFJ kinetics, regardless of surface height change. The peak PFJ_{stress} was decreased by 8.8% and increased by 23% when leaning the trunk ~15° forward and backward, respectively, from the self-selected posture. These findings accord with those of a previous study (33), despite the utilization of a narrower range of trunk-leaning angles. In comparison, Teng and Powers (33) reported a 6.0% decrease and a 7.4% increase in peak PFJ_{stress} by forward and backward trunk-leaning, respectively, during even running. Moreover, given the PFJ_{CA} remained substantially unchanged across the running postures in our study, the variations in the PFJ_{stress} can be attributed to the changes in the PFJ_{CF}. Compared with the self-selected condition, the PFJ_{CF} decreased by 9.8% and increased by 21% with forward and backward trunk-leaning, respectively. The same trend for the peak knee extension moment, namely a 12% decrease by forward trunk-leaning and a 16% increase by backward trunk-leaning. In addition to the previous within-session analyses of the association between the trunk posture and PFJ kinetics, Teng, et al. (32) recently reported ~17% reductions in peak PFJ_{stress} and knee extension moment following a four-week trunk modification program (i.e., 10° increase in trunk flexion angle) for running.

We also observed kinetic changes proximal to the PFJ. Compared to the self-selected running posture, the hip moment was systematically higher and lower throughout the stance phase of both analyzed steps when leaning the trunk forward and backward, respectively (Figure 5). The finding that a forward-leaning trunk during running shifts mechanical demand from knee extensors to hip extensors has been mirrored in the previous studies (2, 10, 32, 33). Given the hip's contribution to sagittal plane total work is small (less than 15%) during the stance phase of running (17), a distal-to-proximal load shift by a forward trunk lean during running can be an effective strategy to decrease PFJ_{stress}. Moreover, runners with a stronger hip extensor muscles tend to adopt a more flexed trunk posture during running (35). This is shown to be associated with a lower load on the knee extensors. However, the causal relations between the trunk posture, hip-extensor strength and lower-limb biomechanics during running have not yet been determined. Thus, in the absence of the musculature dysfunction of the lower-limb or lower back, forward trunk-leaning might be used to train the hip extensors without weights or other interventions (26). However, this reasoning needs further investigations.

Different modifications to the running pattern have been proposed to mitigate running-related PFP. A recent study reported a greater reduction in peak PFJ_{stress} when switching to a forefoot strike (~23%), or when increasing the step-rate (~12%), compared with forward trunk-leaning (~5%) during treadmill running (10). However, these modifications were associated with greater mechanical demands on the ankle and foot joints (10, 19, 27) without taking into consideration the impact of environmental demands. This is significant since most running is performed outdoors and is essentially challenged by uneven terrain. Rather, forward trunk-leaning during running enforces a distal-to-proximal shift in lower-limb mechanical demands (2, 32-34). Hence, one can argue that such postural modification may be a safer strategy for running-related PFP management.

When interpreting the findings of this study, several limitations need to be acknowledged. First, the runway utilized in the present study cannot resemble running outdoors, which represents a wide range of (un)expected and/or varied magnitudes of surface perturbations that may elicit altered PFJ kinetics. Second, given that individuals with low back pain or musculoskeletal problems were excluded from this study, the future studies should address the impact of an altered trunk posture in runners on lumbar moments in prospective and/or longitudinal experiments. Third, only healthy runners at a fixed running velocity were examined. Therefore, the generalization of the results to the symptomatic runners and/or to the conditions under different running velocities should be taken with caution. Fourth, our analysis was limited to the mechanical behavior of perturbed limb (right) during the stance phase. Therefore, future investigations that involve the contribution of the contralateral limb would provide a more detailed profile of the lower-limb joint mechanics under perturbation. Fifth, the computation of PFJ kinetic used in this study did not consider many relevant parameters, including kinematics and kinetics in the frontal and transverse planes, individual lever arm lengths and their changes due to the applied force on the tendon or the effect of agonist and antagonistic muscles force generation on the patella tendon forces. Sixth, we did not control the runners' foot strike pattern when accommodating the drop-step. Therefore, our findings do not exclude the possibility of foot strike effects on patellofemoral loading. For a more comprehensive understanding of PFJ kinetics during functional activities, future methods should account for above-mentioned parameters.

In conclusion, leaning the trunk by nearly ±15° potentially impacts the kinetic pattern of lower-limb joints. The peak patellofemoral joint contact force and stress are significantly influenced by changes to the running posture, through a distal-to-proximal joint load redistribution, but not by expected small changes to the running surface height. In line with the reported efficacy of forward trunk-leaning in mitigating PFJ stress while even (32, 33) or decline (15) running, this acute intervention appears effective during uneven running.

PRACTICAL APPLICATIONS

Given patellofemoral pain is the most frequently reported running-related symptom, increased knowledge of interventions aimed at reducing patellofemoral pain in runners is crucial for runners, coaches, and health professionals. Modification of sagittal trunk orientation has the potential to re-organize lower-limb joint mechanical. The findings of the present study suggest that leaning the trunk forward helps reducing the patellofemoral joint contact force and stress, irrespective of small changes to the running surface. Given most of running is performed outdoors, where changes in ground level surface are frequently encountered, adopting such running technique may help runners to reduce their PFJ loading. Secondly, running gait retraining protocols with a focus on using feed-forward control strategies (anticipatory postural adjustments) could improve compensatory mechanisms (e.g., a distal-to-proximal shift in lower-limb mechanical demands) for protecting the patellofemoral joint in response to running surface perturbation.

Finally, in absence of musculature dysfunction of the lower limb or lower back region, runners may consider a forward trunk-leaning technique as a supplementary training modality to strengthen the hip extensors without weights or other interventions.

ACKNOWLDGMENTS

The authors declare no conflicts of interest.

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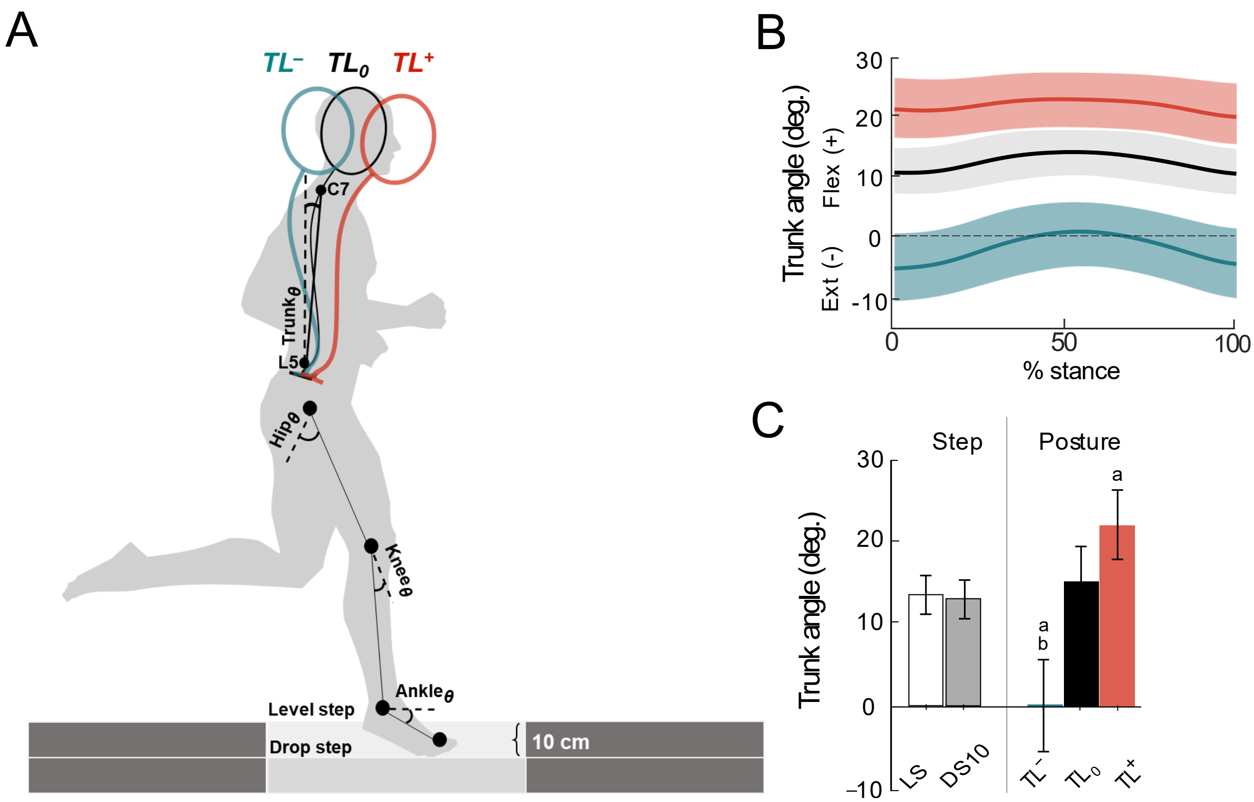
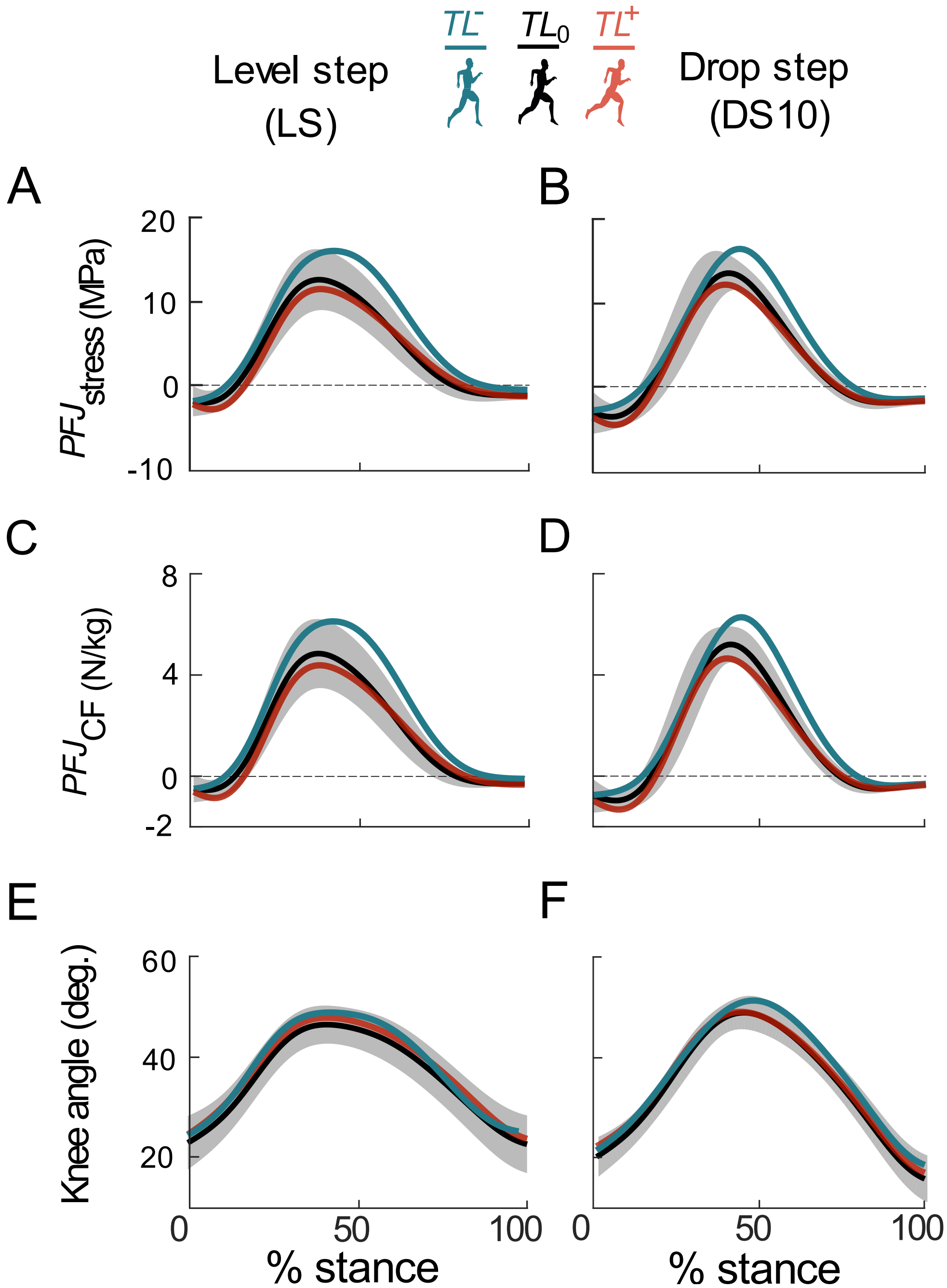
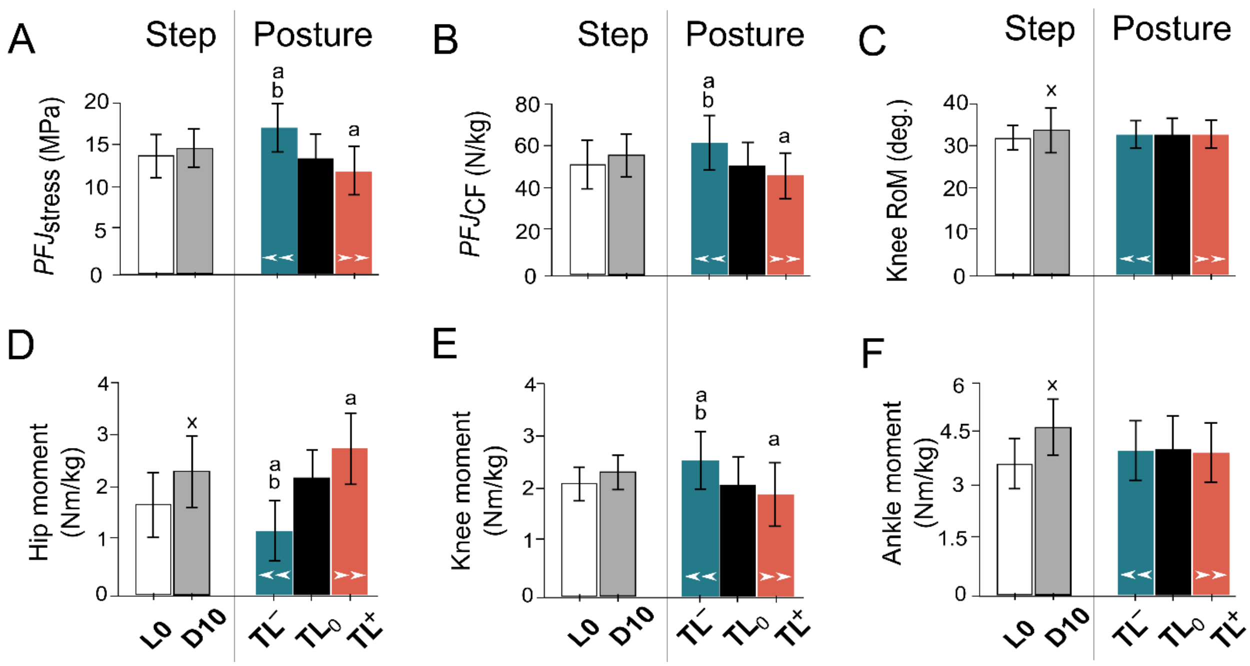
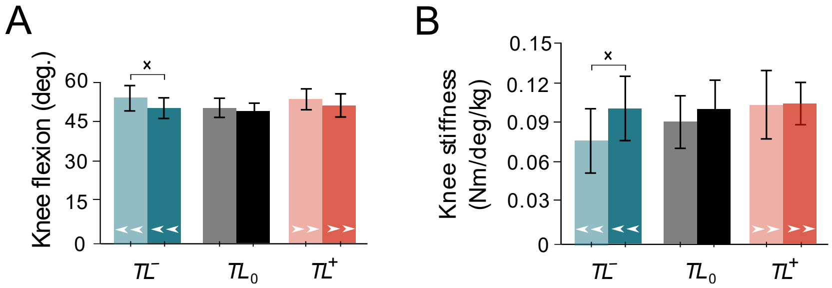
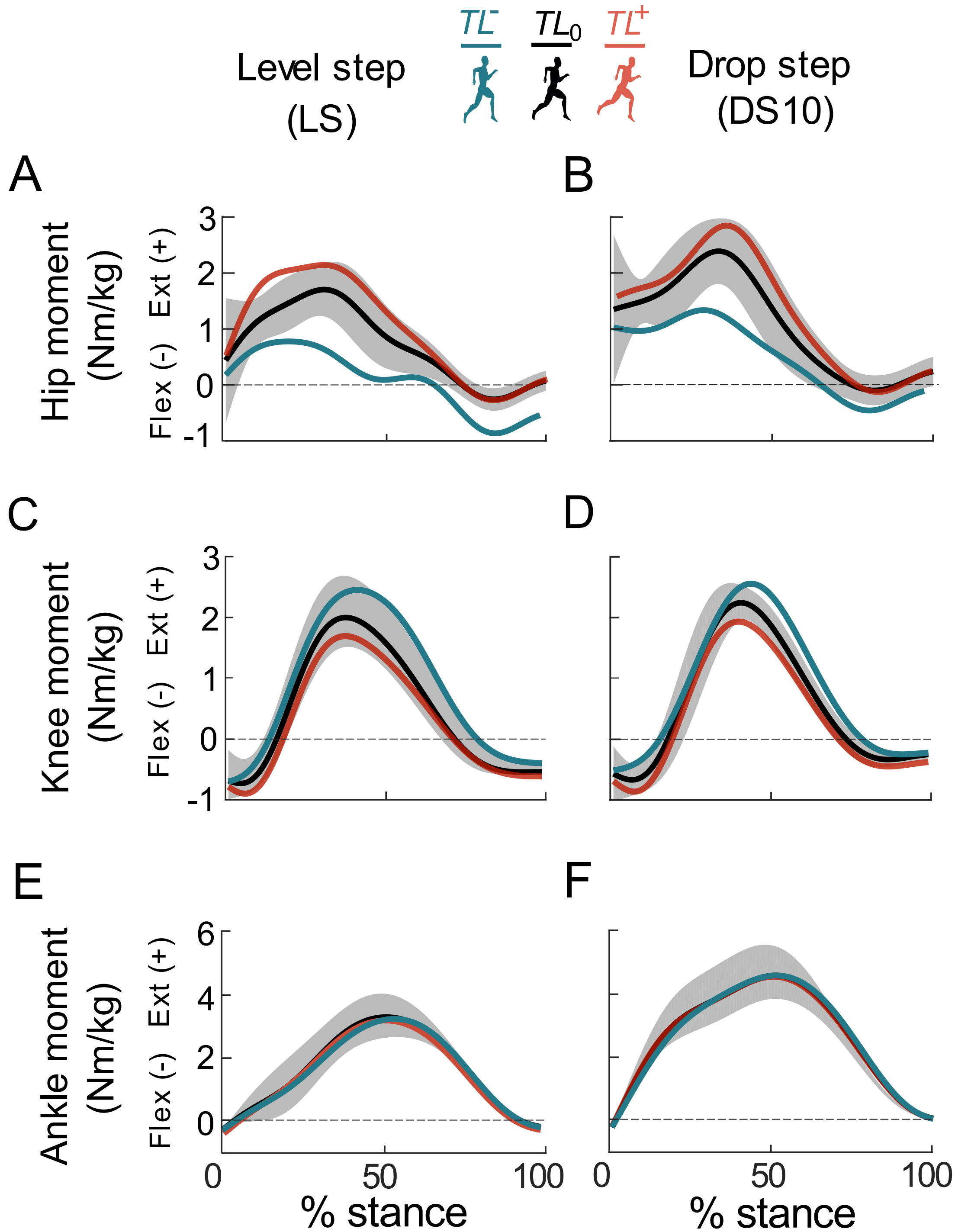




Level step (LS)
TL^-
TL_0
TL^+
Drop step (DS10)
A
B
C
D
E
F
PFJ_{stress} (MPa)
20
10
0
-10
PFJ_{CF} (N/kg)
8
4
0
-2
Knee angle (deg.)
60
40
20
0
50
100
% stance
0
50
100
% stance


A
Step
Posture
PFJstress (MPa)
B
PFJCF (N/kg)
C
Knee RoM (deg.)
D
Hip moment (Nm/kg)
E
Knee moment (Nm/kg)
F
Ankle moment (Nm/kg)
L0
D10
TL−
TL0
TL+


A
Knee flexion (deg.)
0
15
30
45
60
×
TL⁻
TL₀
TL⁺
B
Knee stiffness (Nm/deg/kg)
0
0.03
0.06
0.09
0.12
0.15
×
TL⁻
TL₀
TL⁺



TL⁻
TL0
TL⁺
Level step
(LS)
Drop step
(DS10)
A
B
C
D
E
F
Hip moment
(Nm/kg)
Knee moment
(Nm/kg)
Ankle moment
(Nm/kg)
Flex (-) Ext (+)
3
2
1
0
-1
6
4
2
0
0
50
100
% stance

401

Figure 1. Schematic illustration of uneven running and trunk kinematics. (A) The schematic representation of a runner with three different trunk-leaning conditions while running across the level step (LS) and drop-step (DS10: 10 cm drop of the second force plate). (B) The ensemble-averaged trunk kinematics across the level step stance phase (shaded area: ±1 *SD*). (C) Main effects of *Step* and *Posture* on the mean trunk kinematics (‘a’ and ‘b’: significant differences from TL_0 and TL^+; error bars: SD). Abbreviation: TL^-, backward-leaning trunk; TL_0, self-selected trunk-leaning; TL^+, forward-leaning trunk; θ, angle; L5, fifth lumbar spine vertebrae; C7, seventh cervical spinous process.

Figure 2. PFJ kinetics and knee range of motion**.** Shown are ensemble-averaged patellofemoral joint stress (PFJ_{stress}, top row), patellofemoral joint contact force (PFJ_{CF}, middle row) and knee joint angle (bottom row) across the level step (LS; left) and the drop-step (DS10; right) during the stance phase of three running conditions: backward-leaning trunk (TL^-, green), self-selected trunk-leaning (TL_0, black) and forward-leaning trunk (TL^+, red). The shaded area represents ±1 *SD* for the self-selected condition.

Figure 3. Main effects of *Step* and *Posture*. Shown are the main effects (mean ± *SD*) on variables that two-way repeated measures ANOVAs did not reveal *Step*-by-*Posture* interaction. Significant bilateral differences (LS and DS10) are indicated by ‘×’. Accordingly, significant differences from TL_0 and TL^+ are indicated by ‘a’ and ‘b’, respectively ($p < 0.05$). Error bars denote *SD*. Abbreviation: TL^-, backward-leaning trunk; TL_0, self-selected trunk-leaning; TL^+, forward-leaning trunk; *PFJ*, patellofemoral joint; *CF*, contact force; RoM, range of motion; LS, level step; DS10, drop-step.

Figure 4. *Step*-by-*Posture* interaction. (A) Peak knee flexion angle (B) knee joint stiffness. The bar graphs in light colours represent level step (LS), while those in dark colours represent drop-step (DS10). ‘×’ indicates a significant difference between LS and DS10 ($p < 0.05$). Error bars denote *SD*. Abbreviation: TL^-, backward-leaning trunk; TL_0, self-selected trunk-leaning; TL^+, forward-leaning trunk.

Figure 5. Lower-limb joint moments. Shown are ensemble-averaged hip (top row), knee (middle row) and ankle (bottom row) moments across the level step (LS; left) and the drop-step (DS10; right) during the stance phase of three running conditions: backward-leaning trunk (TL^{-}, green), self-selected trunk-leaning (TL_0, black) and forward-leaning trunk (TL^{+}, red). The shaded area represents ±1 *SD* for the self-selected condition.