THE TRADE-OFF BETWEEN CUMULATIVE JOINT LOADING AND COST OF TRANSPORT WHEN ALTERING ANTERIOR TRUNK LEAN ANGLE IN RUNNING

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The study aimed to establish dose-response relationships between systematically altered anterior trunk leaning (ATL) and lower extremity cumulative joint loading (angular impulse x number of strides) as well as cost of transport (COT) in distance running. Twenty-eight recreational runners underwent a series of six treadmill runs (2.5 m/s) with five predefined ATL conditions (from -4° extension to 28° flexion) and one self-selected ATL condition for five minutes with 3D motion capture and spirometry. Increasing ATL systematically decreased cumulative knee joint loading and increased cumulative hip joint loading in all conditions. However, running outside the preferred running style increased COT. Designing ATL-based overuse load management interventions shows promise, but clinical implementation requires careful consideration of the COT and joint loading trade-offs.

KEYWORDS: trunk leaning, cumulative loading, load management, running injury, locomotion.

INTRODUCTION: Distance running is associated with a high risk of running-related overuse injuries (ROIs), particularly at the knee (van Gent et al., 2007). In light of the notable occurrence of ROIs, research has considered altering a person's running pattern (e.g.) - to redistribute lower extremities joint loading (Dos Santos et al., 2016). Altering the anterior/posterior trunk leaning (ATL/PTL) during running may be a particularly interesting variable to modulate lower extremity joint loading, as the trunk has the largest segmental mass (de Leva, 1996). There is some evidence that running with ATL leads to a redistribution of joint loading with decreasing knee joint loading and increasing hip joint loading (Teng & Powers, 2015; Warrener et al., 2021), but a potential trade-off with cost of transport (COT) is currently uncertain.
ROIs seem to result from accumulated tissue damage over time, determined by the loading magnitude and the number of loading cycles (Hreljac et al., 2000). However, biomechanical parameters usually consider only a single stance phase of running and may be insufficient to fully explain the accumulation of tissue damage and the appearance of ROIs over time. Therefore, this study also includes spatiotemporal parameters to provide an adequate surrogate for cumulative joint loading (cJL) that considers both the magnitude and frequency of loading. This study aimed to establish dose-response relationships and reveal potential trade-offs between systematically modified ATL, lower extremity cJL, and COT in distance running to guide overuse load management in clinical practice.

METHODS: Twenty-eight recreational runners (14 males and 14 females, age 30.5/12.3 years, height 1.75/0.14 m, weight 69.0/15.8 kg, and average running distance 25.0/21.3 km/week; values as median/interquartile range) provided written informed consent and participated in the study. All participants were free of cardiovascular disease and ROIs for at least six months before data collection and were required to maintain a weekly running volume of ≥ 7.5 km. The local Ethics Board had approved the procedures of the study. Each participant went through a five-minute warm-up on the treadmill at a running speed of 2 m/s. Subsequently, participants completed a series of six treadmill runs (2.5 m/s) of five minutes each. These runs included five predefined (ATL28°, ATL20°, ATL12°, ATL4°, PTL4°, Fig. 1 A-E) and one self-selected ATL condition. Auditory feedback was provided instantaneously by the instructor with precise instructions on how much to lean forward or backward when participants deviated more than 5° from the predetermined trunk lean condition for three consecutive strides.

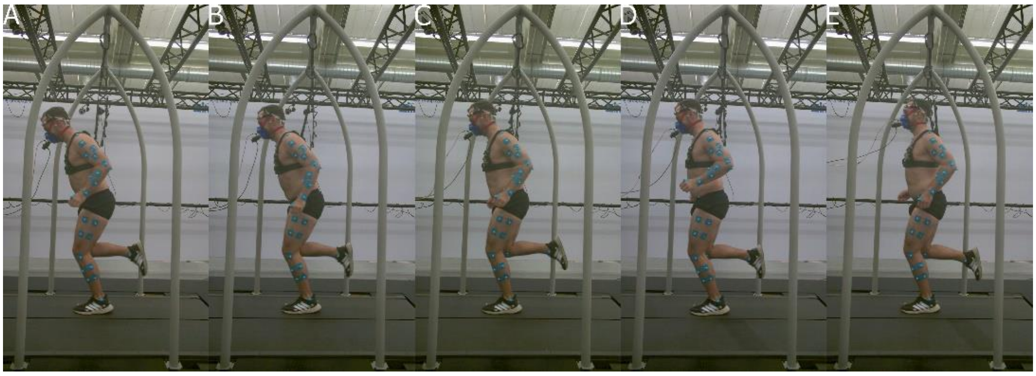

Figure 1: Experimental setup with the five predefined sagittal plane trunk lean conditions: (A) ATL28°, (B) ATL20°, (C) ATL12°, (D), ATL4°, (E) PTL4°.

The last minute of each condition was recorded for data analysis to allow participants to familiarize themselves with the trunk lean condition. A marker-based motion capture system (200 Hz, 24 Miqus M3 cameras, Qualisys AB, Gothenburg, Sweden) captured the position of 78 retroreflective markers using a full-body marker set (Willwacher et al., 2016). A 3D force-instrumented treadmill (2000 Hz, Bertec Corporation, Columbus, OH, USA) collected ground reaction forces (GRFs). 3D marker data were live-streamed to Visual3D (C-Motion, Inc., Germantown, MD; USA) using a custom-made model to provide real-time feedback on ATL angle to the participants. Rates of oxygen uptake (VO_2) and carbon dioxide production (VCO_2) under submaximal conditions were measured using a spirometry system (Vyntus CPX, Vyaire, Höchberg, Germany). Marker trajectories and GRFs were smoothed using a recursive, fourth-order digital Butterworth filter with a cut-off frequency of 20 Hz (Mai & Willwacher, 2019). A three-dimensional inverse dynamics model consisting of five rigid body segments (pelvis, thigh, shank, rearfoot, forefoot) was used to calculate joint kinematics and kinetics (Willwacher et al., 2016). The trunk angle was calculated as the orientation of the rigid trunk segment relative to the vertical axis of the global coordinate system. The trunk segment was tracked by anatomical landmarks, specifically the 7th cervical vertebrae (C7), 10th thoracic vertebrae (Th10), sternum, and clavicula. Joint moments were expressed as internal resultant moments and normalized to the participant's body mass. Cumulative loading (impulse) per joint was calculated by multiplying the time integral of the dominant direction of the joint moment curve (stance phase angular impulse; e.g., extension angular impulse at the knee) with the number of strides required to complete a 1000-m distance. VO_2 and VCO_2 were normalized to body weight and were used to calculate the mass-specific total metabolic rate (Péronnet & Massicotte, 1991). The standing (rest) metabolic rate was subtracted to obtain the net metabolic rate. Finally, the net metabolic rate was divided by the running speed to determine the net COT.
Statistical analyses were conducted in R Studio (RStudio PBC, Boston, Massachusetts) at a level of significance of 0.05. One-way repeated-measures ANOVAs (rmANOVAs) were used to identify main effects of ATL for each parameter. In cases where Mauchly's test of sphericity yielded significance, the Greenhouse-Geisser correction was applied. Partial eta squared (η_p^2) indicated effect size for the main effects and were interpreted as small (0.01 - 0.059), medium (0.06 - 0.139), and large (> 0.139) (Cohen, 1988). Bonferroni-corrected post-hoc tests were conducted to identify significant differences between factor levels if a main effect was detected.

RESULTS: The modification in ATL systematically altered cumulative knee joint loading (cKJL) ($p < .001$, $\eta_p^2 = 0.69$) and cumulative hip joint loading (cHJL) ($p < .001$, $\eta_p^2 = 0.96$) in all conditions. Linear regression revealed that each degree of ATL increased cHJL by 3.23 Nm·s/kg/1000 m (Fig.2A) and reduced cKJL by 1.12 Nm·s/kg/1000 m (Fig.2B). No main effect of ATL on cumulative ankle joint loading (cAJL) was found (Table 1). Consequently, the linear fitting of cAJL indicates almost no relationship between ATL and cAJL (Fig.2C).

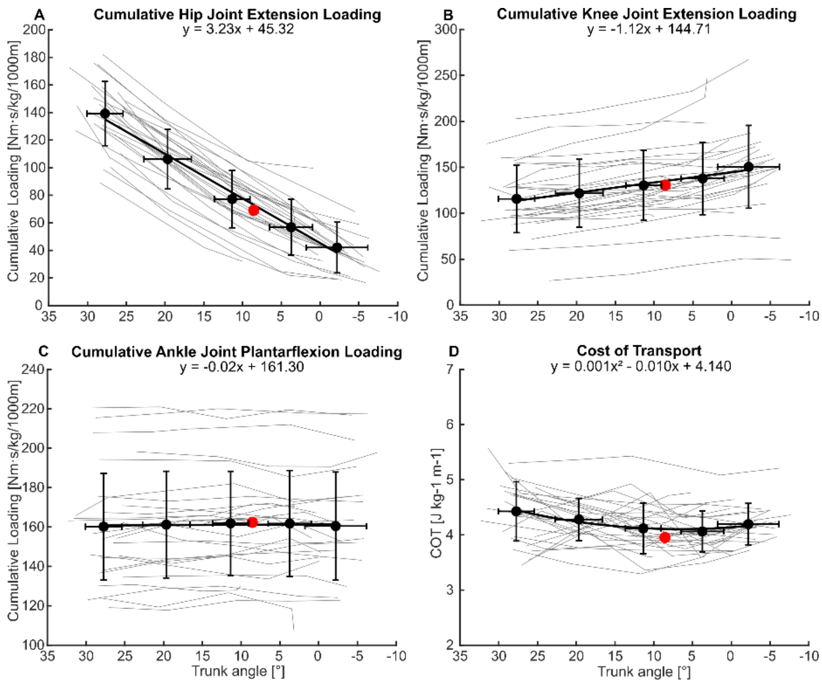

Figure 2: Means ± standard deviation of the cumulative loading for the hip, knee, and ankle joints (A, B, C) and net cost of transport (D). Horizontal error bars highlight the mean of the trunk angle condition ± standard deviation. The red dot is the self-selected ATL condition. Cumulative knee joint extension loading values were negative when described in the proximal coordinate system but are presented here in absolute terms for easier comparison.

The spirometric findings revealed that running outside the preferred ATL increases COT. E.g., ATL28° is characterized by an increase of 12.15% ($p_{bonferroni}$ < .001) in COT compared to the self-selected ATL condition (Fig.2D).

Table 1: Results of lower extremity kinetics and COT (mean ± SD)

	ATL28°	ATL20°	ATL12°	ATL4°	PTL2°	p
Trunk angle, °	27.7 ± 2.3 $^{20°, 12°, 4°, -2°}$	19.7 ± 3.1 $^{12°, 4°, -2°}$	11.6 ± 2.5 $^{4°, -2°}$	3.7 ± 2.8 $^{-2°}$	-2.2 ± 4.0	**< .001$_a$**
Stance phase angular impulse, Nm·s/kg						
Hip	0.25 ± 0.04 $^{20°, 12°, 4°, -2°}$	0.19 ± 0.04 $^{12°, 4°, -2°}$	0.14 ± 0.04 $^{4°, -2°}$	0.10 ± 0.04 $^{-2°}$	0.08 ± 0.03	**< .001$_a$**
Knee	-0.21 ± 0.07 $^{12°, 4°, -2°}$	-0.22 ± 0.07 $^{12°, 4°, -2°}$	-0.24 ± 0.07 $^{-2°}$	-0.25 ± 0.08 $^{-2°}$	-0.27 ± 0.08	**< .001$_a$**
Ankle	0.29 ± 0.05	0.30 ± 0.05 $^{-2°}$	0.30 ± 0.05 $^{-2°}$	0.29 ± 0.05	0.29 ± 0.05	**< .001$_a$**
Cumulative impulse, Nm·s/kg/1000 m						
Hip	139.22 ± 23.24 $^{20°, 12°, 4°, -2°}$	106.21 ± 21.70 $^{12°, 4°, -2°}$	77.18 ± 20.78 $^{4°, -2°}$	56.79 ± 20.20 $^{-2°}$	42.08 ± 18.42	**< .001$_a$**
Knee	-115.63 ± 36.85 $^{12°, 4°, -2°}$	-121.69 ± 37.02 $^{12°, 4°, -2°}$	-130.35 ± 38.12 $^{4°,-2°}$	-137.84 ± 39.37 $^{-2°}$	-150.49 ± 45.09	**< .001$_a$**
Ankle	160.18 ± 27.13	161.20 ± 27.08	161.84 ± 26.42	161.75 ± 26.96	160.45 ± 27.45	0.31
Cost of transport, J kg^{-1} m^{-1}	4.43 ± 0.54 $^{12°, 4°, -2°}$	4.28 ± 0.38 $^{4°}$	4.12 ± 0.46	4.06 ± 0.37	4.20 ± 0.38	**< .001$_a$**

Note: Bold parameter names indicate a significant ANOVA effect (p <.05). $^{28°, 20°, 12°, 4°, -2°}$ revealing significant (p < 0.05) bonferroni-corrected post-hoc tests. $_a$Mauchly's test of sphericity indicates that the assumption of sphericity is violated (p <.05) and the value is adjusted with Greenhouse-Geisser correction.

DISCUSSION: When altering from PTL2° to ATL28°, the cJL characteristics systematically increased at the hip joint and, conversely, decreased at the knee joint. These findings are consistent with previous studies indicating a redistribution of joint loading from distal to more proximal joints (Teng & Powers, 2015; Warrener et al., 2021). The dose-response relationships between ATL and lower extremity cJL are relevant in guiding clinical practice to enable more precise overuse load management strategies for recreational runners. However, the lack of relationship between cAJL and ATL suggests that altering ATL angle is only effective in modulating cKJL and cHJL, but not cAJL at the relatively low running speed chosen in this study. Based on a patient's overuse-injury situation and its self-selected ATL, clinical practitioners could suggest suitable lower extremity load management strategies incorporating ATL. Furthermore, we found that deviating from self-selected ATL increased COT, suggesting

that humans self-select the ATL orientation that minimizes COT in distance running. Consequently, when implementing load-management interventions based on ATL in clinical practice, the trade-off between lower extremity cJL and COT must be considered because COT may limit how long individuals can maintain an artificial running style based on ATL. However, designing interventions based on ATL seems promising because of the potential real-world quantification, e.g., implementing inertial measurement unit based feedback.
The results of this study must be interpreted in the light of some limitations. The calculation of cumulative impulse assumes an equal contribution of angular impulse and loading cycles to cumulative loading (Petersen et al., 2015). This approach is insufficient because the relationship between stress magnitude and loading cycles in biological tissues cannot be accurately described with a linear relationship. Future research should address this limitation by using musculoskeletal modeling and incorporating specific biological tissue characteristics. Furthermore, by reporting only changes within the sagittal plane, our study cannot elucidate the effects in the frontal and transversal planes, which is not negligible as potential risk factors for ROIs have been frequently identified outside the sagittal plane (Willwacher et al., 2022).

CONCLUSION: ATL can redistribute joint load from distal (cKJL) to proximal (cHJL), at the cost of cHJL and COT during distance running. Modulating lower extremity cJL by altering trunk lean angle is an effective strategy to redistribute cJL between/ within the knee and hip joint. When implementing ATL in clinical practice, the increased demands on the hip musculature and trade-off with running economy should be considered.

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