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# Does high-intensity running to fatigue influence lower limb injury risk?

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ABSTRACT

*Objectives:* The aim of this study was to quantify changes in peak bending moments at the distal tibia, peak patellofemoral joint contact forces and peak Achilles tendon forces during a high-intensity run to fatigue at middle-distance speed.
*Design:* Observational study.
*Methods:* 16 high-level runners (7 female) ran on a treadmill at the final speed achieved during a preceding maximum oxygen uptake test until failure (~3 min). Three-dimensional kinetics and kinematics were used to derive and compare tibial bending moments, patellofemoral joint contact forces and Achilles tendon forces at the start, 33 %, 67 % and the end of the run.
*Results:* Average running speed was 5.7 (0.4) $m \cdot s^{-1}$. There was a decrease in peak tibial bending moments (−6.8 %, $p = 0.004$) from the start to the end of the run, driven by a decrease in peak bending moments due to muscular forces (−6.5 %, $p = 0.001$), whilst there was no difference in peak bending moments due to joint reaction forces. There was an increase in peak patellofemoral joint forces (+8.9 %, $p = 0.026$) from the start to the end of the run, but a decrease in peak Achilles tendon forces (−9.1 %, $p < 0.001$).
*Conclusions:* Running at a fixed, high-intensity speed to failure led to reduced tibial bending moments and Achilles tendon forces, and increased patellofemoral joint forces. Thus, the altered neuromechanics of high-intensity running to fatigue may increase patellofemoral joint injury risk, but may not be a mechanism for tibial or Achilles tendon overuse injury development.


## Practical implications

- Patellofemoral joint contact forces increase during a high-intensity run to failure.
- Mechanical loading of the tibia and Achilles tendon decreases during a high-intensity run to failure.
- Altered neuromechanics as a result of high-intensity running is unlikely to be a mechanism for injury of the tibia or Achilles tendon in healthy runners.

## 1. Introduction

Running is associated with a high risk of overuse injury. Overuse injuries of the tibia, patellofemoral joint and Achilles tendon are three of the most common and burdensome injuries amongst runners.[1] In the case of both bone and tendon, the simplified mechanism for overuse injury development is understood to be an accumulation of microdamage which outpaces the remodelling of the tissue.[2,3] The magnitude of loading is more important than the quantity and duration of loading cycles in terms of failure of bone[4] and tendon[5] tissues.

Middle-distance runners are at high risk of injury, and the lower leg is reportedly the most common site of injury amongst both male and female middle-distance runners.[6] During a demanding run to fatigue, the loading of the knee, tibia and Achilles tendon may change as a result of altered neuromechanics, including muscular force production.[7,8] In order to better understand and mitigate against overuse injury

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development, quantification of the changes in structural loading throughout the duration of a run is required. However, alterations in lower limb loading during running exercise have not been well-established, particularly during high-intensity middle distance runs (e.g. 800 and 1500 m), that are inherently fatiguing.

The influence of running activity on tibial loading has previously been investigated, yet the findings are inconsistent. For example, a 10 km treadmill run at 105 % of season's best time resulted in a 5 % reduction in peak tibial loading in male runners.[9] This may have been influenced by the concomitant reduction in mechanical work done at the ankle following the 10 km run.[10] A similar finding was also observed after a longer run (~19 km,[11]). Another study observed decreased peak tibial strain after almost 2 h of running.[12] However, direct in vivo strain gauge measurement showed increased tibial strain after just 2 km of running at a self-selected pace,[13] which was presumably faster than in the studies of longer distance runs. The conflicting findings may suggest that loading is influenced by the duration and/or speed/intensity of the run.

Patellofemoral joint contact forces were unchanged in both male and female runners as a result of a short run (~12 min) at 3.5 $m \cdot s^{-1}$.[14] However, it has been suggested that knee joint kinematics and kinetics change during a run. For example, positive mechanical work done at the knee increased during a 10 km run,[15] and knee flexion angle increased during an exhausting high-intensity run.[16] An increase in mechanical work would likely increase the musculotendinous forces, thereby increasing knee joint contact forces.

Knowledge regarding changes in Achilles tendon loading throughout a run is also limited. Farris et al.[17] reported that Achilles tendon strain remained constant throughout a 30-minute run at a recreational pace, whilst Fletcher and MacIntosh[18] observed reduced Achilles tendon stiffness following a 90-minute run, assessed via ultrasound during dynamometry. It is unclear whether this reduced stiffness was a result of changes to the loading of the tendon that may have occurred during the run. The reduced mechanical work previously reported at the ankle joint following a long run (10–19 km)[11,15] may reduce the Achilles tendon forces but it is unclear whether this would occur after a shorter, faster run.

Overall, there is a lack of understanding of how lower limb loading changes throughout a high-intensity middle-distance run. The aim of this study was to quantify the changes in loading at the distal tibia, the Achilles tendon and the patellofemoral joint during an exhausting high-intensity run to fatigue at middle-distance speed. It was hypothesised that there would be an increase in tibial and patellofemoral loading and no change in Achilles tendon loading as the run progressed, based on previous studies at moderate-high running speeds/intensities.

## 2. Methods

### 2.1. Participants

Sixteen participants (9 males, 7 females, mean (SD) height: 1.75 (0.09) m; mass: 61.7 (7.1) kg; 23.8 (4.4) years) participated in this study, which was part of a larger project on middle-distance running biomechanics.[8] Participants were injury-free, high-level runners with season's best 800–1500 m times, equivalent to 1500 m times of 3:56 $\pm$ 0:08 min:s (males) and 4:33 $\pm$ 0:13 min:s (females). All participants provided informed consent and the study was approved by the Loughborough University Ethics Approvals (Human Participants) Sub-Committee.

### 2.2. Experimental design

The protocol has previously been reported in detail.[8] In brief, during a familiarisation session, a $VO_{2max}$ test was conducted to determine the prescribed running speed for the treadmill run, which was the final speed achieved during the $VO_{2max}$ test. In a second visit, data were collected during running on a force-instrumented treadmill (3DI, Treadmetrix, Utah, US) at a 1 % gradient.[19] This test was conducted after a warmup consisting of a 10-minute run at 60 % of the test run speed, a 15 s practice run at the test speed, and 5 min stretching. During the test run, the speed was set prior to the participant mounting the treadmill. During the run, the participant was encouraged to stay on the treadmill for as long as possible until they could no longer maintain the predetermined speed. Synchronised kinetics (1000 Hz) and whole-body kinematics (250 Hz) were captured throughout. Three-dimensional kinetics were captured using four force transducers (MC3A, AMTI, Watertown, US) embedded in the treadmill. Kinematics were captured using twelve Vicon (Oxford) cameras positioned around the treadmill. Forty-seven retroreflective markers and clusters were attached to the participant by a single assessor. A static trial was collected whilst participants stood on the level treadmill. Additional metrics were obtained[8] that are not relevant to the outcome measurements here.

### 2.3. Data analysis

Data were analysed from the right leg and averaged over 10 steps at each of four time points during the run: the start (strides 6–15); approximately 33 % and 67 % of the run; and at the end (strides 15–6 from the end). The first and last 5 strides were excluded to avoid interference resulting from mounting and dismounting the treadmill. Kinematic and kinetic data were low-pass filtered with a 2nd order Butterworth filter with a cutoff frequency of 12 Hz based on residual analysis of marker positions. Joint moments were calculated by inverse dynamics in Visual3D (C-Motion, Germantown, US) software, where stance was defined as the vertical ground reaction force exceeding 40 N.

Tibial bending moments about the medial-lateral axis were calculated at the distal 1/3rd of the tibia, as previously reported.[20] Muscular forces were estimated from eleven muscles[21,22] using static optimisation constrained to the sagittal plane joint moments, with a cost function minimising the sum of cubed muscle stresses. Resultant bending moments were the sum of the moments due to muscular forces and joint reaction forces.

Patellofemoral joint contact forces were estimated based on methods outlined by Starbuck et al.[23] Briefly, quadriceps muscle forces were determined as the sum of the hamstring and gastrocnemius forces multiplied by their respective moment arms at the knee joint and the knee joint moment[24,25] divided by the quadriceps effective lever arm calculated as a function of knee joint angle.[26] Gastrocnemius force was estimated as the proportion of Achilles tendon force attributed to the gastrocnemius[25] based on the cross-sectional area (CSA) relative to the soleus.[27] Hamstring force was calculated as the proportion of the hip joint moment generated by the hamstrings,[25] considering the CSA of the hamstrings relative to the combined CSA of both the hamstrings and the gluteus maximus,[27] as well as the muscle moment arms in relation to the hip joint angle.[28] Patellofemoral joint contact forces were estimated as the product of the quadriceps muscle forces and a constant *k*, which determined the relationship between the quadriceps muscle forces and patellofemoral joint contact forces.[29]

Achilles tendon force was estimated by dividing the sagittal ankle joint moment by the Achilles tendon moment arm. Achilles tendon moment arms were computed from non-normalised ankle joint angles based on previous regression equations.[30]

Tibial bending moments, patellofemoral forces and Achilles tendon forces were calculated in MATLAB (MathWorks, Natick, MA, USA). The peak tibial bending moment, peak patellofemoral joint contact force and peak Achilles tendon force were then extracted from 10 steps per person at each time point and averaged. The values throughout each analysed stance phase were also averaged and time-normalised for visualisation. Additionally, the components used to derive resultant bending moments at the distal 1/3 tibia (bending due to muscular forces and due to joint reaction forces) were analysed to further aid understanding.

### 2.4. Statistical analysis

A repeated measures ANOVA was conducted to assess the influence of run duration on the key outcome variables. All statistical analyses were conducted in Python Anaconda software (Version 3) and were considered to be significant if $p < 0.05$. Greenhouse–Geisser corrected p-values were considered where the assumption of sphericity was violated, which was the case with all variables in the present study. It had previously been confirmed that there were no interaction effects between time and sex for all variables, thus data from male and female participants were combined. Partial eta squared ($\eta_p^2$) effect sizes were reported where there were significant main effects, interpreted as: 0.01 < small ≤ 0.06; 0.06 < medium ≤ 0.14; and 0.14 < large.[31] Where there were significant main effects for time, post-hoc two-tailed paired *t*-tests with Bonferroni correction were conducted to compare 33 %, 67 % and the end of the run with the start of the run. Results are presented as mean (standard deviation). In addition, the reliability of the key outcome measures was assessed using average within-participant between-stride coefficient of variation (CV) calculated for 10 successive strides at the start of the run.

## 3. Results

Average running speed was $5.7 \pm 0.4\ \mathrm{m \cdot s^{-1}}$ ($20.6\ \mathrm{km \cdot h^{-1}}$) and average duration was $184.9 \pm 42.9$ s. The time points for data collection that were intended to occur at 33 % and 67 % of the run occurred at $35.3 \pm 3.3$ % and $66.0 \pm 2.0$ %. Within-participant CV values were 3.4 %, 2.3 % and 7.0 % for peak tibial, Achilles tendon and patellofemoral joint loading variables respectively.

### 3.1. Tibial bending moments

There was a main effect of time ($F_{(3,45)} = 11.642$, $p < 0.001$, $\eta_p^2 = 0.437$) on peak tibial bending moment (Fig. 1). Post-hoc tests identified a difference between Start–67 % ($p = 0.013$) and Start–End ($p = 0.004$). 14/16 (88 %) participants had reduced peak tibial bending moments from the start to the end of the run, where the mean reduction was 6.8 % across the group.

There was a main effect of time on peak bending due to muscular forces ($F_{(3,45)} = 18.266$, $p < 0.001$, $\eta_p^2 = 0.549$) which decreased throughout the run (Fig. 3). Post-hoc tests revealed differences between Start–67 % ($p = 0.002$) and Start–End ($p = 0.001$). The peak bending due to muscular forces was reduced by 6.5 % on average across the group between the Start and End of the run. There was no main effect of time on the peak bending of the tibia due to joint reaction forces ($F_{(3,45)} = 3.226$, $p = 0.07$, Fig. 3).

### 3.2. Achilles tendon force

There was a main effect of time on Peak Achilles tendon force ($F_{(3,45)} = 44.559$, $p < 0.001$, $\eta_p^2 = 0.748$, Fig. 4). Paired *t*-tests revealed a significant difference between all time point comparisons (all $p < 0.001$). There was a mean reduction of 9.1 % from the start to the end of the run.

### 3.3. Patellofemoral joint force

There was a main effect of time on patellofemoral joint force ($F_{(3,45)} = 6.730$, $p = 0.008$, $\eta_p^2 = 0.310$), showing an increase throughout the duration of the run. Post-hoc tests revealed differences between Start–67 % ($p = 0.034$) and Start–End ($p = 0.026$). There was a mean increase of 8.9 % from the start to the end of the run.

## 4. Discussion

This study demonstrated that changes to lower limb peak loading occurred throughout a high-intensity middle-distance treadmill run, with decreased tibial bending moments and Achilles tendon forces, whilst patellofemoral joint contact forces increased. This increased peak loading was observed to occur at the Achilles tendon and tibia after approximately 1 and 2 min of high-intensity running, respectively. The average reductions in peak tibial and Achilles tendon loading (6.8 % and 9.1 %, respectively) were greater than the average respective CVs (3.4 % and 2.3 %). The 8.9 % increase in peak patellofemoral joint forces between the start and end of the run was also greater than the average CV (7.0 %) for this variable. However, these within-participant changes are not large relative to the between-participant standard deviations of the whole group, as can be observed in Figs. 1, 3 and 4.

The finding of a 6.8 % reduction in bending moments observed at the tibia aligns with previous findings, where a 5 % reduction in peak tibial bending moments was observed throughout a 10 km run[9] and a 6.4 % reduction in 95th percentile von Mises equivalent strain was observed after 2 h of running.[12] This suggests that fatiguing running results in reduced tibial loading, that appears to be independent of the running speed, distance and duration. The reduced tibial bending moments in the present study were predominantly the result of the reduction in bending due to muscular forces - the greatest contributor to distal tibial bending (Fig. 2) - as the contribution from the joint reaction forces was unchanged throughout the run. Stance phase plantarflexor kinetics was previously reported from the same dataset to decline during this run (peak moments −9.0 %, positive work −13.9 %[8]), indicative of a net change in the muscular forces that act around the ankle joint. As EMG

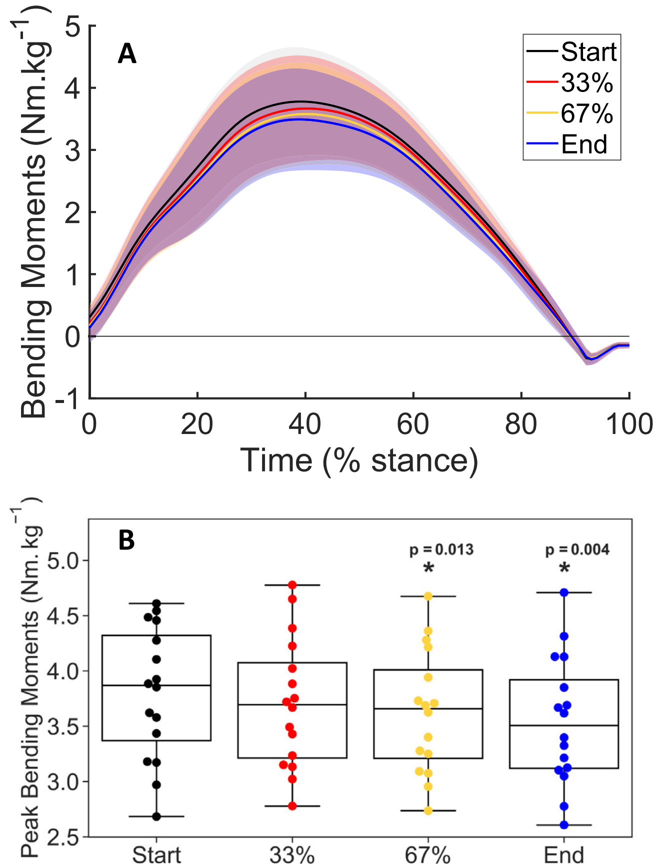


**Fig. 1.** A: Time-normalised bending moments at the distal 1/3 tibia about the medial-lateral axis at four time points during the run. Time series lines represent the means; shading represents the standard deviations. Positive bending moments indicate bending in the concave posterior direction, suggesting anterior tension and posterior compression. B: Box plot displaying peak tibial bending moment at each of the four time points during the run ($n = 16$). Box plots represent the median and 1st and 3rd quartiles, whilst the error bars represent the range, excluding outliers (none present). Mean values for each participant are overlaid in filled circles as a swarm plot at each time point. *Significantly different ($p < 0.05$) from Start.

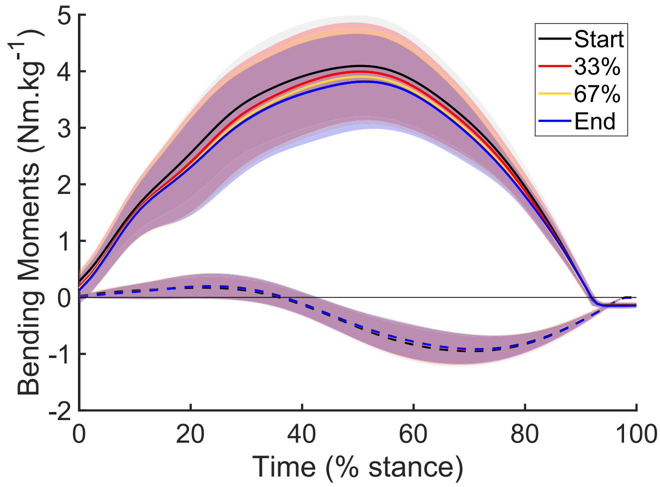


**Fig. 2.** Time-normalised resultant tibial bending moments are the sum of the bending due to muscular forces (solid lines) and bending due to joint reaction forces (dashed lines). Time series lines represent the means; shading represents the standard deviations. Positive bending moments indicate bending in the concave posterior direction, suggesting anterior tension and posterior compression.

activity of the plantar flexor muscles during stance remained high (>1.5 times EMG during isometric maximum voluntary contraction (also from the same dataset,[8])) and stable throughout the run (i.e. neuromuscular activation did not appear to change), it seems likely that the contractile capacity of the plantarflexor muscles declined due to progressive peripheral fatigue during the run, which in turn led to the reduced joint kinetics and tibial bending moments. Alternatively, the peak ankle joint moments could conceivably have decreased throughout the run without a concurrent change in muscle activity as a result of increased antagonist dorsiflexor muscle activity. However, this suggestion is speculative as dorsiflexor EMG data were not collected, and this is not supported by data from a similar run, in which tibialis anterior EMG activity decreased.[32]

The 9.1 % reduction in Achilles tendon forces was likely influenced by similar mechanisms to those discussed in relation to the distal tibia, i.e. the reduction in ankle joint moments and therefore reduced plantarflexor muscle force production resulted in a reduction in Achilles tendon force. In summary, the findings from the present study suggest that running to exhaustion *per se* may not be causative in the development of tibial and Achilles tendon overuse injuries. Given the importance of the magnitude of tissue loading to the risk of failure,[5] it seems that the mechanical changes that occur as a result of a fatiguing high-intensity run do not further increase the risk of tibial and Achilles tendon overuse injury. However, it should be recognised that the aetiology of overuse injuries may be more complex, influenced by the combination of loading magnitude and duration, and the capacity of the tissues to recover between bouts.[33] Therefore, the cumulative stress of fatiguing exercise could still be an important risk factor for overuse injury despite decreases in tibial bending moments and Achilles tendon forces during the run.

Conversely, peak patellofemoral joint contact forces increased throughout the run by 8.9 % which coincided with a 10.3 % increase in

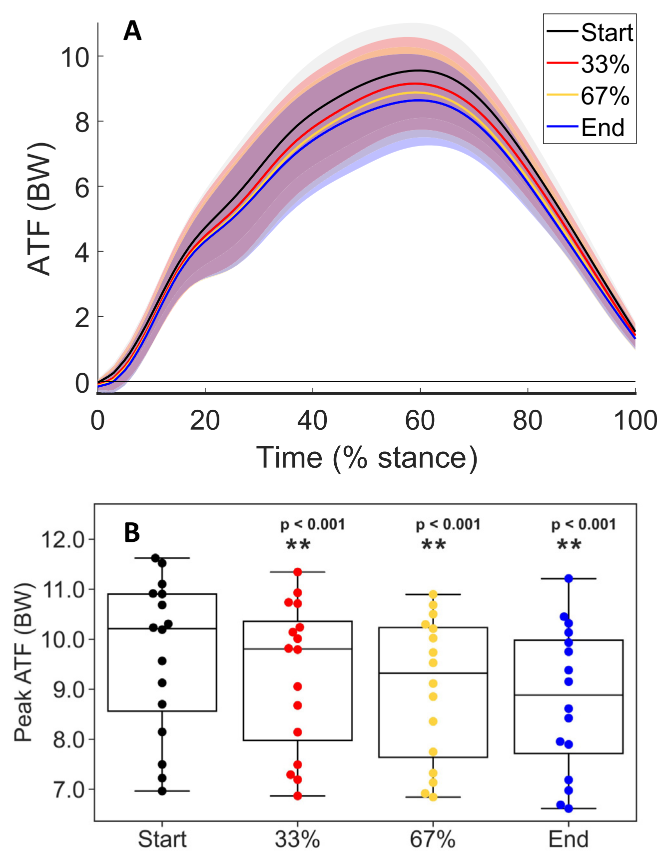


**Fig. 3.** A: Time-normalised Achilles tendon force (ATF) at four time points during the run. Time series lines represent the means; shading represents the standard deviations. B: Box plot displaying peak Achilles tendon force at each of the four time points during the run (n = 16). Box plots represent the median and 1st and 3rd quartiles whilst the error bars represent the range, excluding outliers (none present). Mean values for each participant are overlaid in filled circles as a swarm plot at each time point. **Significantly different ($p < 0.001$) from Start.

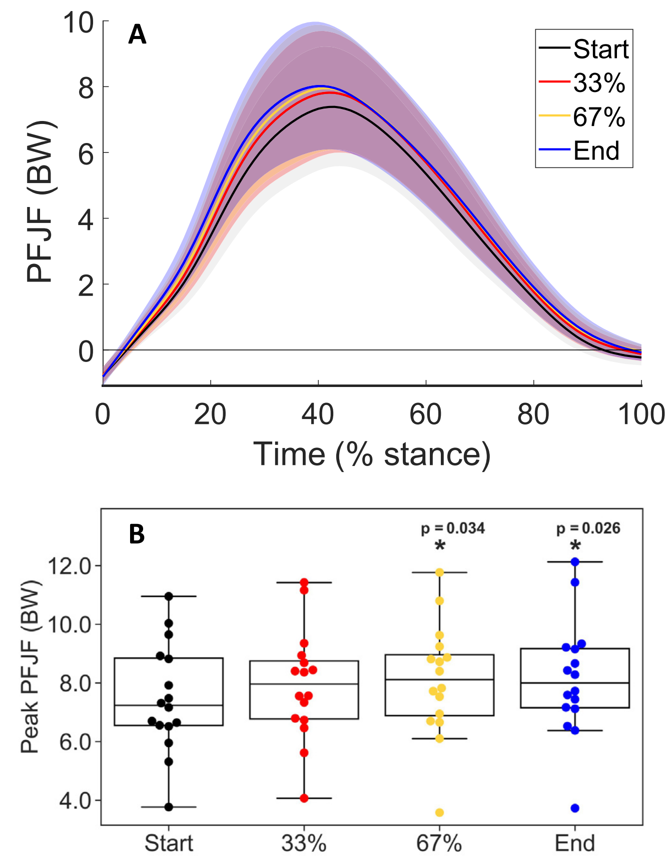


**Fig. 4.** A: Time-normalised patellofemoral joint force (PFJF) at four time points during the run. Solid lines represent the means; shading represents the standard deviations. B: Box plot displaying peak PFJF at each of the four time points during the run (n = 16). Box plots represent the median and 1st and 3rd quartiles whilst the error bars represent the range, excluding outliers. Mean values for each participant are overlaid in filled circles as a swarm plot at each time point. *Significantly different ($p < 0.05$) from Start.

stance phase peak knee extension moment and a 33.3 % increase in positive knee extension work.[8] This is in contrast to previous findings where patellofemoral joint contact forces were unchanged.[14] The differences between studies may be due to the slower running speed in the previous study. In the present study, these increases were concomitant with increased EMG activity of the *vastus lateralis* and *vastus medialis* muscles during late stance (previously reported from the same dataset[8]), although the changes in joint forces occurred close to midstance. Nonetheless, the findings are indicative of increased knee extensor forces that resulted in increased loading on the quadriceps tendon and therefore the patellofemoral joint. It could be assumed that this increased knee joint loading contributes to increased injury risk under prolonged and/or repeated fatigued running, although quantifying the practical consequence of this potential increase in injury risk is highly challenging.

In this study there was a reduction in distal ankle joint loading with a concomitant increase in proximal knee joint loading. It seems likely that a reduction in loading of one musculoskeletal structure leads to an increase in loading of other structures, particularly when running speed is kept constant. The observed changes appear to be systematic and align with previous findings which showed that the mechanical work done was redistributed from the ankle to the knee and hip throughout a 10 km run.[10] The current findings highlight the importance of considering the overall musculoskeletal effects of fatigue, rather than focusing on one isolated musculoskeletal structure. A so-called ‘reduced risk’ at one bodily site may lead to an increased risk at another, making it difficult to provide broad recommendations.

Peak tibial bending moments were 3.8 $\mathrm{Nm \cdot kg^{-1}}$ in the present study, higher than values of ~2.2 $\mathrm{Nm \cdot kg^{-1}}$ previously reported[9] although this difference is likely partly explained by the faster running speeds in the present study (5.7 $\mathrm{m \cdot s^{-1}}$ vs 4.5 $\mathrm{m \cdot s^{-1}}$). Achilles tendon and patellofemoral joint contact forces of 7.7 BW and 7.8 BW, respectively, have previously been reported when running at 5.6 $\mathrm{m \cdot s^{-1}}$,[23] whereas in the present study they were 9.7 BW and 7.5 BW respectively at the start of the run.

### *4.1. Limitations*

The present study includes limitations in addition to those that have previously been discussed in relation to this same dataset.[8] The modelling approaches used here are simplified estimates of tendon, bone or generic joint loading that do not account for indicators of tissue quality, or other participant-specific injury risk factors. In the case of patellofemoral pain, the generic modelling approach here does not consider the different tissues that are loaded and that are susceptible to injury at the knee, each of which may be subjected to damage through unique mechanisms. Furthermore, the modelling approaches employed here have not been evaluated in prospective injury studies which could confirm whether these indices of loading are associated with injury risk.

It is important to note that individual loading patterns and magnitudes may be critical for understanding injury risk and have not been considered in depth here. Whilst the magnitudes of change observed were small compared with the variability observed between participants, it is not currently clear what magnitude of change can be considered clinically meaningful, given that the magnitudes are well below the failure thresholds for the involved tissues. Moreover, the importance of cumulative loading (i.e. accumulated over time) rather than instantaneous peak loads is not well understood. It would be valuable to establish through prospective study whether higher magnitudes of estimated cumulative loading translate to increased risk of overuse injury or whether higher magnitudes of cumulative loading are associated with indicators of better tissue quality, greater remodelling or differences in structure.

In all three of the musculoskeletal structures in the present study, only sagittal plane loading was considered, whereas these tissues experience three-dimensional loading that may be crucial for injury development. In terms of study design, the present study provides insight into mechanisms for changes in tissue loading, but these mechanisms may only be representative of running at a fixed running speed on a treadmill. When fatigue mechanisms occur in-field, runners may select a different strategy such as reducing their running speed, and thus different changes in joint loading may be observed. Finally, higher loading should not necessarily be assumed to be negative, as loading within the adaptive capability of the tissues can lead to beneficial remodelling.

## 5. Conclusions

High-intensity running at a fixed speed resulted in an increase in patellofemoral joint contact forces, but a concurrent decrease in distal tibial bending moments and Achilles tendon forces. Thus, patellofemoral joint injury risk may increase under prolonged and/or repeated exhaustive running, whilst altered running mechanics due to fatigue may not in itself be a mechanism for Achilles tendon or tibial injury development. The findings highlight the importance of considering multiple tissues and joints when introducing recommendations for injury prevention, as altered loading of one bodily site may have converse implications for others.

## Funding information

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## Confirmation of ethical compliance

The study received ethical approval from the Loughborough University Ethics Approvals (Human Participants) Sub-Committee.

## CRediT authorship contribution statement

**Hannah Rice:** Conceptualization, Software, Formal analysis, Writing – original draft, Visualization. **Chelsea Starbuck:** Conceptualization, Formal analysis, Software, Writing – review & editing. **Jasmin Willer:** Conceptualization, Software, Methodology, Investigation, Writing – review & editing, Project administration. **Sam Allen:** Conceptualization, Investigation, Writing – review & editing. **Christopher Bramah:** Methodology, Writing – review & editing. **Richard Jones:** Methodology, Supervision. **Lee Herrington:** Methodology, Writing – review & editing. **Jonathan Folland:** Conceptualization, Formal analysis, Methodology, Investigation, Writing – review & editing, Visualization.

## Declaration of interest statement

There are no reported conflicts of interest.

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