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ORIGINAL RESEARCH ARTICLE

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Identifying Fatigue-Related Gait Patterns Using Multiple Inertial Measurement Units and Statistical Parametric Mapping: A Continuous Analysis of an Outdoor Full Marathon in Male Recreational Runners

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Abstract

Background Running is an effective exercise for personal fitness, yet many recreational runners suffer from running-related injuries. Prolonged running induces neuromuscular fatigue, interfering with an individual's preferred running gait and increasing the injury risk. This study aimed to examine gait patterns associated with fatigue in runners during a full marathon by analyzing lower limb segment and pelvis kinematics captured via multiple inertial measurement units (IMU).

Methods Three IMUs were attached to measure the rearfoot, shank, and pelvis kinematics of 23 male recreational runners during an outdoor marathon. Data were extracted for nine time points: the baseline, and at the 5th, 10th, 15th, up to the 40th kilometer. Each segment's free acceleration and angular velocity during the stance phase at these nine timelines were analyzed using statistical non-parametric mapping.

Results Male recreational runners exhibited a lower running speed (1.13 km/h, $p<0.001$), lengthened stance time (0.009 s, $p\leq 0.001$), and prolonged stride time (0.014 s, $p<0.05$) after 35 km of running, alongside a smaller anterior and superior acceleration of rearfoot and shank during the propulsion phase ($p<0.05$). With increasing running mileage, the rearfoot demonstrated a gradual increase in lateral acceleration and external rotation velocity during the propulsion phase ($p<0.01$). The shank exhibited a progressive decline in anterior tilt velocity during the loading response phase ($p<0.05$). Additionally, the pelvis displayed significantly greater anterior acceleration during propulsion at the 40 km mark ($p<0.01$).

Conclusions Male recreational runners exhibit a marked decline in performance only after 35 km. The progressive increase in rearfoot lateral acceleration and external rotation velocity during the propulsion phase, may be associated with a compensatory distal strategy to maintain balance stability. The gradual reduction in anterior tilt velocity of the shank during the loading response likely reflects a stiffness-enhancing mechanism in the lower limb to preserve locomotor efficiency under fatigue. The increased anterior acceleration of the pelvis at the 40 km mark suggests a proximal shift in propulsion mechanics due to fatigue. These findings underscore the necessity of long-distance protocols, continuous kinematic monitoring, and full stance-phase analysis to study running fatigue.

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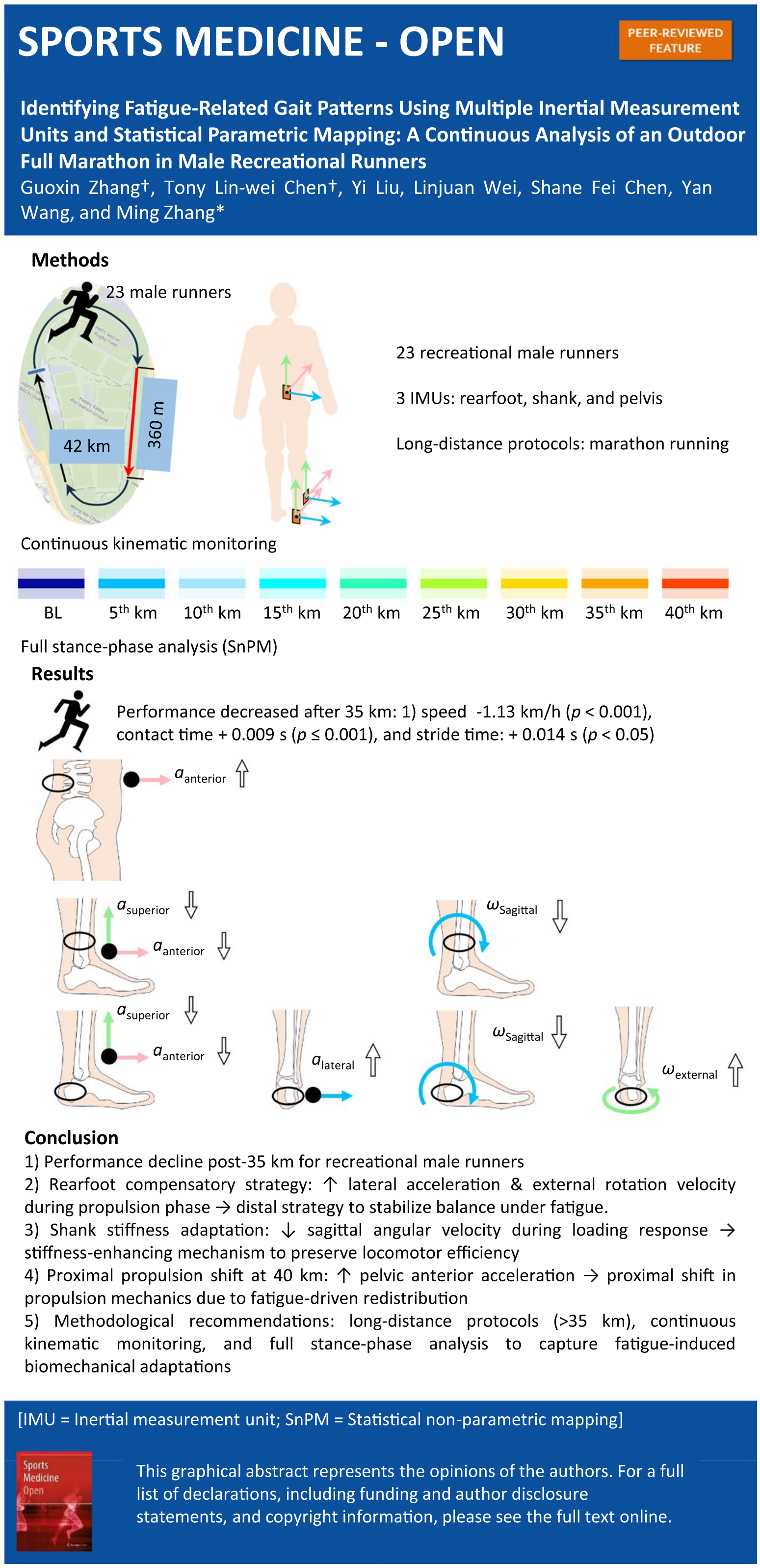
Key Points

- Male recreational runners maintained a stable pace for most of the marathon, but speed declined and stance time lengthened after approximately 35 km, signaling the first objective manifestation of fatigue.
- With fatigue, the rearfoot adopted a more lateral push-off, the shank advanced with reduced rotation speed, and the pelvis generated additional propulsive work—adjustments that appear to enhance balance, conserve energy through increased leg stiffness, and redistribute mechanical load proximally.
- These findings underscore the necessity of full-distance study protocols, continuous kinematic monitoring, and comprehensive stance-phase analysis to characterize authentic fatigue patterns and to inform evidence-based training and injury-prevention strategies for marathon runners.

Graphical Abstract



Background

Running is a simple and effective exercise method that requires minimal technical skills and low economic costs, significantly improving health and longevity [1]. Over the past decade, the number of runners has increased by approximately 60% [2]. Particularly after the outbreak of COVID-19, more individuals have taken up running as it strengthens the body, reduces chronic diseases, and allows for the enjoyment of fresh air and scenic views [3]. However, over 40% of runners, especially long-distance runners, suffer musculoskeletal injuries, most of which affect the lower limb [4]. Typical injuries include plantar fasciitis, Achilles tendinopathy, medial tibial stress syndrome, iliotibial band syndrome, and patellofemoral pain syndrome [4, 5]. One reason leading to these injuries is the interruption of an individual's preferred gait pattern, as the preferred gait pattern is the optimal movement path for an individual, minimizing joint and tendon loading through muscle tuning [6]. Prolonged and strenuous running, such as a full marathon, often leads to neuromuscular fatigue, impairing motor control ability and alignment [6], disrupting the individual's preferred running gait pattern, as reported in our previous study [7], and ultimately increasing the risk of injuries [8, 9]. Therefore, investigating changes in running gait patterns during prolonged running can clarify the mechanisms of musculoskeletal injury risk and has clinical significance for reducing the risk of running-related musculoskeletal injuries.

Changes in running gait patterns often occur due to neuromuscular fatigue during long-distance running. Positive work redistributes from the distal ankle to the more proximal knee and hip during prolonged runs [10]. The plantar loading pattern is altered as fatigue in the local muscles of the foot and shank reduces the involvement of the toes and increases dorsiflexion in the metatarsophalangeal joints [8, 9]. Consequently, plantar loading distribution shifts from the toes to the metatarsal heads, inducing higher peak pressure and impulse at the medial and central metatarsal heads and increasing the incidence of stress fractures in the metatarsals. Endurance runs also change joint angles, including increased ankle dorsiflexion and eversion [11, 12], tibial and knee internal rotation [12], knee flexion [13], hip adduction [14], and trunk flexion [15]. Recreational runners are very likely to change their strike pattern from non-rearfoot to rearfoot in the late stages of a marathon to cope with fatigue [16]. Step length may increase [17], and while stride time and contact time are only slightly affected by fatigue [9] but their predictability is significantly reduced [18].

Although previous studies have extensively examined important gait parameters, they typically have four limitations. Firstly, related research is usually conducted on treadmills in laboratories, while the outdoor environment is much more complex. Overground running can result in different foot strike angles at initial contact than treadmill running [19, 20]. Secondly, the distance protocols adopted in previous studies typically ranged from 5 to 10 km, which may not be long enough to induce fatigue [21]. Thirdly, most studies only report the outcome measures before and after the fatigue protocol. The patterns of changes that occurred between the two observation points were rarely documented. Fourthly, discrete data points related to prominent kinetic events (e.g., initial foot contact or peak values) are usually selected to represent a gait cycle, which cannot capture the full spectrum of changes throughout the stance phase. These limitations can be adequately addressed by a study that involves continuous monitoring of the gait patterns during overground running that covers extended running mileage, such as a full marathon.

Given the latest advancements in wearable motion tracking sensors, a study in this fashion is feasible. A nine-axis inertial measurement unit (IMU) that integrates accelerometers, gyroscopes, and magnetometers can measure three-dimensional segment kinematics in an outdoor research environment. Our previous studies demonstrated the utility of IMUs in assessing fatigue-related gait patterns [22, 23]. Other than that, statistical parameter mapping (SPM) allows for statistical analysis of continuous data, enabling the observation of changes in gait patterns throughout the entire gait cycle [24]. Therefore, we experimented to investigate the effects of running mileage on lower limb segment kinematics in runners during a full marathon. Outcome variables of interest, including three-dimensional segment linear acceleration and rotation velocity, were reported at selected mileage checkpoints (every five kilometers accomplished). These variables were computed as a function of a stance phase and compared between the baseline and the other checkpoints using SPM. We hypothesize that fatigue-induced alterations in movement patterns predominantly localize to the foot during the loading response and the propulsion phases of the stance phase in the later stages of prolonged marathon running.

Methods

Participants

Given the known physiological and biomechanical differences between males and females—particularly in pelvis-hip mechanics, neuromuscular fatigue patterns, and metabolic processes—this study exclusively recruited male runners to eliminate potential sex-related influences on the results [25, 26]. We recruited 23 male recreational runners (age: 45 ± 6 years, range: 30–57 years; height: 172.3 ± 3.8 cm; weight: 61.7 ± 3.9 kg) from

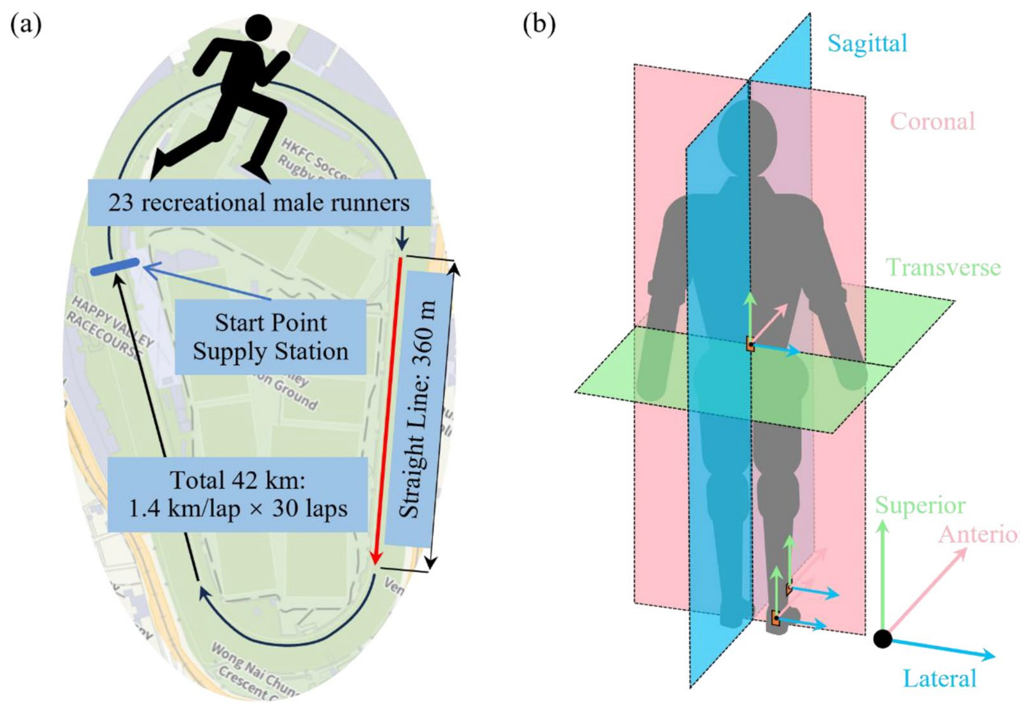


Fig. 1 Schematic diagrams of the running track (**a**), sensor installation location and coordinate system (**b**)

the local community by posters and flyers with a best marathon time under 210 min. The sample size of 23 was determined using G*Power (Version 3.1.9.7, Universität Düsseldorf, Düsseldorf, Germany) [27] for an F-test of within-factors with nine repeated measurements. Parameters included: significance level $(\alpha)=0.05$, power $(1-\beta)=0.8$, and a small-to-medium effect size (Cohen's $f=0.2$) [28]. This conservative effect size estimate was selected due to the absence of directly comparable prior studies, following Cohen's conventions [28] (small: $f=0.10$, medium: $f=0.25$, large: $f=0.40$) and recommendations for behavioral/biomedical sciences when preliminary data are limited. Each participant received 500 HKD as compensation. The participants had an average of 13 ± 11 marathon race experiences, with their best and average performance being 181 ± 14.97 and 202 ± 24.71 min, respectively. Their weekly running mileage was 67.39 ± 28.16 km, with a frequency of 5 ± 2 times per week and an average pace of 5:01 ± 1:09 min per kilometer. None of the participants had musculoskeletal injuries in the past 6 months. All participants were fully informed of the research protocols and provided written informed consent before the study commenced. The study was reviewed and approved by the University Human Subjects Ethics Sub-Committee (Reference No: HSEARS20210125006).

Experiment Protocol

Each participant performed an outdoor marathon run (Fig. 1a). The running track is predominantly flat and obstacle-free, featuring an asphalt-paved surface. All participants wore uniform compression pants with three IMUs (Xsens DOT 2nd Generation, Movella, Enschede, Netherlands) attached to the right foot, right shank, and pelvis [22, 29–31] (Fig. 1b). We selected the posterior heel to represent the foot segment because approximately 89% of recreational runners adopt a rearfoot strike pattern [16]. This anatomical location is the first to endure impact forces during the loading phase, rendering it biomechanically superior to alternative shoe placements (e.g., tongue) for detecting strike kinematics and fatigue-related adaptations. The IMU on the posterior of heel was secured using double-sided adhesive skin interfaces (Sensor tape, Delsys Inc., Natick, Massachusetts, USA) and kinesiology tape (KMB 50, McDavid Inc., Fountain Valley, California, USA); the IMU on the shank was fixed above the lateral malleolus with double-sided adhesive skin interfaces and a manufacturer-provided strap (Strap set, Movella, Enschede, Netherlands); the IMU on the pelvis was anchored via double-sided adhesive skin interfaces and a customized pocket sewn onto compression pant, positioned approximately in the L3–L5 lumbar vertebrae region. The nine-axis IMUs weigh only 11.2 g and consist of a triaxial accelerometer (range: ± 16 g,

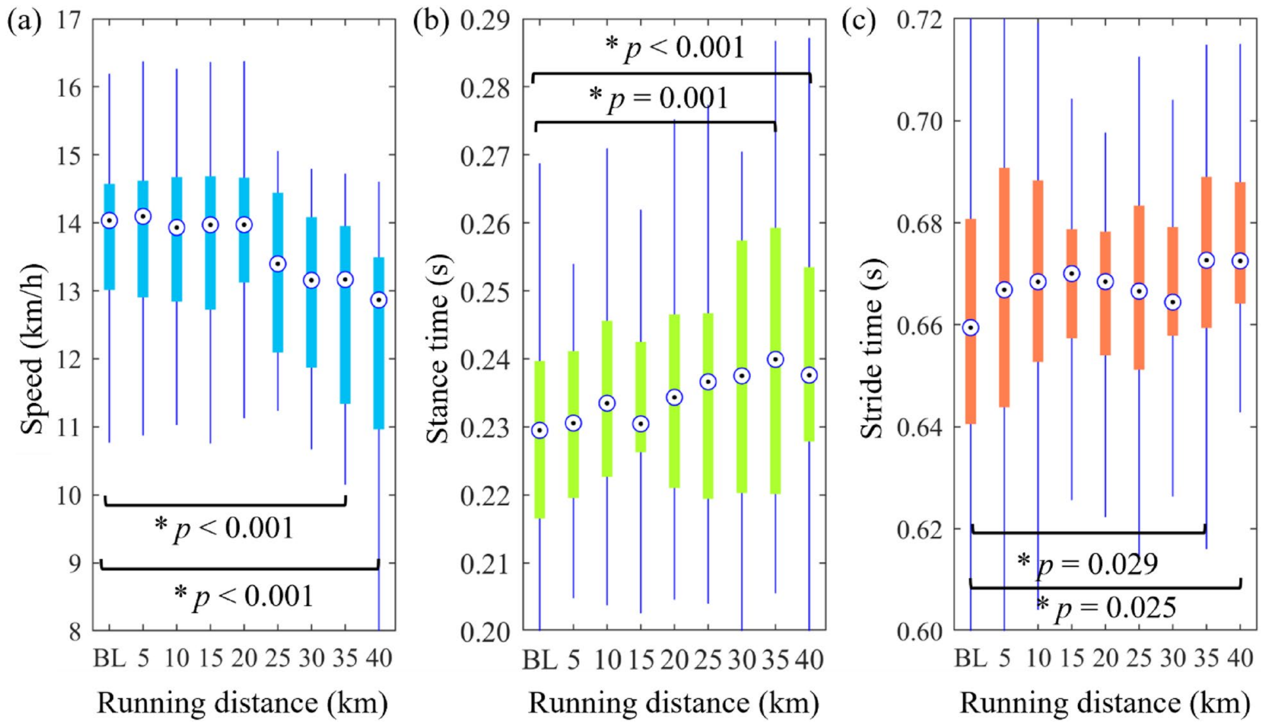


Fig. 2 Changes in running speed (**a**) and stance time (**b**). Box plots display the median (blue circles and points) and interquartile range (box edges: 25th–75th percentiles). Asterisks (*) indicate statistically significant differences between baseline (BL) and subsequent time points (*$p<0.05$)

resolution: 0.49 mg, sample frequency: 800 Hz), a triaxial gyroscope (range: ± 2000°/s, resolution: 0.06°/s, sample frequency: 800 Hz), and a triaxial magnetometer (range: ± 8 Gauss, resolution: 0.25 mGauss, sample frequency: 60 Hz). After the sensor's embedded signal processing architecture dynamically synchronizes and standardizes all sensor signals, the maximal recoding and export frequency is capped at 120 Hz.

Data were collected synchronously at a sampling rate of 120 Hz throughout the entire marathon run. Two sets of IMUs were used for data collection, as each IMU can record for a maximum of 3 h at 120 Hz. The IMUs were replaced at the halfway point of the marathon, with the exchange time minimized to ensure negligible impact on the results. Participants had ample time to warm up before the marathon test began while wearing their appropriate running shoes. Participants were divided into seven groups (3–5 individuals per group) to complete the running experiment separately. Participants were instructed to complete the running task with maximal effort to ensure the collected data closely approximated real race data. Mimicking an official marathon, a supply station was set up for runners to replenish water and energy (Fig. 1a) freely.

Data Processing and Outcome Variables

Data were first exported from the IMUs (Movella DOT Data Exporter 2023.6, Movella, Enschede, Netherlands) and then processed using custom code in MATLAB (2024a, MathWorks Inc., Natick, Massachusetts, USA). Since the IMUs integrated strap-down integration of inertial data and sensor fusion using the Movella Kalman filter core algorithm, no additional data filtering was conducted in this study. Free acceleration (a_{free}) is calculated as the acceleration vector in the sensor-fixed coordinate system (a_{sensor}) minus the components of gravity (g) on each axis (Eq. 1):

$$a_{free} = a_{sensor} - R_{SL} \times g \tag{1}$$

where R_{SL} is the rotation matrix from the local earth-fixed reference coordinate system to the sensor coordinate system, and it is equal to the transposition of the rotation matrix from the sensor coordinate system to the local earth-fixed reference coordinate system (R_{LS})) (Eq. 2):

$$R_{SL} = (R_{LS})^T. \tag{2}$$

A quaternion (Eq. 3) is an efficient and non-singular way to describe three-dimensional rotations, containing information about the rotation angle (q_0) and the

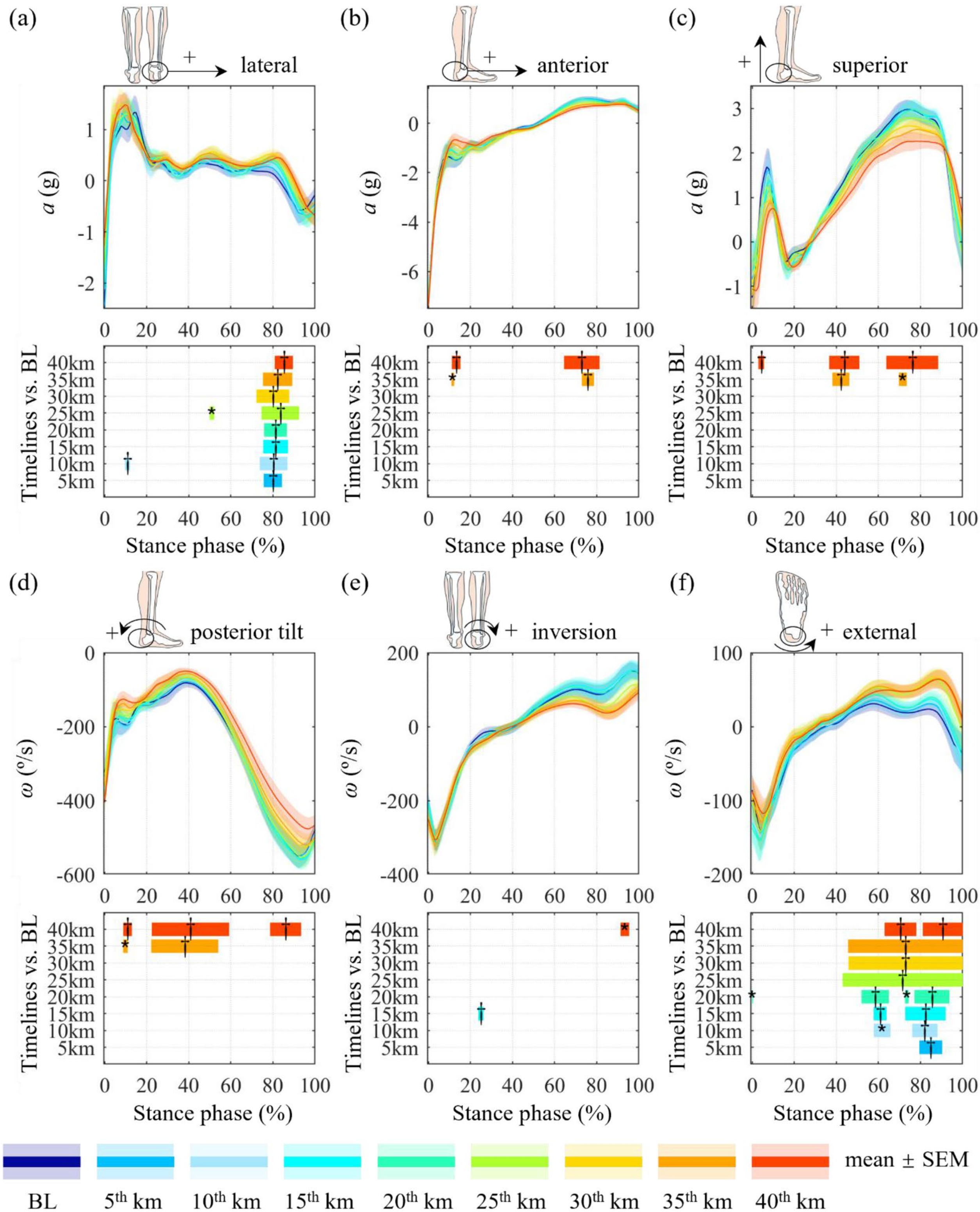


Fig. 3 Changes in three-axis free acceleration and angular velocity of the rearfoot: **a** medial–lateral acceleration; **b** anterior–posterior acceleration; **c** superior-inferior acceleration; **d** sagittal rotation speed; **e** coronal rotation speed; **f** transverse rotation speed. Asterisks (*) and daggers (†) denote statistically significant differences (*$p < 0.05$; †$p < 0.01$) between baseline (BL) and subsequent time points. *SEM* Standard error of the mean

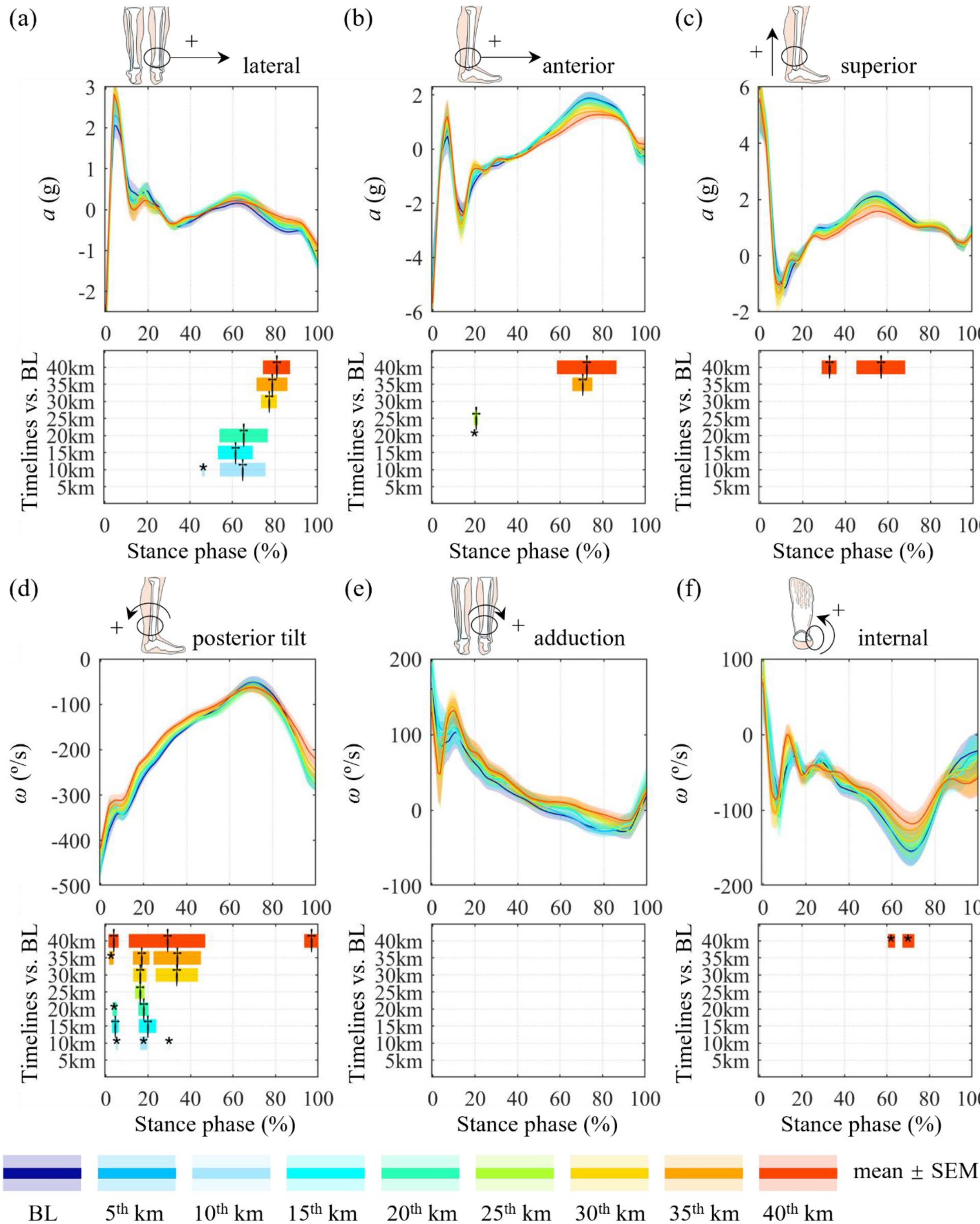


Fig. 4 Changes in three-axis free acceleration and angular velocity of the shank: **a** medial–lateral acceleration; **b** anterior–posterior acceleration; **c** superior-inferior acceleration; **d** sagittal rotation speed; **e** coronal rotation speed; **f** transverse rotation speed. Asterisks (*) and daggers (†) denote statistically significant differences (*$p<0.05$; †$p<0.01$) between baseline (BL) and subsequent time points. *SEM* Standard error of the mean

rotation axis in a three-dimensional coordinate system (q_1, q_2, q_3):

$$q_{LS} = [q_0, q_1, q_2, q_3]. \tag{3}$$

According to the manufacturer, the R_{LS} can be described using quaternions (Eq. 4) [32]:

$$R_{\mathrm{LS}} = \begin{bmatrix} 2 \times q_0^2 + 2 \times q_1^2 - 1 & 2 \times q_1 \times q_2 - 2 \times q_0 \times q_3 & 2 \times q_1 \times q_3 + 2 \times q_0 \times q_2 \\ 2 \times q_1 \times q_2 + 2 \times q_0 \times q_3 & 2 \times q_0^2 + 2 \times q_2^2 - 1 & 2 \times q_2 \times q_3 - 2 \times q_0 \times q_1 \\ 2 \times q_1 \times q_3 - 2 \times q_0 \times q_2 & 2 \times q_2 \times q_3 + 2 \times q_0 \times q_1 & 2 \times q_0^2 + 2 \times q_3^2 - 1 \end{bmatrix}. \tag{4}$$

Finally, the free acceleration in the three-axis was calculated (Eq. 5):

$$a_{free} = \begin{bmatrix} a_{x,free} \\ a_{y,free} \\ a_{z,free} \end{bmatrix} = \begin{bmatrix} a_{x,sensor} - 2 \times q_1 \times q_3 + 2 \times q_0 \times q_2 \\ a_{y,sensor} - 2 \times q_2 \times q_3 - 2 \times q_0 \times q_1 \\ a_{z,sensor} - 2 \times q_0^2 - 2 \times q_3^2 + 1 \end{bmatrix} \tag{5}$$

where $a_{x,\ sensor}$, $a_{y,\ sensor}$, and $a_{z,\ sensor}$ are the three-axis accelerations obtained from the IMUs sensor with the unit converted to gravity acceleration g.

Data analysis was based on gait cycles, with gait events detected using accelerations of the shank. Specifically, initial contact was identified using the peak of resultant acceleration, and toe-off was determined by the average timestamp of local minima of the anterior–posterior and medial–lateral accelerations in the 20%–50% gait cycle [33, 34]. The coordinate system was defined as anterior ($+x$), proximal ($+y$), and lateral ($+z$) (Fig. 1b) [35, 36]. The outcomes included running speed, stance time, triaxial free accelerations, and angular velocities. Each stance phase's free accelerations and angular velocities were time-normalized to 101 points (0 to 100% gait cycle). The stance phase is typically divided into two primary tasks: weight acceptance (0–50% of the stance phase) and propulsion (51–100% of the stance phase). These two tasks are further subdivided into functional intervals: the loading response period (around 0–35% of the stance phase) within weight acceptance, and the pre-swing period (around 75–100% of the stance phase) during propulsion [37].

All variables were averaged across 50 strides for each timeline, including the baseline, the 5th km, the 10th km, …, and the 40th km, for nine timelines. The data of each timeline were extracted from the 360 m straightaway illustrated in Fig. 1a. The baseline was defined as the 2nd km to avoid the potential effect of the adaptation phase [38]. Similarly, to avoid potential end-phase effects, we excluded the final lap of each measurement. Specifically, the baseline, 5km, 10km, 15km, 20km, 25km, 30km, and 40km timepoints correspond to laps 2, 5, 8, 11, 14, 18, 22, 26, and 29, respectively.

Statistical Analysis

Since some outcomes did not satisfy the assumptions of normal distribution or sphericity, nonparametric tests were used to compare the baseline with the other timelines. Specifically, the triaxial free accelerations and angular velocities of each segment were analyzed using statistical non-parametric mapping (SnPM) [39]. Running speed and stance time were analyzed using the Friedman test and multiple pairwise comparisons. The significance level was set to $p < 0.05$ with the Bonferroni correction.

Results

Speed, Stance Time, and Stride Time

All participants completed the marathon with an average time of 203.26 ± 20.53 min (average speed: 12.46 ± 1.14 km/h), which is very close to their previous average performance (202 ± 24.71 min). The median running speed displayed significant declines after 35 km due to cumulative fatigue, decreased from 14.04 km/h at baseline to 12.87 km/h at the 40th km, a reduction of 1.13 km/h (Fig. 2a, $p < 0.001$). The stance time increased from 0.229 s at baseline to 0.238 s at the 40th km, with an increase of 0.009 s (Fig. 2b, $p < 0.001$). Meanwhile, the stride time increased from 0.659 s at baseline to 0.673 s at the 40th km, with an increase of around 0.014 s (Fig. 2c, $p = 0.025$).

Rearfoot

The changes in triaxial free acceleration and angular velocity at the rearfoot, along with the corresponding statistical results, are shown in Fig. 3. The medial–lateral acceleration significantly shifted laterally during the propulsion phase throughout the marathon (Fig. 3a). At the 85% stance phase, the medial–lateral acceleration changed significantly from 0.11 g medially at baseline to 0.39 g laterally at the 40th km ($p < 0.01$). Although insignificant, the posterior acceleration increased from 6.90 g at baseline to 7.40 g at the 40th km at the initial contact (Fig. 3b). However, compared to the baseline, the posterior acceleration significantly decreased during the middle loading response (12–15% stance phase, $p < 0.01$), and the anterior acceleration decreased during the middle of

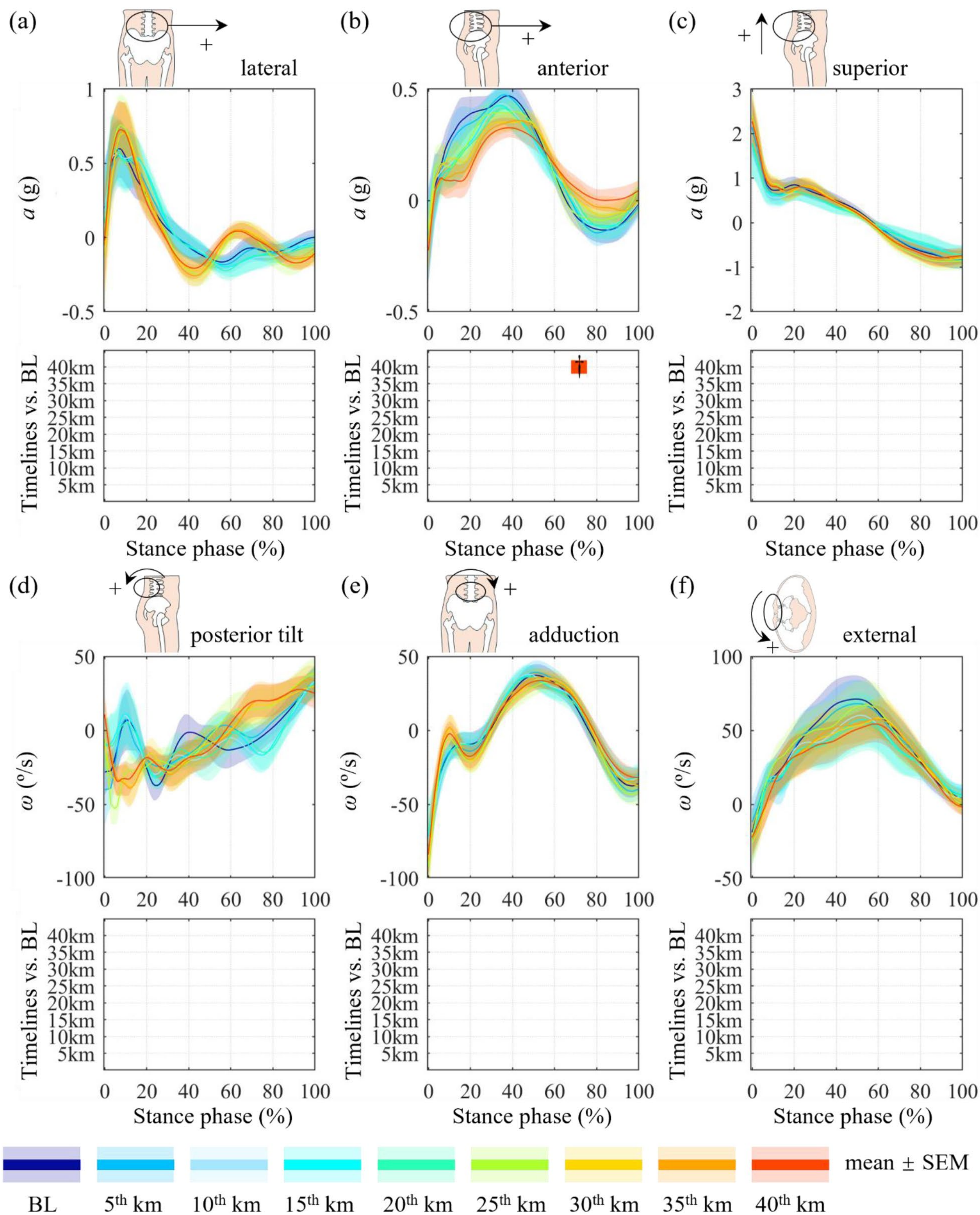


Fig. 5 Changes in three-axis free acceleration and angular velocity of the pelvis: **a** medial–lateral acceleration; **b** anterior–posterior acceleration; **c** superior-inferior acceleration; **d** sagittal rotation speed; **e** coronal rotation speed; **f** transverse rotation speed. Asterisks (*) and daggers (†) denote statistically significant differences (*$p < 0.05$; †$p < 0.01$)) between baseline (BL) and subsequent time points. *SEM* Standard error of the mean

the propulsion phase (65–81% stance phase, $p < 0.01$) at the 40th km. The superior-inferior acceleration showed a delayed trend and was reduced during early loading response ($p < 0.01$), late weight acceptance (37–50% stance phase, $p < 0.01$), and middle propulsion phase (64–88% stance phase, $p < 0.01$) at 40th km, compared to the baseline (Fig. 3c).

The anterior tilt velocity increased during the early loading response phase, though not significant (Fig. 3d). For example, the anterior tilt speed increased from 323°/s at baseline to 405°/s at the 40th km at the initial contact. This rotation velocity significantly decreased during the second half weight acceptance (23–50% stance phase, $p < 0.01$) and propulsion phase (79–93% stance phase, $p < 0.01$). The inversion velocity significantly decreased during the propulsion phase (Fig. 3e). At the 95% stance phase, the inversion speed decreased from 149°/s at baseline to 68°/s at the 40th km ($p < 0.05$). The external rotation velocity displayed a significant trend towards external during the propulsion phase throughout the marathon (Fig. 3f), increasing from 20 and − 36°/s in baseline to 48°/s and 10°/s in the 40th km at 70% and 100% stance phase, respectively ($p < 0.01$).

Shank

The changes in triaxial acceleration and angular velocity at the shank, along with the corresponding statistical results, are shown in Fig. 4. The medial–lateral acceleration of the shank showed a similar trend to the rearfoot, significantly shifting laterally during the propulsion phase (Fig. 4a). At the 85% stance phase, the lateral acceleration increased from − 0.54 g at baseline to 0.23 g at the 40th km ($p < 0.01$). Though not significant, the posterior acceleration also increased from 4.73 g at the baseline to 5.68 g at the 40th km (Fig. 4b). The anterior acceleration decreased during the early propulsion phase (59–86% stance phase, $p < 0.01$). At initial contact, the superior acceleration increased from 5.51 g at baseline to 5.58 g at the 40th km, though not significantly (Fig. 4c). The superior acceleration significantly decreased during the late loading response phase (29–35% stance phase, $p < 0.01$) and early propulsion phase (51–67% stance phase, $p < 0.01$), decreasing from 0.99 g and 1.98 g at baseline to 0.59 g and 1.52 g at the 40th km at the 30% and 60% stance phase, respectively.

The anterior tilt velocity decreased during the weight acceptance phase (2–6% and 11–46% stance phase, $p < 0.01$) and the terminal propulsion phase (94–100% stance phase, $p < 0.01$), compared to the baseline (Fig. 4d). The internal speed of the shank decreased during the early propulsion phase (61–63% and 68–72% stance phase, $p < 0.05$), from 154°/s at baseline to 118°/s at the 40th km (70% stance phase) (Fig. 4f).

Pelvis

The changes in triaxial acceleration and angular velocity at the pelvis, along with the corresponding statistical results, are shown in Fig. 5. Unlike the accelerations at the rearfoot and shank, which showed significant differences in all three planes, the pelvis only displayed differences in the anterior–posterior direction during the mid-propulsion phase (68–75% stance phase, $p < 0.01$), changing from 0.07 g posteriorly to 0.06 g anteriorly at the 70% stance phase (Fig. 5b). The angular velocity of the pelvis showed no significant difference in all three planes (Fig. 5d–f).

Discussion

This study investigated how prolonged distance running induced fatigue alters gait patterns. We hypothesized that movement pattern alterations would predominantly manifest in the foot during the propulsion phase of the stance phase. We found that fatigue-induced impacts on motor performance in recreational male marathon runners primarily manifest after the 35 km mark, characterized by increased stride time and contact time, alongside decreased forward velocity of the body, as well as gradually reductions in anterior and superior acceleration and anterior tilt velocity of the foot and shank during the propulsion phase. Furthermore, as the running distance progressed, the rearfoot exhibited a progressive lateral deviation and increased external rotation velocity during propulsion. Notably, the pelvis exhibited a significant increase in anterior acceleration exclusively during the propulsion phase at the 40 km mark, with no such changes observed at earlier stages of the marathon. These results generally supported our hypothesis and validated the necessity of the methodological framework—marathon-distance protocols, continuous kinematic monitoring, and full stance phase analysis—for capturing dynamic fatigue-related movement adaptations in male runners.

The decrease in running speed after the marathon indicates a decline in motor performance in the sagittal plane due to neuromuscular fatigue [40]. This reduced motor ability is also evident in the increased stance time and stride time, similar results were also reported in previous studies [8, 9], which allows runners more time to adjust posture and maintain balance while fatigued [41]. Although changes in stride time and contact time were minimal, fatigue reduces the precision, consistency, and coordination of optimal movement patterns, leading to increased unpredictability in stride timing [18]. Generally, the acceleration and rotation speed of the lower limb and pelvis should decrease as speed decreases. This is one reason why many runners choose to reduce their speed to mitigate impact after fatigue during prolonged races [42]. However, the forward rotation speed increased during early loading response. As more than 90% of recreational and sub-elite marathon runners adopt a rearfoot strike pattern in the second half of the marathon [16], a landing pattern associated with a decrease in arch height [6, 43], greater impact may impair the intrinsic muscles and fascia of the foot, potentially leading to conditions such as medial tibial stress syndrome, Achilles tendinopathy, and plantar fasciitis [4, 5]. The paradoxical result of increased impact despite decreased running speed can be attributed to the fatigue of the quadriceps, which leads to a reduced cushioning ability [11] and decreased running economy after fatigue [13, 44].

Unlike the more rapid anterior tilt velocity of the rearfoot during early loading response, the shank exhibited significantly reduced anterior tilt velocity during this period. The decreased rotation speed of the shank can be interpreted as increased stiffness in the lower extremity to counteract neuromuscular fatigue [41]. However, this opposite behavior of the distal and proximal ankle joints may reflect reduced coordination. This worsening coordination can inhibit running performance and lead to impairments in the ankle and knee joints [7]. Interestingly, the decline in ankle joint coordination in the sagittal plane did not appear in our previous half-marathon study [7], which may be related to the onset mechanism of neuromuscular fatigue. During the second half of the weight acceptance phase, the accelerations of the rearfoot and shank significantly decreased in the superior-inferior directions after prolonged running, especially after 35 km. This decrease is consistent with the observed reduction in forward velocity and can be attributed to neuromuscular fatigue [40], such as the decreased muscle activities in gastrocnemius and soleus [9]. The anterior tilt speed of the rearfoot and shank slowed during the late weigh acceptance phases, indicating diminished kinematics of the foot and shank in the sagittal plane.

During the propulsion phase, it is noteworthy that the rearfoot's external rotation speed gradually increased, which may contribute to increased internal rotation of the knee, a phenomenon commonly observed during fatigued running [12, 14]. One possible explanation is the application of a distal strategy of gait modification, which is often the first choice to reduce running-related injuries [45]. Compared to the knee, ankle fatigue has a smaller impact on postural stability [46]. However, we believe this pronounced lateral movement and faster external rotation imply decreased gait stability in the transverse plane [23]. The reduced control ability in the transverse plane triggers Achilles tendinopathy [47]. Previous studies demonstrated that running-induced fatigue increases ankle dorsiflexion [11] and eversion [12, 14, 15]. Although this study did not directly measure these two parameters, the observed gradually reductions in rearfoot's anterior tilt and inversion speed during the propulsion phase indirectly support these findings. The increased ankle dorsiflexion is the alteration of plantar loading patterns that the loading redistributed from toes to metatarsophalangeal joints with higher plantar force, peak pressure, impulse, and contact time at the metatarsal regions [8, 9].

It is also worth noting that the anterior acceleration of the pelvis gradually increased during the propulsion phase, which might be a manifestation of the core and proximal muscle compensation mechanism following lower extremity muscle fatigue [15, 48]. The core and proximal muscles have greater volume and cross-sectional area than the distal muscles and are more tolerant to fatigue [49, 50]. During the push-off phase of fatigued running, the proximal muscles take on more responsibility as the distal muscles fatigue, with positive work shifting from ankle to the knee and hip joints by external ground-reaction force lever arm and joint torque [10, 15, 48–50]. Following fatigue, a temporal delay in rearfoot kinematics (superior-inferior acceleration and external rotation velocity) during the terminal propulsive phase were observed alongside increased anterior pelvic acceleration. This coordination pattern—akin to the pelvis 'pulling' the foot forward—supports the theory of proximal compensation under distal muscle fatigue. The acceleration and rotation of the rearfoot displayed a gradually lateral and external shift, respectively, indicating a gradual increase in joint power at the ankle in the transverse plane during the terminal part of the propulsive phase after fatigued running [44].

This study facilitates a better understanding of changes in gait patterns related to potential neuromuscular fatigue during outdoor overground endurance running. Alterations in certain movement patterns of the lower limbs and trunk were observed only after 35 km of running, which suggests that shorter distances, such as 5–10 km, may not sufficiently induce fatigue in recreational male runners. Continuous monitoring of gait kinematics reveals fatigue-driven progressive changes in lower-limb gait during prolonged distance running, which is superior to analyses that only compare pre-fatigue and post-fatigue states. SnPM analysis confirmed that fatigue-induced alterations in gait patterns predominantly manifest during the propulsion phase of the stance phase, which provides more meaningful insights than analyses focusing solely on isolated time points (e.g., peak values). Runners aiming to enhance performance should prioritize endurance training targeting the lower limbs, including the gastrocnemius and soleus, as improving the capacities of the plantar flexor muscle–tendon unit may contribute to enhanced athletic performance [10]. The findings of this study (e.g., excessive external rotation velocity of the rearfoot during the propulsion phase) indicate that targeted strengthening of hip abductors (e.g., gluteus medius) may mitigate running-related injuries caused by fatigue.

This study has several limitations. Firstly, it included only male recreational runners aged 30–57 years, which means the findings may not apply to other groups of runners. While sex-specific recruitment enhanced internal validity for detecting fatigue-related gait changes, it inherently limits the generalizability of our findings to female runners. Previous studies have demonstrated that sex differences exist in fatigue resistance and biomechanics between males and females, due to factors such as skeletal muscle physiology, muscle perfusion, voluntary activation, and metabolic mechanisms [25, 26]. Although sex differences exist in biomechanical parameters during running, the trends in movement pattern changes remain consistent during fatigued running. Secondly, the study was conducted on a track-like playground, which did not account for influencing factors such as inclination angle, surface flatness, and obstacles commonly found on pavements, which have a non-negligible impact on gait patterns [51]. Future studies should consider recruiting a more diverse group of runners, including females and individuals of different ages, to investigate the effects of sex and age on running gait patterns during prolonged running. Additionally, future research could benefit from incorporating multimodal sensing systems to deepen our understanding of how fatigue affects gait. For example, integrating techniques such as electromyography and oxygen sensing may allow researchers to assess changes in muscle activation patterns, coordination, and localized metabolic processes.

Conclusion

This study focused on the changes in gait patterns during prolonged outdoor running and used multiple IMUs to monitor the rearfoot, shank, and pelvis kinematics. As the running mileage increases, recreational male runners experienced a decrease in forward velocity and an increase in stance time and stride time after 35 km of running, alongside the decrease of anterior and superior acceleration of rearfoot and shank during the propulsion phase. During the propulsion phase, the significant reduction in anterior tilt velocity of the rearfoot may reflect a biomechanical adaptation where the foot moderates plantarflexion speed, enabling the metatarsal joints to assume a greater role in propulsion. Concurrently, the progressive increase in rearfoot external rotation velocity during this phase could be linked to compensatory internal knee rotation, potentially serving as a distal strategy to stabilize transverse plane balance. In the loading response phase, the shank exhibited a gradual decline in anterior tilt velocity, likely indicative of a fatigue-mitigation mechanism wherein the lower limb enhances stiffness to preserve locomotor efficiency. Notably, at the 40 km mark, the pelvis demonstrated a pronounced increase in anterior acceleration during propulsion, suggesting a shift in propulsion dynamics from distal to proximal structures as fatigue accumulates in prolonged marathon running. The findings of this study validate the necessity of employing a methodological framework that integrates long-distance protocols, continuous kinematic monitoring, and full stance phase analysis to investigate fatigued running mechanics in male runners. Male recreational runners may benefit from targeted strengthening of the plantar flexors and hip abductors to optimize running performance and reduce injury risk.

Abbreviations

IMU	Inertial measurement units
SnPM	Statistical non-parametric mapping
SPM	Statistical parameter mapping

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Author Contributions

GZ and TLC participated in the research design, data collection, data analysis, results interpretation, and wrote the manuscript; YL, LW, SFC, and YW contributed to participant recruitment, data collection, data analysis, and proofreading of the manuscript; MZ conceived this study, participated in research conceptualization, results interpretation, and helped to write and revise this manuscript. All authors have read and approved the final version of the manuscript and agree with the order of the presentation of the authors.

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Data Availability

Please contact the corresponding author for data requests.

Declarations

Ethics Approval and Consent to Participate

This study complied with the Declaration of Helsinki and was reviewed and approved by the Hong Kong Polytechnic University Human Subjects Ethics Sub-Committee (Reference No: HSEARS20210125006). All participants were fully informed of the research protocols and provided written informed consent before the study commenced.

Consent for Publication

Not applicable.

Competing interests

The authors declare that they have no competing interests.

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