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Effect of foot pronation during distance running on the lower limb impact acceleration and dynamic stability

LIANGLIANG XIANG[1, 2, 3], YAODONG GU[1, 2, 3]*, ALAN WANG[3, 4], QICHANG MEI[1, 2], PEIMIN YU[1, 2, 3], VICKIE SHIM[3], JUSTIN FERNANDEZ[2, 3, 5]

[1] Faculty of Sports Science, Ningbo University, Ningbo, China.
[2] Research Academy of Grand Health, Ningbo University, Ningbo, China.
[3] Auckland Bioengineering Institute, The University of Auckland, Auckland, New Zealand.
[4] Faculty of Medical and Health Sciences, The University of Auckland, Auckland, New Zealand.
[5] Department of Engineering Science, The University of Auckland, Auckland, New Zealand.

Purpose: Foot pronation is not an isolated factor influencing lower limb functions. Exploring gait variability and impact loading associated with the foot posture are crucial for understanding foot pronation-related injury mechanisms. This study aimed to evaluate how foot posture affects impact loading and running variability during running. *Methods*: Twenty-five male participants were recruited into this study. Pressure under the foot arch, acceleration and marker trajectory were recorded in the right limb for each runner after 1, 4, 7 and 10 km running, respectively. Linear mixed effects models were used to analyze the statistical difference of the data. *Results*: FPI-6 has significantly increased after the 10 km running ($p < 0.01$). For the tibial acceleration, peak resultant acceleration after 10 km running was significantly increased than after 4 km running ($p = 0.02$). At the dorsum of the foot, the short-time largest Lyapunov exponent (LyE) after 10 km running decreased 0.28 bit/s compared with LyE after 7 km running ($p = 0.03$). In the tibia, LyE after 4 km and 10 km running was decreased significantly ($p < 0.01$ and $p = 0.01$). *Conclusions*: The foot was significantly pronated at the middle and at the end of running. Foot pronation during distance running increased the distal tibia peak impact acceleration but did not increase running instability.

Key words: *foot pronation, running, impact loading, tibial acceleration, local dynamic stability (LDS)*

1. Introduction

Pronation and supination are considered natural movements, stabilizing the foot and retaining flexibility during locomotion [39]. Foot pronation as the main movement of the foot during running improves adaptation to contact surfaces during the early stance phase and absorbs foot-ground contact loading [12], [15]. It is also a common topic regarding running-related injuries and running shoe design [1], [27]. The decreased foot flexibility in the subtalar and midtarsal joints leads to foot supination during the push-off phase. The period of pronation is prolonged for hyper-pronators, therefore, delaying the onset of supination in the gait cycle [28].

When the foot pronates, it is inward rotated on its subtalar joint axis [1]. Generally, pronation occurs during the first 40–50% of foot contact during locomotion [26]. Abnormal foot postures may alter joint loading and are risk factors for lower limb injuries. Excessive pronation has been typically seen as an overuse injury contributor during gait cycles [12], [24], [26], such as medial tibia stress syndrome and patellofemoral pain. Moreover, pronation after long-distance running may impact foot lever arm function and reduce ankle plantarflexion during late stance [33].

* Corresponding author: Yaodong Gu, Faculty of Sports Science, Ningbo University. No. 818, Fenghua Road, Jiangbei District, Ningbo, Zhejiang, China. 315211. E-mail: guyaodong@nbu.edu.cn
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Colapietro et al. [4] demonstrated that runners with chronic ankle instability (CAI) showed a decreased pronation excursion than healthy counter partners during running. It was also found that foot posture changed to a more pronated position after a half marathon [5]. Dos Santos et al. [7] found that increased running velocity increases foot pronation. Mizrahi et al. [22] revealed that fatigue after running was associated with increased acceleration in the tibia and increased the risk of stress fracture. However, Nielsen et al. [25] argued that novice runners with moderate foot pronation did not exhibit increased running-related injuries in a 1-year prospective cohort study. Foot pronation is initially aimed to increase the ground contact surface, but how it influences impact forces is lacking investigation.

The six-item foot posture index (FPI-6) is a validated method in a clinical setting for quantifying the static foot posture during full weight-bearing through assessing foot postural variation in the forefoot, midfoot and rearfoot in three cardinal body planes [32]. In past decades, the FPI-6 was typically utilized as a biomechanical tool for the static foot posture before statistical analysis [29]. For each item, a score is labeled from –2 to +2. Total FPI-6 score above +9 or below –2 may be regarded as abnormal or pathological foot posture [31]. According to Nielsen et al. [25], the FPI-6 more than +6 is defined as pronated foot, and less than –1 – as supinated.

Other common variables used to assess foot posture or movement around subtalar joints are the longitudinal arch angle, Achilles tendon angle and rearfoot angle [1], [2], [7], [28]. An Achilles tendon angle of 7–10° was defined as foot pronation, and above 10° – as foot hyper-pronation with higher injury risk [7]. Rabiei et al. [30] have evaluated three-dimensional foot pronation during running by extracting feature values utilizing principal component analysis. However, foot pronation evaluation during running is a challenging task, as it is a combined movement of dorsiflexion, forefoot abduction and calcaneal eversion [1], [30].

Local dynamic stability (LDS) estimates gait variability and rates the slight perturbations of gait over time [6]. It was introduced in the gait biomechanics field, initially evaluating gait instability of walking and risk of falling, and has gained growing attention for investigating running gait in recent years [8], [9], [13], [14]. Franks et al. [9] illustrated that running stability in different midsole thickness and stiffness conditions of running shoes did not present significant differences. However, trained runners showed increased running stability compared to novice and recreational runners [9], [13]. LDS could also be employed to estimate neuromuscular control when transitioning from one running condition to the other immediately. Ekizos et al. [8] revealed decreased running stability due to an instant transition from the shod to barefoot running condition. The running stability increases at the middle and end of distance-running via calculating the largest Lyapunov exponent (λ) from the lower limb angular velocities [13]. Nonetheless, Hollander et al. [14] found that running instability increased throughout a 15-minute run.

While the above-mentioned studies assessed running stability, no study has investigated how foot pronation during running affects the LDS of running gait. Furthermore, foot pronation is not an isolated factor influencing lower limb functions. Exploring gait variability and impact loading associated with the foot posture are crucial for understanding foot pronation-related injury mechanisms. Therefore, this study aimed to: (1) determine foot pronation changes during and after 10 km treadmill running by comparing the pronated angle, pressure beneath the foot arch, and peak resultant acceleration; and (2) estimate if running variability changes while the foot is pronated during running. It is hypothesized that: (1) plantar pressure, impact acceleration, and local dynamic stability would be different depending on the running distance and foot posture changes; (2) at the end of 10 km running, both acceleration on the foot and tibia would increase, and running stability may decrease at the middle and end of distance running.

2. Materials and methods

2.1. Participants

Twenty-five male heel strike recreational runners (height: 174.4 ± 5.6 cm; mass: 68.6 ± 8.2 kg; BMI: 22.6 ± 2.5 kg/m^2; age: 24.5 ± 3.0 years) were recruited through the university and local running clubs via posters and social media. Power analysis was performed in the SIMR package in R (Version 4.0.5, R Foundation for Statistical Computing, Vienna, Austria), and it was shown that the sample size in this study is enough to detect differences statistically (power >0.8). All participants were able to run a minimum running volume of 20 km/week and were free from lower limb musculoskeletal injuries and neural disorders. Subjects with pronated or supinated feet and flat feet or pes cavus were excluded. Participants signed written informed consent and were informed of the test procedures and study objectives. The study was conducted in accordance with the Declaration of Helsinki and approved by the University's Institutional Review Board (RAGH20201137).

2.2. Experiment design and protocol

Participants were given ten minutes to warm up and become familiar with the experimental settings. FPI-6 was initially measured for each participant before the test while standing on the floor, a shoulder-width between feet. As shown in Fig. 1a, four markers were placed at the heel and shank. Pressure under the foot arch (navicular) was recorded via the Pliance S2005 pressure measurement system (Novel GmbH, Munich, Germany; Range: 3–200 kPa), which has been proven to have high test–retest and inter–rater reliability (ICC ≥ 0.995) [18] (Fig. 1b). All participants wore the same running shoes during running, prototypes of neutral running shoes with ethylene-vinyl acetate (EVA) midsole and heel height of 33 mm.

Wearable sensors are often placed on the dorsal foot and medial distal tibia to assess lower limb impact acceleration. In this study, Nine-axial inertial measurement unit (IMU) sensors (IMeasureU, Auckland, New Zealand; mass: 12 g; range: ±16 g and ±2000 °/s; resolution: 16 bit; sample rate: 100 Hz) were attached using straps on the dorsum of the foot and the vertical axis of the distal anteromedial tibia, 3 cm away from the crest of the medial malleolus. Data were gathered in the IMeasureU Research (IMeasureU, Auckland, New Zealand) through Bluetooth connection using an iPad 2018 (Apple Inc. California, USA).

The modified Borg Rating of Perceive Exertion (RPE) was used to control the running intensity during running due to the difference between participants [41]. The average running velocity was 11.2 ± 1.2 km/h. After 10 km of running on a treadmill (Quasar, h/p cosmos®, GmbH, Germany), 3 minutes of rest were allowed, then, FPI-6 was rechecked. A single experienced practitioner conducted all experimental settings and foot posture checks.

2.3. Data collection and process

The Pliance pressure measurement system was used to measure and process peak force and pressure under the foot arch. Trial-axial acceleration and angular velocity data were recorded in the foot dorsum

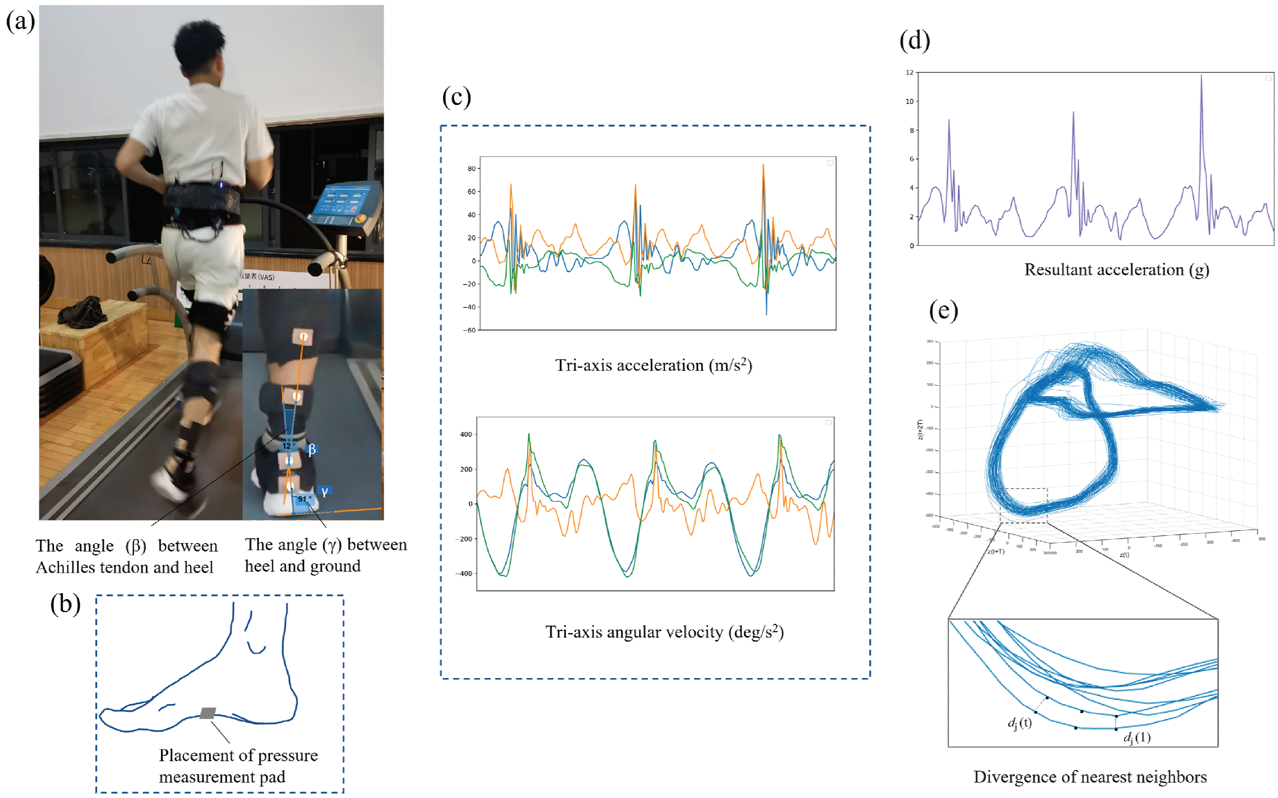

Fig. 1. (a) Experimental setup and β and γ angles, (b) placement of pressure pad for the peak force and pressure measurement under the foot arch, (c) trial-axial acceleration and angular velocity, (d) resultant acceleration, (e) evaluation of the local dynamic running stability

and distal tibia utilizing accelerometer and gyroscope in the IMU sensors (Fig. 1c). We used a Nikon D750 camera (NIKON Corp., Tokyo, Japan) to capture the trajectory of markers. The angle (β) between the Achilles tendon and heel and the angle (γ) between the heel and ground were calculated in Kinovea (Version: 0.8.27) (Fig. 1a). Pressure, sensor and marker trajectory were recorded in the right limb (dominant leg) for each runner after 1, 4, 7 and 10 km running. Pressure and video data were recorded for 3 seconds, and sensor data were collected for 80 seconds of each trial. Three consecutive running gaits were chosen for each trial to measure the difference in pressure and peak acceleration. Peak force and pressure were normalized by each participant's body mass for further statistical analysis [16].

2.4. Resultant acceleration and local dynamic stability

Raw acceleration was filtered by a second-order low-pass Butterworth filter with a cut-off frequency of 40 Hz and normalized to gravity (1 g = 9.8 m/s^2). Resultant acceleration (Fig. 1(d)) was calculated using the following equation.

$$\text{Resultant acceleration } = \sqrt{x^2 + y^2 + z^2}\,. \tag{1}$$

Eighty consecutive strides of angular velocity data in each trial were chosen for running stability evaluation. Gait events were detected based on the previously established method. Specifically, the initial foot contact was the local minima prior to the peak vertical acceleration in distal tibial [23]. We then time-normalized 80 gaits to 8000 samples using the data extrapolation method. λ quantifies the exponent divergence between neighboring kinematic trajectories (Fig. 1e) [34]. The short-time largest Lyapunov exponent (LyE) was used to assess the local dynamic stability of running gait. LyE represents the maximum rate of divergence of close trajectories in state-space [8], [9]. The first 5% of running gait was fit to assess LyE in this study using Rosenstein's algorithm [14]. The bigger the LyE, the worse the local dynamic stability. In this study, the state-space was reconstructed to calculate λ based on time-series data:

$$\begin{aligned} S(t) = [&a_i(t),\, a_i(t+\tau),\, a_i(t+2\tau), \\ &\ldots,\, a_i(t+(m-1)\tau)], \end{aligned} \tag{2}$$

in which τ is the time delay, m is the embedded dimension, a_i represents one-dimensional coordinate, i represents sensor axis, and $S(t)$ is the vector of the reconstructed m-dimensional state-space.

State-space reconstruction relies on the time delay and embedded dimension. The average mutual information algorithm [10] and global false nearest neighbour algorithm [17] were adopted separately to calculate the time delay of time-series data and embedded dimension. Time delay and maximum embedded dimension were fixed as 12 (mean across all trials) and 9 (maximum across all trials). Rosenstein's algorithm [34] was employed to calculate LyE as it is robust to experimental noise [20]. In the state-space, the algorithm tracks the Euclidean distance of the initial nearest neighbours of each point by the following equation.

$$d(t) = \| \Delta x(t) \| / \| \Delta x(0) \|\,. \tag{3}$$

To that, the relationship between the j^{th} pair of nearest neighbours exponential diverge and its distance could be expressed as below.

$$d_j(i) \approx C_j * e^{\lambda * (i\Delta\Delta)}\,, \tag{4}$$

where C_j is a constant representing the initial separation.

Therefore, LyE can be calculated as the slope of the logarithm of mean divergence by a linear fit of the divergence curves:

$$LyE = \lim_{n\to\infty} \lim_{\Delta x(0)\to 0} \frac{1}{\Delta t}\left[\ln\left(\frac{\Delta x(t)}{\Delta x(0)}\right)\right]. \tag{5}$$

2.5. Statistical analysis

Descriptive statistics are represented as Mean ± SD. Shapiro–Wilk normality test was performed to check the Gaussian distribution of data. Pre- and post- FPI-6 scores were analyzed using a pair sample t-test. Linear mixed-effects models were used to check the statistical difference of the data during running in the GraphPad Prism® (v8.0.2, San Diego, CA, USA). Runners were included as a random factor. Pressure, angles, resultant acceleration or LyE were included as a fixed effect, separately. If the results reject Mauchly's test of Sphericity, Greenhouse–Geisser was applied to adjust the degree of freedom. Partial eta squared value (η_p^2) was calculated to quantify the magnitude of effect size and classified as small (0.01 < ES ≤ 0.06), medium (0.06 < ES ≤ 0.14), and large (ES > 0.14) [3]. Tukey's honest significance differences (HSD) post hoc analysis was conducted for significant difference tests between groups. The significance level was set at $\alpha < 0.05$.

3. Results

3.1. Foot posture evaluation

FPI-6 has significantly increased after the 10 km running (2.1 ± 1.6 vs. 6.2 ± 1.7, $p < 0.01$, 95%CI: 3.5 to 4.7). As shown in Fig. 2, β angle was significantly changed ($p = 0.01$, $\eta_p^2 = 0.22$): compared to β angle after 1 km (13.0°), it increased to 14.7° after 10 km ($p = 0.01$, 95%CI: –2.9 to –0.4). γ angle was not significantly changed and with no differences between groups.

3.2. Plantar loading characteristics

Peak force during the 10 km treadmill running is significantly increased ($F = 9.35$, $p < 0.01$) (Fig. 3a, Table 1). Peak force reached 0.111 ± 0.042 N, compared to 0.092 ± 0.036 N after 1 km of running, 0.093 ± 0.042 N after 4 km of running, and 0.105 ± 0.042 N after 7 km of running, with $p = 0.01$, $p = 0.01$, and $p = 0.04$, respectively. As presented in Table 1, the peak pressure increased gradually ($F = 20.19$, $p < 0.01$, $\eta_p^2 = 0.21$) (Fig. 3b) and a significant difference exists in every two-group comparison.

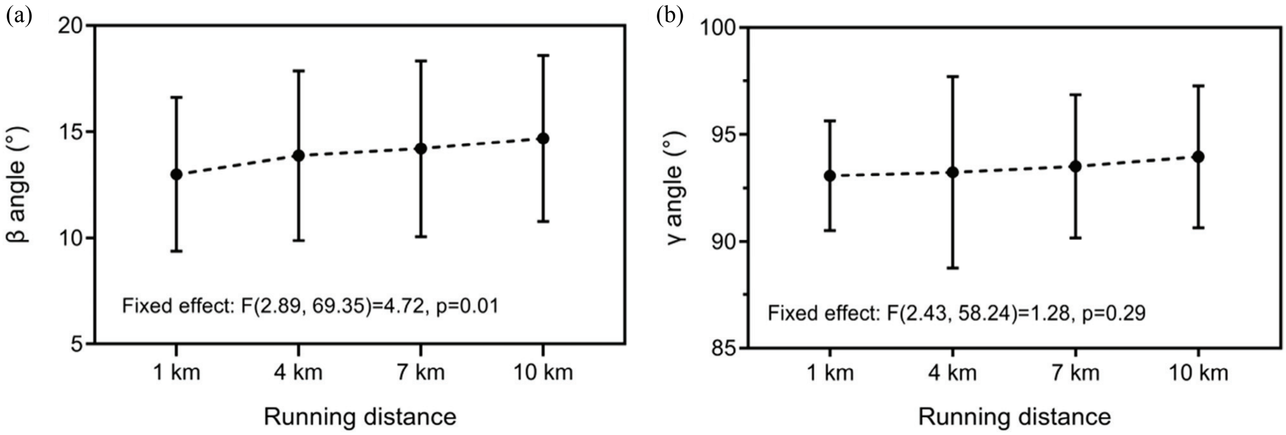

Fig. 2. The angle (β) between the Achilles tendon and the heel (a), and the angle (γ) between the heel and ground (b) during the 10 km running

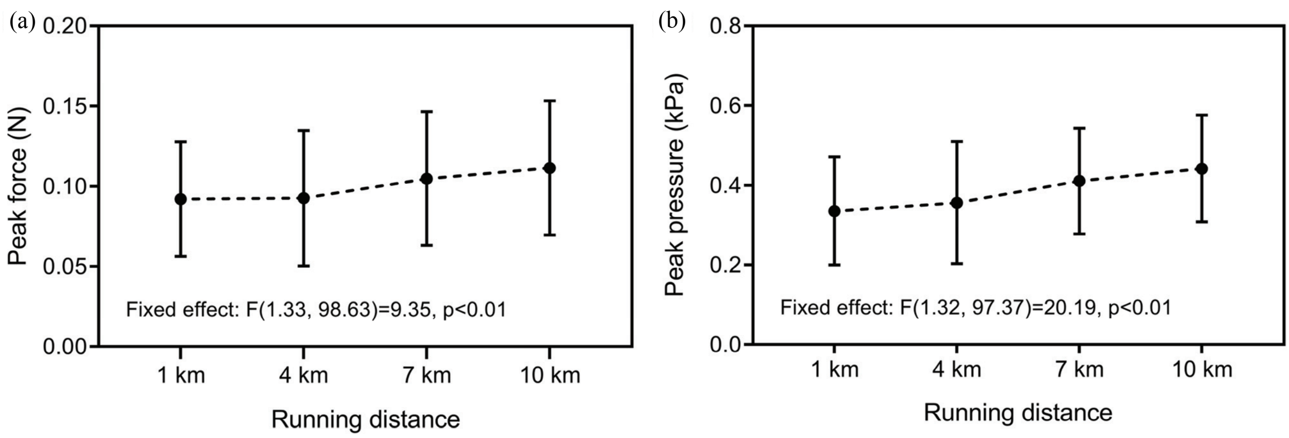

Fig. 3. Peak force (a) and pressure (b) under the foot arch during the 10 km running

3.3. Shock acceleration changes

As depicted in Table 2, there is no statistical difference for the resultant acceleration in the dorsum of foot with F (2.7, 199.8) = 1.34 and p = 0.26. For the distal tibial acceleration, the peak value after 10 km of running was significantly increased, compared to the corresponding value after 4 km of running (mean difference = –0.66, 95% CI: –1.24 to –0.09, p = 0.02) (Fig. 4b).

Table 1. Mixed-effect and post hoc between groups analysis for the peak force and pressure under the foot arch (Mean ± SD)

	Comparisons between groups	Tukey's HSD *post hoc* analysis			Mixed-effect analysis		
		Mean ± SD	Mean difference (95% CI)	P	F	η_p^2	P
Peak force [N]	1 km vs. 4 km	0.092 ± 0.036 vs. 0.093 ± 0.042	–0.001(–0.005, 0.004)	0.99	9.35 (1.33, 98.63)	0.11	**<0.01**
	1 km vs. 7 km	0.092 ± 0.036 vs. 0.105 ± 0.042	–0.013(–0.026, 0)	0.06			
	1 km vs. 10 km	0.092 ± 0.036 vs. 0.111 ± 0.042	–0.019(–0.034, –0.005)	**0.01**			
	4 km vs. 7 km	0.093 ± 0.042 vs. 0.105 ± 0.042	–0.012(–0.025, 0)	0.05			
	4 km vs. 10 km	0.093 ± 0.042 vs. 0.111 ± 0.042	–0.019(–0.033, –0.004)	**0.01**			
	7 km vs. 10 km	0.105 ± 0.042 vs. 0.111 ± 0.042	–0.007(–0.013, 0)	**0.04**			
Peak pressure [kPa]	1 km vs. 4 km	0.336 ± 0.136 vs. 0.356 ± 0.153	–0.02(–0.034, –0.007)	**<0.01**	20.19 (1.32, 97.37)	0.21	**<0.01**
	1 km vs. 7 km	0.336 ± 0.136 vs. 0.411 ± 0.133	–0.075(–0.12, –0.031)	**<0.01**			
	1 km vs. 10 km	0.336 ± 0.136 vs. 0.442 ± 0.134	–0.106(–0.156, –0.057)	**<0.01**			
	4 km vs. 7 km	0.356 ± 0.153 vs. 0.411 ± 0.133	–0.055(–0.099, –0.01)	**0.01**			
	4 km vs. 10 km	0.356 ± 0.153 vs. 0.442 ± 0.134	–0.086(–0.137, –0.034)	**<0.01**			
	7 km vs. 10 km	0.411 ± 0.133 vs. 0.442 ± 0.134	–0.031(–0.054, –0.008)	**<0.01**			

Table 2. Mixed effect analysis for the resultant acceleration and LyE in the dorsum of foot and distal tibia regions during 10 km running (Mean ± SD)

		1 km	4 km	7 km	10 km	F	η_p^2	P
Acceleration [g]	Foot	12.39 ± 3.36	13.00 ± 2.91	12.75 ± 3.16	13.10 ± 3.35	1.34 (2.7,199.8)	0.02	0.26
	Tibia	8.22 ± 2.82	8.02 ± 1.96	8.28 ± 2.76	8.69 ± 2.37	2.36 (2.51, 185.3)	0.04	0.08
LyE [bit/s]	Foot	10.67 ± 1.94	10.52 ± 1.81	10.75 ± 1.71	10.47 ± 1.75	3.19 (2.64, 195.4)	0.04	**0.03**
	Tibia	8.42 ± 2.17	8.12 ± 2.03	8.18 ± 2.20	7.97 ± 2.25	5.55 (2.3, 171.1)	0.07	**<0.01**

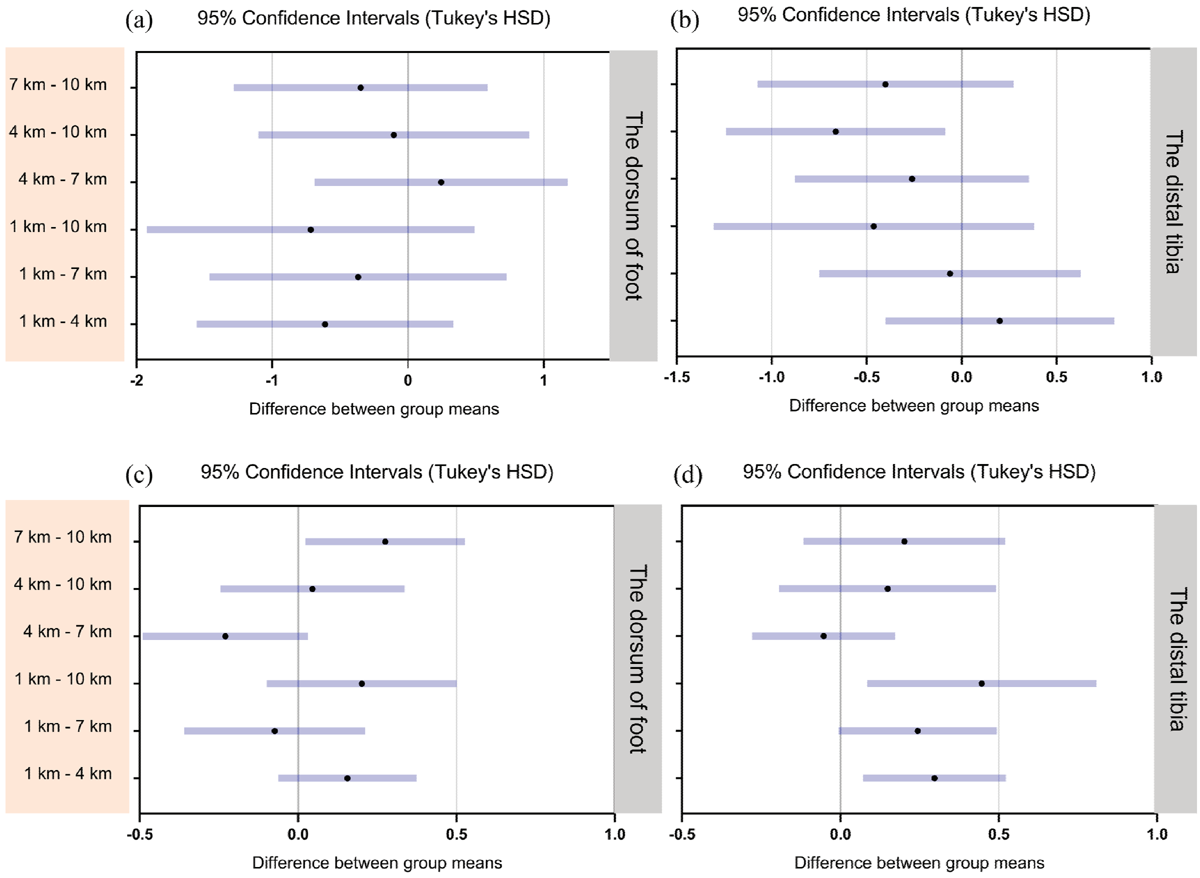

Fig. 4. *Post-hoc* analysis of groupwise comparisons for peak resultant acceleration (a and b) and largest Lyapunov exponent (c and d) of running after 1, 4, 7 and 10 km

3.4. Local dynamic stability assessment

The mixed-effect model exhibits statistical differences of LyE based on the angular velocity data collected at the foot (F = 3.19, p = 0.03, η_p^2 = 0.04) and distal tibia (F = 5.55, p < 0.01, η_p^2 = 0.07) (Fig. 4c, 4d). At the dorsum of foot, LyE after 10 km of running decreased 0.28 bit/s compared with LyE after 7 km of running (mean difference = 0.28, 95%CI: 0.02 to 0.53, p = 0.03). LyE in the tibia showed that after 4 km and 10 km run, it decreased to 8.12 ± 2.03 bit/s (mean difference = 0.30, 95% CI: 0.07 to 0.52, p < 0.01) and 7.97 ± 2.25 bit/s (mean difference = 0.45, 95% CI: 0.08 to 0.81, p = 0.01), respectively, compared to the value after 1 km of running (8.42 ± 2.17 bit/s).

4. Discussion

This study explored foot pronation by employing pressure measurement, marker trajectory monitoring, and wearable sensor technologies during 1, 4, 7 and 10 km of running. The LDS of running was further compared by calculating LyE utilizing the Rosenstein algorithm. The findings showed that the FPI-6 score increased to a significant pronation condition and the extent of pronation was increased step by step. Peak acceleration and running stability also increased, although the foot is more pronated in the middle and end of 10 km running.

Foot pronation is hard to assess during running since the movement around the subtalar joint cannot be evaluated directly when performing dynamic tasks [1]. Combining multiple measurements to assess foot movement is not recommended, as each sub-measurement contains independent information [2]. This study initially evaluated foot pronation via FPI-6, and β and γ angles were monitored during treadmill running. Gradually increased pressure at the region under the foot arch demonstrated that the foot is more pronated during distance-running. During the loading response phase of running, the dorsiflexion at the ankle joint and pronation at the subtalar joint were normal phenomena to increase the ground contact surface area and absorb shock [12], [15]. Increased pronation during the stance phase may reduce ankle plantarflexion [33]. Consistent with kinematic changes of foot pronation, this study found significantly increased peak force and pressure underneath the foot arch at the middle and end of 10 km running.

Malisoux et al. [19] found that runners with pronated feet may reduce overall injury rates when wearing motion control shoes. However, Nielsen et al. [25] illustrated that novice runners with moderate foot pronation taking up running in neutral running shoes did not increase the risk of injuries. Mei et al. [21] found pronated feet after distance running increased knee and ankle joint moments. However, the implications of foot pronation in distance running are still not well-explored. We could not directly correlate running-related injuries with foot pronation. Therefore, this is still a knowledge gap associated with lower limb impact loading and gait variability when explaining the mechanisms of foot pronation.

The load rate of vertical ground reaction force (GRF) has been associated with a variety of running-related injuries. Tibial acceleration is strongly correlated with vertical GRF [40], [44]. Schutte et al. [35] found that runners with a history of medial tibial stress syndrome performed higher peak acceleration than the control group. Sheerin et al. [36] demonstrated a moderate correlation between running velocity and tibial acceleration, and peak resultant acceleration increase with higher velocities. Apart from increased peak pressure data, peak acceleration in the distal tibia was also increased. These findings present an increased risk of overload injuries in the shank during the middle to later phase of distance running after foot pronation. Therefore, the foot pronation effect was unable to mitigate plantar loading and shock acceleration increments. Consistent with a previous study [43], which proved that maximalist shoes have a better shock attenuation function in both time and frequency domains than minimalist shoes, inexperienced runners or runners with stress fractures take cushioning into consideration while choosing running shoes.

We also found that the impact acceleration in the dorsum of the foot is higher than in the tibia. This indicates that the distal lower limb bears a higher impact and may be explained by the effect of ankle joint motion on impact during running [11]. However, besides pronated feet during running, fatigue at the end of running may contribute to increased lower limb impact acceleration, which is not considered in this study [22]. Footwear selections may also influence shock acceleration; for instance, the minimalist shoe increased peak acceleration, whereas acceleration decreased in the ultra-cushioning shoe [37].

As for running gait, λ in the first 0.05 of gait cycle period is sensitive [14]. In this study, LyE was defined as the slope of a linear fit of 0–5% time-normalized samples rather than typical 0–50% walking gait cycle [38]. It was hypothesized that the pronated foot at the end of 10 km running changed the lower limb kinematics and kinetics and may decrease the LDS during running. Inconsistently with our expectation, running stability of the lower limb was increased in this study, illustrating small perturbations generated from gait movement of increased pronation of the foot did not contribute to increased gait variability of running. Furthermore, these findings are consistent with previous studies [9], [13] that local dynamic running instability decreased during running overtime. The LDS was higher in the tibia than the foot, indicating decreased gait stability in distal joints, which is consistent with previous studies [9], [13]. Therefore, foot pronation during distance running may not alter neuromuscular control to the lower limb.

Despite the promising findings, one limitation in this study was that we did not consider fatigue with increased running mileage. Foot pronation after running may associate with motor and neuromuscular control to some extent. Future studies may investigate foot posture changes after distance-running and fatigue interaction and explain how foot pronation affects the lower limb biomechanics and running stability independently. Furthermore, this study focalized the shock acceleration in the time domain. How frequency characteristics (i.e., power spectral density and shock attenuation magnitude) are changed with the foot posture during a prolonged run is worth exploring in the future.

5. Conclusions

Gait variability and impact acceleration characteristics during prolonged running were investigated along with the foot posture. The foot was significantly pronated at the middle and end of running. Foot pronation during distance running increased impact acceleration at the distal tibia but did not increase running instability. These findings suggested that novice runners or recreational runners with a history of tibial stress fractures wear footwear with an impact load absorption function during prolonged running because impact forces increase and foot shape changes.

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References

[1] BEHLING A.V., MANZ S., VON TSCHARNER V., NIGG B.M., *Pronation or foot movement – what is important*, J. Sci. Med. Sport, 2020, 23 (4), 366–371.

[2] BEHLING A.V., NIGG B.M., *Relationships between the foot posture index and static as well as dynamic rear foot and arch variables*, J. Biomech., 2020, 98, 109448.

[3] COHEN J., *Statistical Power Analysis for the Behavioral Sciences*, 2nd ed., Routledge, 2013.

[4] COLAPIETRO M., FRASER J.J., RESCH J.E., HERTEL J., *Running mechanics during 1600 meter track runs in young adults with and without chronic ankle instability*, Phys. Ther. Sport, 2020, 42, 16–25.

[5] COWLEY E., MARSDEN J., *The effects of prolonged running on foot posture: a repeated measures study of half marathon runners using the foot posture index and navicular height*, J. Foot Ankle Res., 2013, 6 (20), DOI: 10.1186/1757-1146-6-20.

[6] DINGWELL J.B., CUSUMANO J.P., CAVANAGH P.R., STERNAD D., *Local dynamic stability versus kinematic variability of continuous overground and treadmill walking*, J. Biomech. Eng., 2001, 123, 27–32.

[7] DOS SANTOS J.O.L., GOMES A.L.R., LIMA A.B., DE PAIVA VIEIRA E., BEZERRA E.D.S., ROSSATO M., *Effect of linear running velocity on the increase on foot pronation*, Foot, 2019, 41, 74–78.

[8] EKIZOS A., SANTUZ A., ARAMPATZIS A., *Transition from shod to barefoot alters dynamic stability during running*, Gait Posture, 2017, 56, 31–36.

[9] FRANK N.S., PRENTICE S.D., CALLAGHAN J.P., *Local dynamic stability of the lower extremity in novice and trained runners while running in traditional and minimal footwear*, Gait Posture, 2019, 68, 50–54.

[10] FRASER A.M., SWINNEY H.L., *Independent coordinates for strange attractors from mutual information*, Phys. Rev. A., 1986, 33, 1137–1140.

[11] GLASSBROOK D.J., FULLER J.T., ALDERSON J.A., DOYLE T.L.A., *Foot accelerations are larger than tibia accelerations during sprinting when measured with inertial measurement units*, J. Sports Sci., 2020, 38 (3), 248–255.

[12] HINTERMANN B., NIGG B.M., *Pronation in runners*, Sport. Med., 1998, 26 (3), 169–176.

[13] HOENIG T., HAMACHER D., BRAUMANN K.M., ZECH A., HOLLANDER K., *Analysis of running stability during 5000 m running*, Eur. J. Sport Sci., 2019, 19 (4), 413–421.

[14] HOLLANDER K., HAMACHER D., ZECH A., *Running barefoot leads to lower running stability compared to shod running – results from a randomized controlled study*, Sci. Rep., 2021, 11, 4376.

[15] HORWOOD A.M., CHOCKALINGAM N., *Defining excessive, over, or hyper-pronation: a quandary*, Foot, 2017, 31, 49–55.

[16] JANDOVÁ S., VOLF P., VAVERKA F., *The influence of minimalist and conventional sports shoes and lower limbs dominance on running gait*, Acta Bioeng. Biomech., 2018, 20 (3), 3–9.

[17] KENNEL M.B., BROWN R., ABARBANEL H.D.I., *Determining embedding dimension for phase-space reconstruction using a geometrical construction*, Phys. Rev. A., 1992, 45 (6), 3403–3411.

[18] LAI C.H.Y., LI-TSANG C.W.P., *Validation of the Pliance X System in measuring interface pressure generated by pressure garment*, Burns, 2009, 35, 845–851.

[19] MALISOUX L., CHAMBON N., DELATTRE N., GUEGUEN N., URHAUSEN, A., THEISEN D., *Injury risk in runners using standard or motion control shoes: a randomised controlled trial with participant and assessor blinding*, Br. J. Sports Med., 2016, 50 (8), 481–487.

[20] MEHDIZADEH S., *The largest lyapunov exponent of gait in young and elderly individuals: a systematic review*, Gait Posture, 2018, 60, 241–250.

[21] MEI Q., GU Y., XIANG L., BAKER J.S., FERNANDEZ J., *Foot pronation contributes to altered lower extremity loading after long distance running*, Front. Physiol., 2019, 10, 573.

[22] MIZRAHI J., VERBITSKY O., ISAKOV E., *Fatigue-related loading imbalance on the shank in running: a possible factor in stress fractures*, Ann. Biomed. Eng., 2000, 28 (4), 463–469.

[23] MO S., CHOW D.H.K., *Accuracy of three methods in gait event detection during overground running*, Gait Posture, 2018, 59, 93–98.

[24] NEAL B.S., GRIFFITHS I.B., DOWLING G.J., MURLEY G.S., MUNTEANU S.E., FRANETTOVICH SMITH M.M., COLLINS N.J., BARTON C.J., *Foot posture as a risk factor for lower limb overuse injury: a systematic review and meta-analysis*, J. Foot Ankle Res., 2014, 7(55), DOI: 10.1186/s13047-014-0053-6.

[25] NIELSEN R.O., BUIST I., PARNER E.T., NOHR E.A., SØRENSEN H., LIND M., RASMUSSEN S., *Foot pronation is not associated with increased injury risk in novice runners wearing a neutral shoe: a 1-year prospective cohort study*, Br. J. Sports Med., 2014, 48, 440–447.

[26] NIGG B., BEHLING A., HAMILL J., *Foot pronation*, Footwear Sci., 2019, 11 (3), 131–134.

[27] NIGG B.M., BALTICH J., HOERZER S., ENDERS H., *Running shoes and running injuries: mythbusting and a proposal for two new paradigms:"preferred movement path"and "comfort filter"*, Br. J. Sports Med., 2015, 49 (20), 1290–1294.

[28] NOVACHECK T.F., *The biomechanics of running*, Gait Posture, 1998, 7, 77–95.

[29] PATERSON K.L., CLARK R.A., MULLINS A., BRYANT A.L., MENTIPLAY B.F., *Predicting dynamic foot function from static foot posture: comparison between visual assessment, motion analysis, and a commercially available depth camera*, J. Orthop. Sport. Phys. Ther., 2015, 45 (10), 789–798.

[30] RABIEI M., ESLAMI M., MOVAGHAR A.F., *The assessment of three-dimensional foot pronation using a principal component analysis method in the stance phase of running*, Foot, 2016, 29, 11–17.

[31] REDMOND A.C., CRANE Y.Z., MENZ H.B., *Normative values for the foot posture index*, J. Foot Ankle Res., 2008, 1 (1), DOI: 10.1186/1757-1146-1-6.

[32] REDMOND A.C., CROSBIE J., OUVRIER R.A., *Development and validation of a novel rating system for scoring standing foot posture: the foot posture index*, Clin. Biomech., 2006, 21, 89–98.

[33] RESENDE R.A., PINHEIRO L.S.P., OCARINO J.M., *Effects of foot pronation on the lower limb sagittal plane biomechanics during gait*, Gait Posture, 2019, 68, 130–135.

[34] ROSENSTEIN M.T., COLLINS J.J., DE LUCA C.J., *A practical method for calculating largest lyapunov exponents from small data sets*, Phys. D Nonlinear Phenom., 1993, 65, 117–134.

[35] SCHÜTTE K.H., SEERDEN S., VENTER R., VANWANSEELE B., *Influence of outdoor running fatigue and medial tibial stress syndrome on accelerometer-based loading and stability*, Gait Posture, 2018, 59, 222–228.

[36] SHEERIN K.R., BESIER T.F., REID D., *The influence of running velocity on resultant tibial acceleration in runners*, Sport. Biomech., 2018, DOI: 10.1080/14763141.2018.1546890.

[37] SINCLAIR J., FAU-GOODWIN J., RICHARDS J., SHORE H., *The influence of minimalist and maximalist footwear on patellofemoral kinetics during running*, J. Appl. Biomech., 2016, 32 (4), 359–364.

[38] STENUM J., BRUIJN S.M., JENSEN B.R., *The effect of walking speed on local dynamic stability is sensitive to calculation methods*, J. Biomech., 2014, 47, 3776–3779.

[39] TANSEY P.A., BRIGGS P.J., *Active and passive mechanisms in the control of heel supination*, Foot ankle Surg., 2001, 7 (3), 131–136.

[40] TENFORDE A.S., HAYANO T., JAMISON S.T., OUTERLEYS J., DAVIS I.S., *Tibial acceleration measured from wearable sensors is associated with loading rates in injured runners*, PM R, 2020, 12 (7), 679–684.

[41] VARGHESE M.A., SAHA P.N., ATREYA N., *A rapid appraisal of occupational workload from a modified scale of perceived exertion*, Ergonomics, 1994, 37 (3), 485–491.

[42] XIANG L., GU Y., RONG M., GAO Z., YANG T., WANG A., SHIM V., FERNANDEZ J., *Shock acceleration and attenuation during running with minimalist and maximalist shoes: a time- and frequency-domain analysis of tibial acceleration*, Bioengineering, 2022, 9, 322.

[43] XIANG L., WANG A., GU Y., ZHAO L., SHIM V., FERNANDEZ J., *Recent machine learning progress in lower limb running biomechanics with wearable technology: a systematic review*, Front. Neurorobot., 2022, 16, 913052.