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Lower extremity stiffness in habitual forefoot strikers during running on different overground surfaces

ZIWEI ZENG[1], LULU YIN[2], WENXING ZHOU[1], YU ZHANG[2], JIAYI JIANG[1], LIN WANG[3]*

[1] School of Kinesiology, Shanghai University of Sport, Shanghai, China.
[2] Department of Critical Care Medicine, Shanghai Tenth People's Hospital, School of Medicine, Tongji University, Shanghai, China.
[3] Department of Sports Rehabilitation, Shanghai University of Sport, Shanghai, China.

Purpose: Sports surface is one of the known external factors affecting running performance and injury. To date, we have found no study that examined the lower extremity stiffness in habitual forefoot strikers running on different overground surfaces. Therefore, the objective of this study was to investigate lower extremity stiffness and relevant kinematic adjustments in habitual forefoot strikers while running on different surfaces. *Methods*: Thirty-one male habitual forefoot strikers were recruited in this study. Runners were instructed to run at a speed of 3.3 m/s (±5%) on three surfaces, named synthetic rubber, concrete, and artificial grass. *Results*: No significant differences were found in leg stiffness, vertical stiffness, and joint stiffness in the sagittal plane during running on the three surfaces ($p > 0.05$). Running on artificial grass exerted a greater displacement in knee joint angle than running on synthetic rubber ($p = 0.002$, 95% CI = 1.52–7.35 degrees) and concrete ($p = 0.006$, 95% CI = 1.04–7.25 degrees). In the sagittal plane, peak knee moment was lower on concrete than on artificial grass ($p = 0.003$, 95% CI = 0.11–0.58 Nm/kg), whereas peak ankle moment was lower on synthetic rubber than on concrete ($p < 0.001$, 95% CI = 0.03–0.07 Nm/kg) and on artificial grass ($p < 0.001$, 95% CI = 0.02–0.06 Nm/kg). Among the three surfaces, the maximal ground reaction forces on concrete were the lowest ($p < 0.05$). *Conclusions*: This study indicated that running surfaces cannot influence lower extremity stiffness in habitual forefoot strikers at current running speed. Kinematic adjustments of knee and ankle, as well as ground reaction forces, may contribute to maintaining similar lower extremity stiffness.

Key words: *running, kinematics, forefoot strike pattern*

1. Introduction

Running is an easily attainable sport to achieve physical fitness and mental health [15], [24], however, it frequently causes injuries. The aetiology of running injuries is multifactorial and can result from the interactions of various extrinsic and intrinsic factors, such as running surface, footwear, and foot strike pattern [14], [27]. Several studies suggested that overground surfaces are one of the extrinsic risk factors for running injuries, especially for lower extremity injuries [1], [8], [26].

Experimental studies demonstrated that running surfaces can influence modifiable lower extremity stiffness among habitual rearfoot strikers (hRFS) [7], [9], [10], [20]. The centre of mass (COM) trajectory is not affected by rapidly increasing leg stiffness (K_{leg}) when runners abruptly transfer onto a softer running surface [9], [10], [20]. However, impact forces are strongly affected when running on different overground surfaces [21]. Several studies concerning foot strike patterns when running on a hard surface [7], [31] indicated that running on asphalt requires a stiffer ankle joint, whereas running on gravel requires a greater stability in ankle joint [7]. An increase in surface hardness reportedly results in decreased hip and knee flexion at contact [16]. However, a study demonstrated that indices of running performance, including sprint time, velocity, and duty factor, are not affected by different running surface hardness nor kinematics and kinetics of ankle or knee joint [34]. As mentioned above, the biome-

* Corresponding author: Lin Wang, Department of Sports Rehabilitation, Shanghai University of Sport, Shanghai, China.
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chanical effects of running on different overground surfaces on lower extremity stiffness among hRFS are controversial.

Stiffness is often regarded as a reaction of the neuromuscular system, that is, the neuromuscular system adjusts stiffness to complete motor response in shifting conditions [22], [23]. Runners modify their lower extremity stiffness to accommodate changes in running conditions, allowing themselves to maintain similar running mechanics and stability on different overground surfaces [11], [22]. Furthermore, increased stiffness may be related to pressure injuries due to overuse because of greater loads, whereas decreased stiffness may be associated with soft tissue injuries due to excessive joint motion in lower extremities [4].

Foot strike pattern has been verified to be a vital factor related to running injuries [37] and appears to be associated with lower extremity stiffness [28]. Prior studies reported a reduction in running injuries in forefoot strikers (FFS) compared to RFS [5]. Most studies explored the stiffness of hRFS [6], [7], [16] or groups of runners whose foot strike patterns were not identified [8], [10], [11], [20], [26]. A recent study indicated that hFFS and hRFS experience different lower extremity stiffness [39]. And hFFS have a greater K_{leg}, a shorter ground contact time, and a lower loading rate than hRFS [39]. Therefore, the findings on the lower extremity stiffness in hRFS running on different overground surfaces may not be applicable to hFFS. Hence, the interaction between lower extremity stiffness in hFFS and different overground surfaces must be investigated.

The purpose of the current study was to determine the effects of three sports surfaces on the lower extremity stiffness in hFFS. We hypothesized that hFFS can adjust lower extremity stiffness according to three natural running surfaces. This study could provide useful information on the influence of sports surfaces on running biomechanics and economy among hFFS.

2. Materials and methods

2.1. Participants

Thirty-one healthy male hFFS (age = 28.7 ± 7.1 years; height = 173.7 ± 4.5 cm; weight = 67.6 ± 10.2 kg; leg length = 0.8 ± 0.04 m; running age = 4.4 ± 3.9 years; running distance per week = 33.9 ± 24.6 km) were recruited from the local running clubs. Upon arrival at the laboratory, the participants completed a verbal questionnaire to ensure that they had no known abnormalities (i.e., flat feet and high arched feet), cardiovascular diseases, visual diseases, vestibular diseases, or recent musculoskeletal injuries that could affect their running performance (within the past 6 months). This study was approved by the Ethics Committee of Shanghai University of Sport. Written informed consent was obtained from each participant before data collection.

Foot strike patterns were determined using the Novel Pedar-X system (Novel Munich, Germany) to identify the centre of pressure (CoP) before the formal test. The sampling frequency was 100 Hz. The participants were instructed to run at a target velocity of 3.3 m/s (±5%) in their habitual landing fashions [6]. The speed was determined using a photoelectric timing system (WittySEM, Microgate, Italy), which was placed in the middle of the 15 m runway. At the point of initial foot landing, the participants whose CoP was located in the anterior third of the foot length were identified as hFFS, whereas those whose CoP was located in the middle third or the posterior third were regarded as hMFS or hRFS, respectively [18]. In this study, only hFFS were included.

Three running surfaces were constructed in the laboratory, namely, synthetic rubber, concrete, and artificial grass. Each type of constructed runway was 15 m long and 1 m wide. The thickness of synthetic rubber and artificial grass was 2 cm. The concrete runway was constructed using concrete tiles with a thickness of 1 cm. Under each kind of surface, a layer of polyvinyl chloride mattress (1.6 mm) was inserted to prevent friction between the running surfaces and the floor [40].

2.2. Experimental protocol

Experiments were conducted at the biomechanics laboratory in Shanghai University of Sport. First, the participants' body height, body mass, and leg length were measured. All participants were right-leg dominant, which was confirmed as one's preferred leg to kick a ball [12]. All participants were then required to wear the same lightweight running shoes (European size 41 to 43, ASICS, SORTIEMAGIC RP 4 TMM467.0790, Japan) without thick heels and additional cushioning structures to ensure no abnormal feeling and interference with running performance [3].

Before the formal test, each participant was instructed to stretch first and then ran on a treadmill for 5 min to warm up at their preferred speeds. After the warm up period, the participants were allowed to perform practice running trials to familiarize themselves with the experimental procedure and adjust their running speed to the prescribed velocity (3.33 m/s, ±5%). The order of three different running surfaces was randomized for each participant. The participants were instructed to make contact with the central portion of the two force plates, which were embedded under the runway, with their right foot without deliberately modifying their running gait. Each participant completed three successful trials for each running surface.

Kinematic and ground reaction force (GRF) data were recorded using a 3D motion analysis with 10 cameras (Vicon T40, Oxford Metrics, UK) and synchronized with two force plates (Kistler Instruments Corp., Switzerland). The force plates were placed in the middle part of the 15 m runway. The acquisition sampling frequency was 200 and 2000 Hz, respectively. Twenty-three reflective markers were placed on the participants' anatomical landmarks according to the full-body plug-in-gait model [25].

2.3. Data analysis

Data were processed in Visual 3D (C-Motion, Rockville, MD, USA). A pelvis and a three-segment lower extremity plug-in-gait model (i.e., thigh, shank, and foot) was constructed to determine related lower extremity kinematic and kinetic features. All data were filtered using a fourth-order low-pass Butterworth filter with a cut-off frequency (kinematics: 10 Hz; GRFs: 50 Hz). Kinematic and kinetic features were analysed throughout the running stance phase. The start of ground contact was defined as the moment at which the vertical ground reaction force (vGRF) became greater than 20 N, whereas the end of ground contact was defined as the moment after the initial ground contact at which the vGRF became lower than 20 N.

Leg stiffness, K_{leg} [kN/m], was calculated as the ratio of the maximal GRF to ΔL during the running stance phase as follows [29], [32]:

$$K_{leg} = \frac{\mathrm{GRF}_{\max}}{\Delta L}, \tag{1}$$

$$\Delta L = L_0 - \sqrt{L_0^2 - \left(\frac{Vt_c}{2}\right)^2} + \Delta y, \tag{2}$$

where the ΔL is the vertical variation in leg length, L_0 is the upright leg length measured as the distance from the trochanter major to the prominence of the lateral malleolus, Δy is the vertical displacement of the COM, and t_c is the ground contact time.

Vertical stiffness, K_{vert} [kN/m], was calculated as the ratio of the maximal GRF to Δy during the running stance phase as follows [4], [10]:

$$K_{vert} = \frac{\mathrm{GRF}_{\max}}{\Delta y}. \tag{3}$$

Joint stiffness, K_{joint} [Nm/kg/degree], was calculated as the ratio of total change in joint moment (ΔM_{joint}) to joint angular displacement ($\Delta\theta_{joint}$) as follows [30]:

$$K_{joint} = \frac{\Delta M_{joint}}{\Delta\theta_{joint}}, \tag{4}$$

where ΔM_{joint} is the displacement of joint moment between initial ground contact and the instant of peak joint moment, and $\Delta\theta_{joint}$ is the angular displacement in the sagittal plane between the initial ground contact and the instant of peak joint moment.

Initial ground contact angle was defined as the angle when vGRF exceeded 20 N. The displacement of joint angle/moment was defined as the total change in angle/moment during the running stance phase. Peak joint moment was defined as the maximal joint moment in the sagittal plane during the running stance phase. The maximal GRF was recorded in three dimensions, namely, anterior-posterior, medial-lateral and vertical directions. All joint moments and GRFs were normalized to body weight (BW).

2.4. Statistical analysis

All data are expressed as mean ± SD. One-way variance analysis of repeated measures (ANOVA) was applied to determine differences in lower extremity stiffness, kinematic, and kinetic parameters among different running surfaces. Bonferroni adjustment was adopted for multiple comparisons. If a significant difference was found between two surfaces, the 95% confidence interval of mean difference (95% CI) was calculated between the two surfaces. The level of statistical significance was determined as $p < 0.05$. Statistical analysis was performed using SPSS version 26.0 (version 26.0, IBM Inc., Chicago, IL, USA).

3. Results

No significant differences were observed in K_{leg}, K_{vert}, or K_{joint} in the sagittal plane during running on the three different surfaces (Table 1).

Table 1. Lower extremity stiffness, kinematic and kinetic variables when running on three overground surfaces (mean ± SD)

Variables	Synthetic rubber	Concrete	Artificial grass	p value
K_{leg} [KN/m]	10.0 ± 2.0	9.5 ± 2.1	9.7 ± 1.8	0.147
K_{vert} [KN/m]	21.3 ± 3.2	21.0 ± 3.3	21.1 ± 3.2	0.735
Knee K_{joint} [Nm/kg/degree]	0.04 ± 0.01	0.04 ± 0.01	0.04 ± 0.01	0.288
Ankle K_{joint} [Nm/kg/degree]	0.06 ± 0.02	0.06 ± 0.02	0.06 ± 0.01	0.286
t_c [s]	0.23 ± 0.02	0.24 ± 0.02	0.23 ± 0.02	0.362
Δy [m]	0.09 ± 0.01	0.08 ± 0.01	0.08 ± 0.01	0.160
Knee $angle_0$ [degree]	–19.4 ± 5.3	–19.3 ± 4.9	–20.2 ± 7.1	0.563
Ankle $angle_0$ [degree]	–14.6 ± 6.5	–14.8 ± 6.4	–15.3 ± 6.8	0.705
Knee Δangle [degree]	84.5 ± 9.4	84.8 ± 11.1	88.9 ± 11.5	**<0.001**
Ankle Δangle [degree]	58.5 ± 27.2	53.6 ± 5.6	52.4 ± 5.5	0.292
Peak knee moment [Nm/kg]	2.01 ± 0.11	1.79 ± 0.09	2.13 ± 0.10	**0.001**
Peak ankle moment [Nm/kg]	0.04 ± 0.04	0.08 ± 0.04	0.08 ± 0.04	**<0.001**
Knee Δmoment [Nm/kg]	3.53 ± 0.56	3.55 ± 0.74	3.59 ± 0.64	0.789
Ankle Δmoment [Nm/kg]	3.09 ± 0.51	3.17 ± 0.61	3.09 ± 0.59	0.185

K_{leg} – leg stiffness; K_{vert} – vertical stiffness; K_{joint} – joint stiffness; t_c – ground contact time; Δy – vertical displacement of the COM; $angle_0$ – initial contact angle; Δangle – displacement of joint ankle; Δmoment – displacement of joint moment.

Runners showed similar Δy and t_c during the running stance phase, initial contact joint angle of knee and ankle joints, and displacement of ankle joint angle on the three different surfaces. ANOVA demonstrated significant differences in the displacement of knee joint angle among surfaces (F [2, 60] = 10.267, $p < 0.001$). Running on artificial grass produced a greater displacement of knee joint angle than running on synthetic rubber ($p = 0.002$, 95% CI = 1.52–7.35 degrees) and on concrete ($p = 0.006$, 95% CI = 1.04–7.25 degrees).

Significant differences were found in peak knee moment (F [2, 60] = 7.333, $p = 0.001$) and peak ankle moment (F [2, 60] = 21.383, $p < 0.001$) in the sagittal plane (Table 1). Bonferonni post-hoc test showed that running on concrete produced lower peak knee moment than running on artificial grass ($p = 0.003$, 95% CI = 0.11–0.58 Nm/kg). Moreover, running on synthetic rubber produced a lower peak ankle moment than running on concrete ($p < 0.001$, 95% CI = 0.03–0.07 Nm/kg) and on artificial grass ($p < 0.001$, 95% CI = 0.02–0.06 Nm/kg). No significant differences in the displacement of knee and ankle joint moment were found among the three running surfaces.

Differences were found in the maximal GRFs for anterior-posterior (F [2, 60] = 9.502, $p < 0.001$), and medial-lateral (F [2, 60] = 36.2, $p < 0.001$) directions (Table 2). The maximal anterior-posterior GRF appeared at about 70% of the stance phase and smaller when running on concrete than on synthetic rubber ($p < 0.001$, 95%CI = 0.02%–0.06%BW) and on artificial grass ($p = 0.044$, 95%CI = 0.01%–0.05%BW). In comparison, the maximal medial-lateral GRF appeared at about 10% of the stance phase. Running on concrete produced a lower maximal medial-lateral GRF than running on synthetic rubber ($p < 0.001$, 95%CI = 0.03%–0.06%BW) and on artificial grass ($p < 0.001$, 95% CI = 0.02%–0.05% BW), and the maximal medial-lateral GRF was greater on synthetic rubber than on artificial grass ($p = 0.036$, 95% CI = 0.01%–0.03% BW). Although no significant differences were observed in vertical GRF among the three surfaces, a greater vertical GRF was found on synthetic rubber than on concrete ($p = 0.016$, 95% CI =

Table 2. The maximal GRFs when running on three running surfaces (mean ± SD)

GRFs (% BW)	Synthetic rubber	Concrete	Artificial grass
A-P GRF (% BW)	0.37 ± 0.05#	0.33 ± 0.05#*	0.36 ± 0.06*
M-L GRF (% BW)	0.10 ± 0.03#^	0.05 ± 0.03#*	0.09 ± 0.04*^
v GRF (% BW)	2.71 ± 0.22#	2.62 ± 0.26#	2.64 ± 0.42

A-P – anterior-posterior; M-L – medial-lateral; # indicates significant differences between concrete and synthetic rubber; * indicates significant differences between concrete and artificial grass; ^ indicates significant differences between synthetic rubber and artificial grass.

0.02%–0.18% BW) and appeared at about 45% of the stance phase. Running on synthetic rubber and artificial grass showed similar GRFs (Fig. 1). However, running on concrete produced a higher anterior-posterior GRF at the posterior direction and a lower medial-lateral GRF at the lateral direction at about 0–45% of the stance phase.

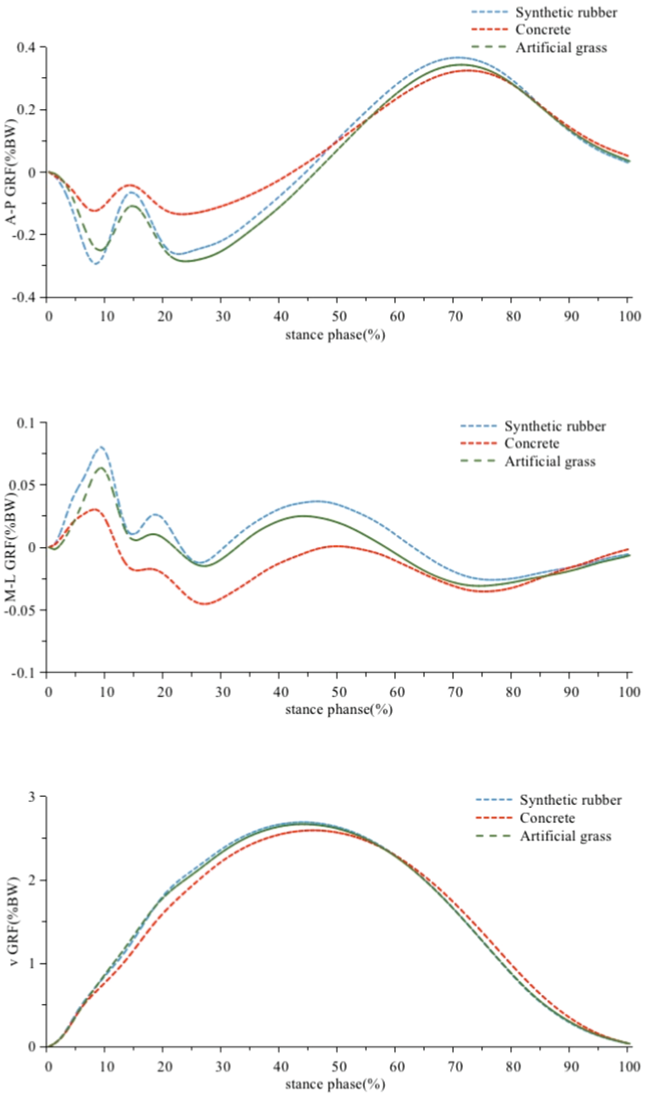

Fig. 1. The GRFs in anterior-posterior, medial-lateral, and vertical directions running on synthetic, concrete, and artificial grass

4. Discussion

The aim of this study was to investigate lower extremity stiffness and kinematic adjustments in hFFS when running on different overground surfaces. To the best of our knowledge, the current study was the first to investigate the influence of different surfaces on lower extremity stiffness in hFFS. The findings rejected the hypothesis that hFFS can accommodate lower extremity stiffness when running on conducted surfaces. Nevertheless, significant differences in kinematics and kinetics were observed. This result partially confirmed the hypothesis that hFFS can maintain similar lower extremity stiffness by adjusting kinematics and kinetics to compensate for differences in the hardness of surfaces.

Previous studies found that, when compensating for the compliance variance of surfaces, the human body adjusts its K_{leg} to maintain a stable trajectory of COM in hRFS [10], [11], [20], [22], [23], and running on a harder surface requires a stiffer ankle joint in hRFS [7]. In our study, the lower extremity stiffness and vertical displacement of COM in hFFS did not change during running on different overground surfaces. A possible explanation is that the prescribed running speed or the constructed surfaces were insufficient to cause biomechanical changes in the lower limbs. In addition, habitual foot strike pattern is an important factor that affects running performance, and hFFS are associated with better leg adjustment and transmission of elastic energy [39]. Furthermore, longitudinal arch (LA) behaves as a “foot spring” during running, thereby storing mechanical energy and subsequently returning via compression and recoil cycle along with changes in GRF [35]. Adjusting LA stiffness can adapt to alterations in surfaces and loading [36]. When running at the same speed, hFFS perform increased LA compliance and loading than hRFS [19]. In the current study, the hFFS maintained similar lower extremity stiffness during running on the three surfaces probably because of the better cushioning effect of LA among hFFS. In a future study, the stiffness of foot arch in hFFS running on different overground surfaces should be investigated as this information may provide insights into the kinematic adjustment in GRF.

Several studies reported differences in the kinematics of lower extremities when running on different surfaces among hRFS [6], [16], [17]. In general, when running on natural grass, the ground contact time is increased to attenuate the impact [17]. Via sensitivity analysis, Morin et al. [29] found that, for both K_{leg} and K_{vert}, the ground contact time is the most sensitive parameter. A 10% reduction in ground contact time leads to a 25% or 20% increase in K_{leg} or K_{vert}, respectively. Interestingly, some studies reported that ground contact time is not influenced by different surfaces [16], [34], which was an observation consistent with our results. Differences in surface materials, running speeds, and individual running strategy may account for the conflicting results.

In our study, the initial contact knee and ankle joint angles and the displacement of ankle joint angle were similar among the three surfaces. These findings were consistent with the observations of Dixon et al. [6], who found no considerable differences in kinematic variables when running on different surfaces. By contrast, Hardin et al. [16] reported that an increase in surface hardness results in a decrease in hip and knee flexion at contact. The present study indicated that the displacement of knee joint angle was greater on artificial grass. K_{joint} is dependent on joint moment and angular displacement. Although no significant differences in the displacement of knee joint moment were observed, the values of K_{joint} for knee were similar. These contradictory results may confirm that hFFS depend mainly on knee joint regulation to adapt to different running surfaces.

The present study found that peak knee moment was decreased when running on concrete and peak ankle moment was decreased when running on synthetic rubber. These results disagreed with the findings of Willwacher et al. [38], who reported that running barefoot at 3.5 m/s increases ankle joint moment and decreases knee joint moment on hard surfaces. Running speed and type may partly explain the reason for the differences [2]. Kerdok et al. [20] found significant differences when running on the softest surface. Only when surface hardness sufficiently or rapidly changes, lower extremity stiffness can be remarkable. In addition, although the experimental process was generally consistent, the foot strike patterns used were different.

Several studies on running on treadmills have been conducted. These works may provide controlled conditions in a more stable running speed versus overground running. These studies confirmed that only small changes in knee angle are related to an increase in K_{leg} when running on the softest surface [20]. Gidley et al. [13] reported that running on a compliant treadmill elicits a substantially more compliant leg than a rigid treadmill, a finding that was inconsistent with that of previous studies [11], [20]. A comparison between treadmill and overground running suggested that kinematics and kinetics seem similar. This comparison observed notable differences in knee kinematics and peak joint moment. These results indicated that running on a treadmill may cause slight differences in running patterns. Thus, the results of running on a treadmill, should be compare carefully, although our focus was on running in real life.

Unsurprisingly, remarkable differences in GRFs were observed herein, consistent with the findings of a previous work [33]. The maximal anterior-posterior and medial-lateral GRFs substantially decreased while running on concrete, whereas the maximal vGRF was smaller while running on concrete than on synthetic rubber. The maximal vGRF would likely remain constant within the range of surface hardness due to changes in lower extremity stiffness and kinematics. Zhang et al. [40] found that running on rubber results in considerably lower plantar pressure in partial areas of the foot compared with running on concrete. However, they found no differences between running on grass and running on rubber or concrete. Their findings were consistent with those of the present study. Results indicated that the regulation of GRFs may be the most important mechanism, especially in the frontal and horizontal planes.

The present study has several limitations. First, we did not quantify the hardness of surfaces. Thus, we were unable to specifically explain the possible absorbing ability of the surfaces to the impact of running. This omission might have affected the results. In addition, we allowed the participants to practice several times before the formal test. Doing so might have induced accommodation to different surfaces. And the 15 m runway might not be enough to observe differences when running on 3 different overground surfaces. Thus, further studies should focus on longer monitoring among hFFS.

5. Conclusions

In summary, this study found no differences in lower extremity stiffness among hFFS running on different overground surfaces. Adjustment of lower extremities on kinematics and kinetics may result in similar lower extremity stiffness. Findings in current study indicate that running on a hard surface induced more loads in knee joint, while running on a soft surface performed more loads in ankle joint among hFFS. Therefore, hFFS should choose the appropriate running surface for themselves to avoid sports injuries and improve running performance. These findings may provide insight into the adjustment mechanism of hFFS when running on diverse terrains in real world.

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References

[1] ADKISON J., REQUA R., GARRICK J., *Injury rates in high school football. A comparison of synthetic surfaces and grass fields*, Clin. Orthop. Relat. Res., 1974, 99, 131–136.

[2] ARAMPATZIS A., BRUGGEMANN G.P., METZLER V., *The effect of speed on leg stiffness and joint kinetics in human running*, J. Biomech., 1999, 32, 1349–1353.

[3] BISHOP M., FIOLKOWSKI P., CONRAD B., BRUNT D., HORODYSKI M., *Athletic footwear, leg stiffness, and running kinematics*, J. Athl Train., 2006, 41, 387–392.

[4] BUTLER R.J., CROWELL H.P. 3rd., DAVIS I.M., *Lower extremity stiffness: implications for performance and injury*, Clin. Biomech., 2003, 18, 511–517.

[5] DAVIS I.S., RICE H.M., WEARING S.C., *Why forefoot striking in minimal shoes might positively change the course of running injuries*, J. Sport. Health. Sci., 2017, 6, 154–161.

[6] DIXON S.J., COLLOP A.C., BATT M.E., *Surface effects on ground reaction forces and lower extremity kinematics in running*, Med. Sci. Sports. Exerc., 2000, 32, 1919–1926.

[7] DOLENEC A., STIRN I., STROJNIK V., *Activation Pattern of Lower Leg Muscles in Running on Asphalt, Gravel and Grass*, Coll. Antropol., 2015, 39, 167–172.

[8] DRAGOO J.L., BRAUN H.J., *The effect of playing surface on injury rate: a review of the current literature*, Sports. Med., 2010, 40, 981–990.

[9] FERRIS D.P., FARLEY C.T., *Interaction of leg stiffness and surfaces stiffness during human hopping*, J. Appl. Physiol., 1997, 82, 15–22, discussion 13–14.

[10] FERRIS D.P., LIANG K., FARLEY C.T., *Runners adjust leg stiffness for their first step on a new running surface*, J. Biomech., 1999, 32, 787–794.

[11] FERRIS D.P., LOUIE M., FARLEY C.T., *Running in the real world: adjusting leg stiffness for different surfaces*, Proc. Biol. Sci., 1998, 265, 989–994.

[12] GHENA D.R., KURTH A.L., THOMAS M., MAYHEW J., *Torque Characteristics of the Quadriceps and Hamstring Muscles during Concentric and Eccentric Loading*, J. Orthop. Sports. Phys. Ther., 1991, 14, 149–154.

[13] GIDLEY A.D., LANKFORD D.E., BAILEY J.P., *The construction of common treadmills significantly affects biomechanical and metabolic variables*, J. Sports. Sci., 2020, 38, 2236–2241.

[14] GIJON-NOGUERON G., FERNANDEZ-VILLAREJO M., *Risk Factors and Protective Factors for Lower-Extremity Running Injuries A Systematic Review*, J. Am. Podiatr. Med. Assoc., 2015, 105, 532–540.

[15] GRIER T.L., CANHAM-CHERVAK M., ANDERSON M.K., BUSHMAN T.T., JONES B.H., *Effects of Physical Training and Fitness on Running Injuries in Physically Active Young Men*, J. Strength. Cond. Res., 2017, 31, 207–216.

[16] HARDIN E.C., VAN DEN BOGERT A.J., HAMILL J., *Kinematic adaptations during running: effects of footwear, surface, and duration*, Med. Sci. Sports. Exerc., 2004, 36, 838–844.

[17] HOLLIS C.R., KOLDENHOVEN R.M., RESCH J.E., HERTEL J., *Running biomechanics as measured by wearable sensors: effects of speed and surface*, Sports. Biomech., 2019, 7, 1–11.

[18] JANDOVA S., SPARKS M., OLESNIEWICZ P., CHAROUSEK J., CHRASTKOVA M., MARKIEWICZ J., *Changes in the foot strike pattern and pressure distribution when running in minimalist and traditional sport shoes*, Medicina Dello Sport, 2018, 71, 257–267.

[19] KELLY L.A., FARRIS D.J., LICHTWARK G.A., CRESSWELL A.G., *The Influence of Foot-Strike Technique on the Neuromechanical Function of the Foot*, Med. Sci. Sports. Exerc., 2018, 50, 98–108.

[20] KERDOK A.E., BIEWENER A.A., MCMAHON T.A., WEYAND P.G., HERR H.M., *Energetics and mechanics of human running on surfaces of different stiffnesses*, J. Appl. Physiol., 2002, 92, 469–478.

[21] LY Q.H., ALAOUI A., ERLICHER S., BALY L., *Towards a footwear design tool: influence of shoe midsole properties and ground stiffness on the impact force during running*, J. Biomech., 2010, 43, 310–317.

[22] MACLELLAN M.J., PATLA A.E., *Adaptations of walking pattern on a compliant surface to regulate dynamic stability*, Exp. Brain. Res., 2006, 173, 521–530.

[23] MARIGOLD D.S., PATLA A.E., *Adapting locomotion to different surface compliances: neuromuscular responses and changes in movement dynamics*, J. Neurophysiol., 2005, 94, 1733–1750.

[24] MARKOTIC V., POKRAJCIC V., BABIC M., RADANCEVIC D., GRLE M., MILJKO M., KOSOVIC V., JURIC I., KARLOVIC VIDAKOVIC M., *The Positive Effects of Running on Mental Health*, Psychiatr. Danub., 2020, 32, 233–235.

[25] MCGINLEY J.L., BAKER R., WOLFE R., MORRIS M.E., *The reliability of three-dimensional kinematic gait measurements: a systematic review*, Gait. Posture, 2009, 29, 360–369.

[26] MCMAHON T.A., GREENE P.R., *The influence of track compliance on running*, J. Biomech., 1979, 12, 893–904.

[27] MEI Q., GU Y., FERNANDEZ J., *A biomechanical assessment of running with hallux unstable shoes of different material stiffness*, Acta Bioeng Biomech, 2019, 21, 121–128.

[28] MELCHER D.A., PAQUETTE M.R., SCHILLING B.K., BLOOMER R.J., *Joint stiffness and running economy during imposed forefoot strike before and after a long run in rearfoot strike runners*, J. Sports. Sci., 2017, 35, 2297–2303.

[29] MORIN J.B., DALLEAU G., KYROLAINEN H., JEANNIN T., BELLI A., *A simple method for measuring stiffness during running*, J. Appl. Biomech., 2005, 21, 167–180.

[30] MRDAKOVIC V., ILIC D., VULOVIC R., MATIC M., JANKOVIC N., FILIPOVIC N., *Leg stiffness adjustment during hopping at different intensities and frequencies*, Acta Bioeng Biomech, 2014, 16, 69–76.

[31] NIGG B.M., KERR B., *Biomechanical aspects of sports shoes and playing surfaces*, Calgary: University of Calgary Press, 1983.

[32] PAPPAS P., DALLAS G., PARADISIS G., *Reliability of Leg and Vertical Stiffness During High Speed Treadmill Running*, J. Appl. Biomech., 2017, 33, 160–165.

[33] RILEY P.O., DICHARRY J., FRANZ J., DELLA CROCE U., WILDER R.P., KERRIGAN D.C., *A kinematics and kinetic comparison of overground and treadmill running*, Med. Sci. Sports. Exerc., 2008, 40, 1093–1100.

[34] STAFILIDIS S., ARAMPATZIS A., *Track compliance does not affect sprinting performance*, J. Sports. Sci., 2007, 25, 1479–1490.

[35] STEARNE S.M., MCDONALD K.A., ALDERSON J.A., NORTH I., OXNARD C.E., RUBENSON J., *The Foot's Arch and the Energetics of Human Locomotion*, Sci. Rep., 2016, 6, 19403.

[36] VEREECKE E.E., AERTS P., *The mechanics of the gibbon foot and its potential for elastic energy storage during bipedalism*, J. Exp. Biol., 2008, 211, 3661–3670.

[37] WEI Z., ZHANG Z., JIANG J., ZHANG Y., WANG L., *Comparison of plantar loads among runners with different strike patterns*, J. Sports. Sci., 2019, 37, 2152–2158.

[38] WILLWACHER S., FISCHER K.M., ROHR E., TRUDEAU M.B., HAMILL J., BRUGGEMANN G.P., *Surface Stiffness and Footwear Affect the Loading Stimulus for Lower Extremity Muscles When Running*, J. Strength. Cond. Res., 2020, DOI: 10.1519/JSC.0000000000003410.

[39] YIN L., HU X., LAI Z., LIU K., WANG L., *Leg Stiffness and Vertical Stiffness of Habitual Forefoot and Rearfoot Strikers during Running*, Appl. Bionics. Biomech., 2020, 2020, 8866340.

[40] ZHANG Z., ZHANG Y., FU W., WEI Z., J. J., WANG L., *Plantar Loads of Habitual Forefoot Strikers during Running on Different Overground Surfaces*, Appl. Sci., 2020, 10, 2271.