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Technologically advanced running shoes reduce biomechanical factors of running related injury risk

Hyunji Kim[1,2] & Jooeun Ahn[1,2,3]✉

Recent footwear technology has led to the development of technologically advanced running shoes (TARS), which improve running performance. However, the effect of TARS on biomechanical risk factor remains unclear. This study compares the effects of TARS with those of conventional cushioned shoes (CON) and minimalist shoes (MIN) on running biomechanics and biomechanical risk factors. We recruited 15 recreational runners, measured their ventilation threshold speeds and habitual strike angles, collected kinematic data and ground reaction forces across shoe conditions, and estimated joint reaction force and muscle force through inverse dynamic analysis. Results show that TARS significantly alter landing patterns by shifting runners toward a forefoot/midfoot strike patterns (mean strike angle decreased by 4.17° compared to CON) and reducing subtalar eversion during loading phase. While MIN increase peak ankle joint reaction force by 3.07 body weight (BW) compared to CON, TARS reduce it by 1.84 BW. TARS also decrease peak soleus and peroneus longus forces by 1.10 BW and 0.43 BW respectively, without increasing demands on any joint. These results suggest that TARS provide distinct biomechanical characteristics that reduce certain mechanical loads associated with running injuries. Our findings further suggest a need for reevaluating footwear classification methods and embracing technological advancements in running shoe design for potentially safer and more efficient running.

Keywords Footwear, Running, Running related injury, Musculoskeletal modeling

Endurance running is one of the most popular forms of exercise. Multiple studies in biological anthropology even suggest that humans have evolved as superior endurance runners for millions of years[1,2]. In modern society, footwear technology has been developed to prevent running-related injury (RRI) and enhance running performance[3–6]. The initial focus of footwear technology was on reducing impact through cushioning and controlling excessive pronation to prevent RRI[4,6]. This approach led to the widespread adoption of conventional cushioned shoes (CON)[7,8]. Although some studies have supported the role of cushioned footwear in preventing RRI[4,7], multiple other studies have questioned its effectiveness and even suggested potential drawbacks probably caused by cushioning[2,8,9]. Accordingly, attention was paid to minimalist shoes (MIN), which have lightweight, reduced cushioning, and low heel-to-toe drop[10,11]. Proponents of MIN suggest that this design encourages a forefoot strike (FFS) or midfoot strike (MFS) pattern, which may reduce injury risk and improve performance[1,8–14]. They claim that cushioned shoes, by contrast, induce rearfoot strike (RFS), which is incompatible with long-evolved human running biomechanics[2,8]. Some researchers have even suggested that this 'unnatural' RFS induced by excessive cushioning may contribute to a higher prevalence of RRI[2,8,12,13]. While such claims remain debated, multiple studies have shown that RFS patterns—frequently observed in cushioned footwear—are associated with elevated vertical loading rates, higher peak vertical ground reaction forces, and increased energy absorption at the knee joint during early stance[6,9,12,15,16]. These features have been proposed as potential biomechanical injury risk factors and are visually summarized in Fig. 1.

More recently, footwear technology focusing on performance enhancement has led to the development of technologically advanced running shoes (TARS), which enabled endurance runners to run with improved running economy[15,17–20]. TARS have highly resilient midsole foam materials, and carbon fiber plates[15,17,21], featuring not only thick elastic soles[22] and low minimalist index[23] similar to CON but also lightweight and heel-to-toe offset more akin to MIN[18,19]. Therefore, TARS cannot be categorized as CON or MIN. Despite these distinctive features of TARS, their impact on running biomechanics has not yet been systematically compared with CON and MIN, whereas the biomechanical effects of CON and MIN have been extensively

[1]Department of Physical Education, Seoul National University, Seoul 08826, South Korea. [2]Soft Robotics Research Center, Seoul National University, Seoul 08826, South Korea. [3]Institute of Sport Science, Seoul National University, Seoul 08826, South Korea. ✉email: ahnjooeun@snu.ac.kr

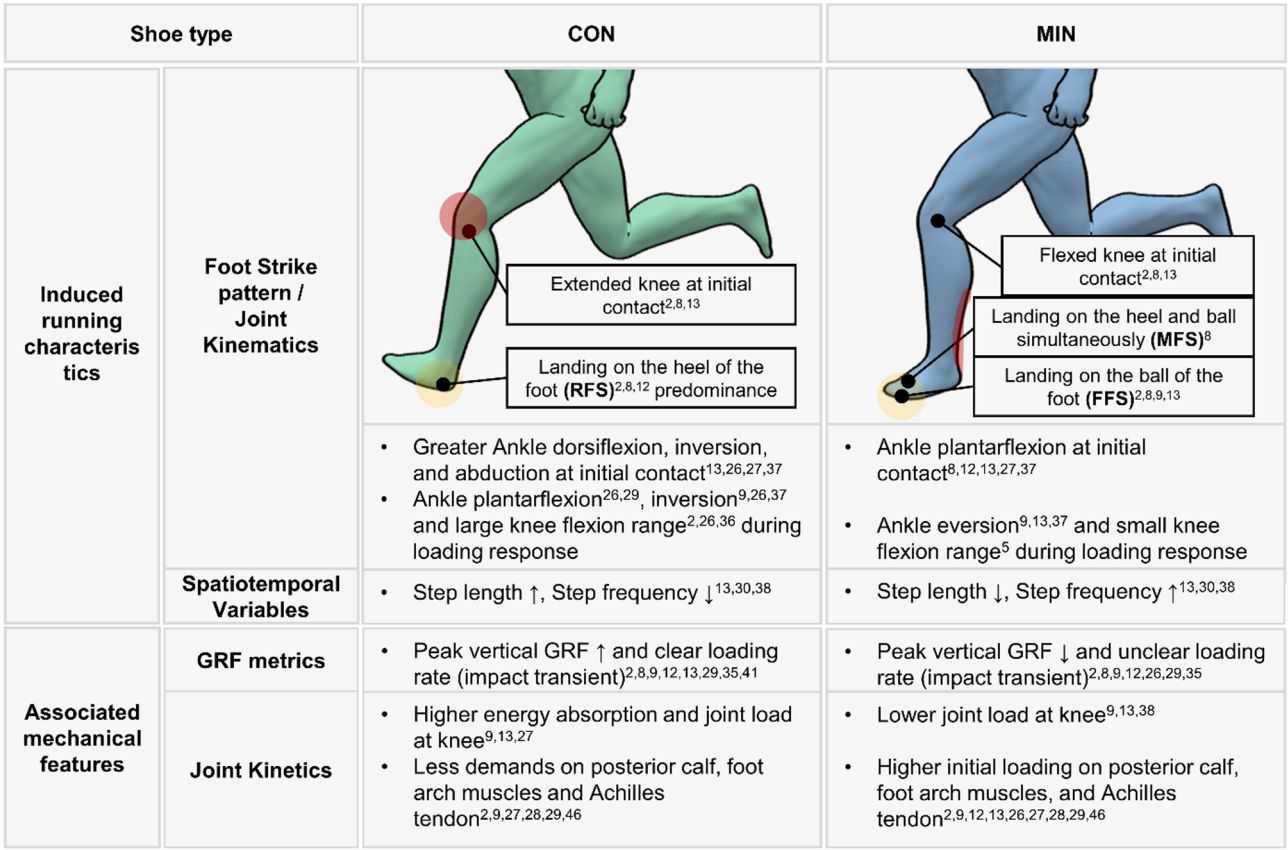


Fig. 1. Previously reported running patterns induced by conventional cushioned shoes (CON) and minimalist shoes (MIN). RFS, rearfoot strike; MFS, midfoot strike; FFS, forefoot strike; GRF, ground reaction force.

studied (Fig. 1)[2,8,9,12,13,24–30]. Although recent studies have reported the kinematics and kinetics of running with TARS[18,22,31,32], they have focused primarily on the effects of TARS on RFS runners. Some articles also raised concerns regarding a potential association between TARS and specific RRI based on anecdotal clinical observations[3,33], but the claimed concerns are based on the presumed trade-off between performance and RRI rather than on biomechanical analyses or a clear understanding of the changes resulting from TARS.

We aim to address this research gap. We systematically evaluate the biomechanical effects of TARS relative to CON and MIN. We examined not only external variables such as spatiotemporal parameters, GRFs, and joint kinematics, but also internal variables including joint reaction forces (JRFs) and muscle forces estimated through inverse dynamics. These variables were selected based on their relevance to known injury mechanisms and performance determinants, as reported in previous literature (Fig. 1)[2,8,9,12,13,26,28,29,34–47]. Our hypothesis was that TARS would result in altered lower-limb biomechanics and affected internal loading metrics—interpretable as biomechanical risk factors—compared with CON and MIN.

Methods

Participants

The required sample size was calculated using G*Power 3.1 software (3.1. version, Heinrich-Heine-Universität Düsseldorf, Düsseldorf, Germany) based on average effect size of F = 0.565, calculated from joint moment values reported in prior studies[18,48]. With a significance level of 5% and a power of 80%, the minimum required sample size for a repeated measures ANOVA with three groups was determined as 12. Considering possible dropping out, we decided to recruit 16 runners. The inclusion criteria were established based on a Delphi consensus study defining recreational runners[49]. These criteria included: (1) age between 20 and 35 years; (2) routinely running 15–50 km per week; (3) completed a 10 K run in 40–50 min within the past year; (4) no acute injury in the preceding 6 months; and (5) no prior experience running in the shoes utilized in this study. We deliberately recruited only male participants to secure the validity of the inverse analysis. The sex-specific skeletal structure and associated biomechanical loading differences significantly influence the resultant joint reaction forces and muscle forces estimated by inverse dynamic analysis[50,51], and the musculoskeletal model used in this study (AnyBody with TLEM 2.2) has been primarily developed and validated using male anatomical data[44,52–55]. Although sex-based anthropometric scaling is possible, it does not fully account for differences in joint geometry, or soft tissue distribution. One participant discontinued participation owing to pain following the use of MIN. Thus, 15 male participants were included in the study (age: 27.6 ± 3.2 years; height: 176.2 ± 4.4 cm; body mass: 75.1 ± 5.3 kg; and 10 K record: 45.7 ± 2.3 min). The Seoul National University institutional review board approved

the study protocol (institutional review board number 2112/001-012), which conformed to the ethics described in the Declaration of Helsinki. All participants provided written informed consent prior to participation, and they also signed informed written consent forms for the publication of any identifying information or images in an online open-access publication.

Experimental design

All participants underwent an incremental test to exhaustion wearing their habitual shoes before participating in the main experiment. The incremental test began after a 5-min warm-up run at 2 m/s, after which the speed was increased by 0.28 m/s every minute. The ratings of perceived exertion, heart rate, and $\dot{V}O_2$ of the participants were monitored as the speed increased. The test was terminated when the ratings of perceived exertion achieved a value of 20, indicating extreme exertion, or when a further increase in the heart rate and $\dot{V}O_2$ was not observed.

For footwear model selection, we performed a focus group interview with two professional and two recreational runners, who did not participate in the running experiment, to assess the popularity and practical relevance of candidate models within each category of CON, TARS, or MIN. These discussions helped identify widely used models in the running community. The selected shoe models were cross-validated with prior literature[19,23,49] to ensure that their material and structural characteristics (e.g., mass, midsole height, and minimalist index) align with properties of shoes in each category. Supplementary Table S1 summarizes the selected shoe models and their properties.

At least 2 and up to 10 days after the incremental test, participants completed six 7-min sub-maximal steady state runs, with two runs under each shoe condition. The order of the six runs was randomized and counterbalanced to minimize any possible order effects. Each 7-min run was composed of a 1-min run at 70%, a 1-min run at 80%, and a 5-min run at 90% of the ventilation threshold speed (3.97 ± 0.30 m/s) that was measured in the preceding incremental test. A minimum rest period of 20 min was provided between 7-min runs. The rest period was extended until the participant reported readiness and the heart rate went below 100 beats per minute.

Data analysis

We followed the reporting guidelines for running biomechanics and footwear studies using 3D motion capture, which was proposed by Hébert-Losier et al.[56]. The guidelines include detailed information on sampling frequency, data processing, motion capture system configuration, marker placement, biomechanical model specifications, and calibration.

Marker data were collected at 200 Hz using a 12-camera motion capture system (Optitrack Prime 13, Natural Point, OR, USA), which has been validated with spatial resolution errors below 200 μm[57]. GRF data were acquired at 1000 Hz via an instrumented treadmill (Bertec Corporation, OH, USA), which provides reliable vertical GRF measurements with a coefficient of variation under 3%[58]. Metabolic data were recorded breath-by-breath using a portable respirometry analyzer (K5, COSMED, RM, Italy), which was reported to have a coefficient of variation of 4.5% and a concordance correlation coefficient of 0.91 for $\dot{V}O_2$ measurement[59]. A heart rate monitor (Garmin Ltd., KS, USA) was used in synchronization with the metabolic data collection system.

Twenty-six reflective markers were attached to the lower extremities during the incremental test. The foot strike angle (FSA) was defined as the angle in the sagittal plane obtained by subtracting the angle between the foot and the ground in the standing posture from the angle between the foot and the ground at initial contact (IC) (Fig. 2a)[10]. The habitual FSA of each participant was calculated as the average FSA over a 40-second period during which the treadmill belt speed was closest to 90% of the ventilation threshold speed of the participant. Fifty reflective markers, including the plug-in gait marker set, were placed on the whole body during the 7-minute sub-maximal steady state runs. The marker and GRF data collected during the last 2 min of the entire 7-minute run were filtered using a zero-lag low-pass fourth-order Butterworth filter with a 20 Hz cut-off frequency[60]. Each event of foot strike was determined based on the vertical GRF threshold of 20 N[41].

The AnyBody Modeling System (ver. 7.3.2; Anybody Technology, Aalborg, Denmark) with a full-body musculoskeletal model, including the TLEM 2.2 model, was used to calculate the inverse kinematics and kinetics. The joint angles, joint velocities, internal JRFs, joint moments, and muscle forces[61,62] were calculated using parameter identification, marker tracking, and inverse dynamic analysis. The accuracy of this modeling approach has been supported by validation studies. Marra et al. reported joint force prediction accuracy with RMSE < 0.4 body weight (BW) and $R^2 > 0.8$[44]. Damsgaard et al. demonstrated strong agreement between simulated muscle activations and experimental EMG[53].

Mechanical power was computed as the product of joint moment and angular velocity. The negative and positive power were integrated to calculate the amount of absorbed and generated mechanical energy, respectively. The amount of absorbed energy was calculated during the loading response (defined from IC to peak knee flexion angle during stance)[38,63], whereas the amount of generated energy was calculated during the entire stance phase. The relative contribution of each of hip, knee, ankle, and subtalar to the total energy absorption and generation was calculated. The peak JRFs and muscle forces were normalized to the body weight of each participant.

We measured the average time between each IC of the foot using the GRF data during the last two minutes, and defined the time interval as the average step time. Then, we calculated step frequency as the inverse of the average step time, and determined dimensionless step frequency by normalizing the step frequency to the natural frequency of the lower limb using the following formula:

$$\text{dimensionless step frequency} = \text{step frequency} / \sqrt{g/l},$$

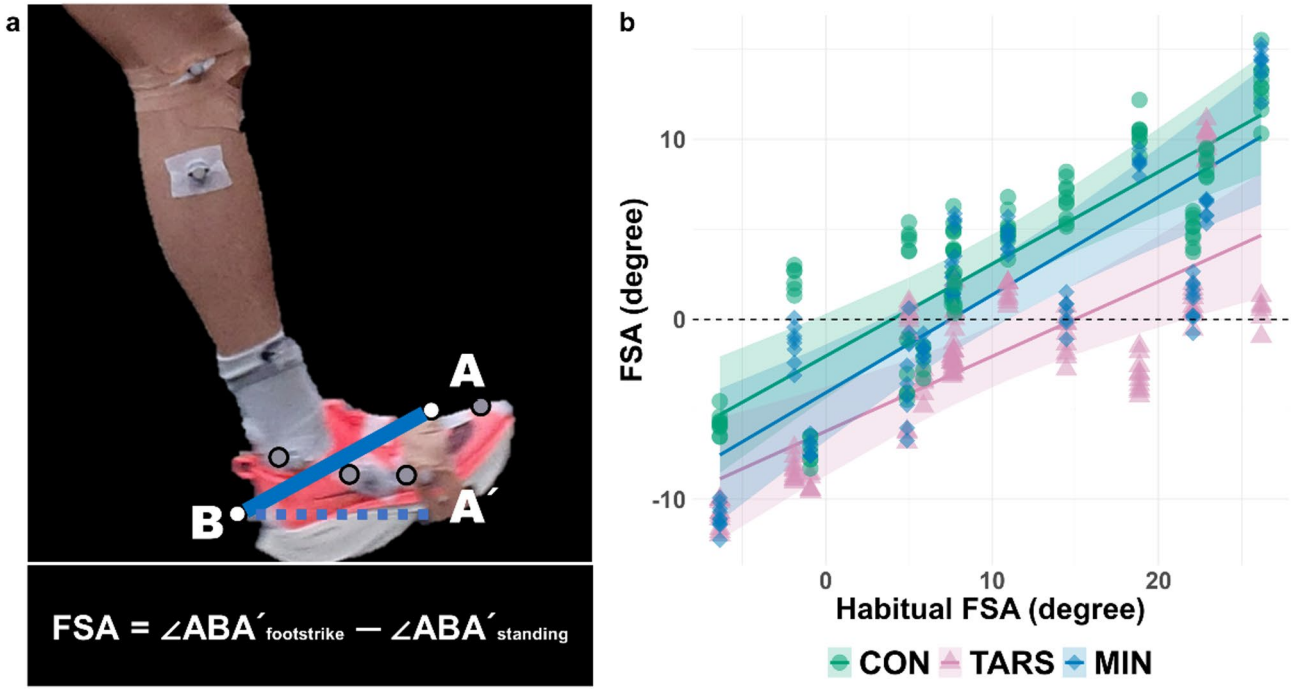


Fig. 2. Foot strike angle (FSA) and the effect of shoe condition and habitual FSA on the resultant FSA. (**a**) FSA defined as the angle obtained by subtracting the angle between the foot and the ground in the standing posture from the angle between the foot and the ground at initial contact[10]. (**b**) Estimated marginal means of FSA for each shoe condition across the range of habitual FSA values with 95% confidence intervals presented by shaded areas. The observed data points for each shoe condition are overlaid in a scatter plot.

where g denotes the gravitational acceleration, and l represents the length from the hip to the ankle joint in a standing position[30]. We calculated the step length by dividing the treadmill belt speed by the step frequency. Then, we normalized the step length by dividing it by l. The contact time was calculated as the average time between IC and toe-off of both feet. The horizontal distances from the center of mass to the ankle and from the knee to the ankle were normalized to l. The braking impulse was calculated as the integral of the negative anteroposterior GRF from IC to midstance, whereas the propulsion impulse was calculated by integrating the positive anteroposterior GRF from midstance to toe-off. The highest value of the vertical GRF was additionally identified.

Statistical analysis

The experimental design involved repeated measures with three shoe conditions (CON, TARS, and MIN) as within-subject variables. All statistical analyses were performed using R (v 4.4.0, R Core Team). The dependent variables included the kinematics and kinetics of running, internal loads and joint power. The influence of shoe conditions on each dependent variable was assessed using linear mixed-effects models with the lme4 package in R[64]. The CON condition was set as the reference level for comparing the effects of the TARS and MIN conditions. The habitual FSA measured during the incremental test was included as a covariate to account for the potential interaction effects with shoe conditions. Fixed effects (shoe condition, habitual FSA, and their interaction) and random effects (intercept and shoe condition within subjects) were included in the mixed-effects model to account for the within-subject shoe condition correlation induced by repeated measures using an unstructured covariance matrix. The normality assumption of residuals was not met for distances in the sagittal plane at IC and running kinetics variables, so generalized linear mixed models with a Tweedie distribution were used to accommodate their specific distributions. Statistical significance for interaction and main effect terms was set a priori at $p < 0.05$, and the p value for statistical significance was adjusted for multiple comparisons using a simultaneous interference procedure[65].

Results

Physiological characteristics of participants

The physiological characteristics of the participants, which were obtained during the incremental test prior to the main experiments, are summarized in Table 1. The average $\dot{V}O_{2max}$ across the cohort was 56.60 ± 6.74 ml/kg/min, indicating a generally high aerobic fitness level consistent with trained recreational runners. Ventilatory threshold speeds ranged from 3.31 to 4.29 m/s, and the average heart rate at $\dot{V}O_{2max}$ was 192.40 ± 4.66 bpm.

Participant ID	$\dot{V}O_{2max}$/kg (ml/kg/min)	Ventilatory threshold speed (m/s)	Heart Rate at $\dot{V}O_{2max}$ (bpm)
P01	65.82	3.75	180
P02	52.97	3.55	188
P03	53.51	3.82	188
P04	60.72	4.29	194
P05	60.48	3.76	198
P06	64.83	4.00	196
P07	62.81	4.29	197
P08	65.34	4.00	193
P09	46.77	3.76	188
P10	56.26	4.07	193
P11	53.57	4.07	193
P12	49.95	3.82	193
P13	51.09	3.31	196
P14	45.43	4.06	194
P15	59.54	4.06	195

Table 1. Incremental test results: $\dot{V}O_{2max}$ and ventilatory threshold speeds for each participant.

FSA

Considering the potential influence of runners' habitual foot strike patterns on shoe-induced changes, we investigated the effects of shoe conditions, the habitual FSA, and their interaction on biomechanical variables during running using linear mixed-effects models. The distribution of habitual FSA of all participants measured during the incremental tests is summarized in Supplementary Fig. S2. The analysis revealed that FSA resulting from TARS was significantly lower than FSA resulting from CON ($\beta = -4.174$, $p < 0.05$). In contrast, the linear mixed-effects models did not conclude a statistically significant difference between FSA resulting from MIN and FSA resulting from CON (Fig. 2b, Supplementary Table S2). These findings indicate that TARS are more effective than MIN in reducing FSA and thus inducing an FFS or MFS pattern. The habitual FSA also affects FSA during running; individuals with higher habitual FSA exhibit higher FSA during running regardless of the shoe conditions ($\beta = 0.512$, $p < 0.001$). No significant interaction is observed between habitual FSA and shoe conditions.

Running mechanics

Shoe conditions affect foot kinematics and spatiotemporal variables. TARS significantly increased ankle plantarflexion angle ($\beta = 6.076$, $p < 0.05$) and subtalar eversion angles ($\beta = 4.731$, $p < 0.01$) at IC compared with CON (Fig. 3a, Supplementary Table S3). During loading response, TARS significantly increased ankle plantarflexion angle ($\beta = 1.408$, $p < 0.05$) and decreased subtalar eversion angle ($\beta = -1.808$, $p < 0.05$) (Supplementary Table S3). No significant difference in the kinematics of the limbs above the ankle joint complex was observed among the different shoe conditions. In contrast, habitual FSA exhibits significant effects on hip flexion ($\beta = 0.201$, $p < 0.001$) and knee flexion ($\beta = -0.236$, $p < 0.001$). Higher habitual FSA is also associated with increased knee flexion during loading response ($\beta = 0.190$, $p < 0.01$) (Supplementary Table S3). MIN significantly increased step frequency and dimensionless step frequency, and decrease step length and normalized step length (Supplementary Table S4).

We also observed that MIN reduce the peak vertical GRF compared with CON ($\beta = -0.088$, $p < 0.001$), whereas TARS exhibit no significant effect ($\beta = 0.010$, $p > 0.05$) (Fig. 3b, Supplementary Table S5). We additionally found that the effect of MIN on the reduction in the peak vertical GRF diminishes as habitual FSA increases with significant interaction ($\beta = 0.004$, $p < 0.05$) (Supplementary Table S5).

Our estimation from inverse dynamic analysis indicates that TARS significantly reduce the peak resultant ankle JRF compared with CON ($\beta = -1.835$, $p < 0.01$), whereas MIN significantly increase it ($\beta = 3.074$, $p < 0.001$) (Fig. 3c, Supplementary Table S6). MIN also lead to significant increases in peak gastrocnemius ($\beta = 0.936$, $p < 0.01$) and peak soleus forces ($\beta = 1.510$, $p < 0.001$) compared with CON, whereas TARS significantly reduce the peak soleus ($\beta = -1.096$, $p < 0.001$) and peak peroneus longus forces ($\beta = -0.433$, $p < 0.01$) (Fig. 3e, Supplementary Table S6).

In addition, MIN significantly increase the ankle joint's contribution to energy generation compared with CON ($\beta = 3.193$, $p < 0.05$), whereas TARS do not (Fig. 3d, Supplementary Table S6). Furthermore, TARS do not increase the demands on the knee or hip either (Supplementary Table S6).

Discussion

Our findings demonstrated that TARS promoted an FFS or MFS with decreased FSA and increased plantarflexion angle at IC. However, the kinematic pattern induced by TARS is distinct from that typically observed in habitual FFS or MFS runners. Notably, TARS increase subtalar eversion at IC but decrease it during the loading response. A previous study suggested that excessive subtalar eversion during the loading response can increase the risk of injury, including medial tibial stress syndrome, commonly known as shin splints[36,66]. It was additionally

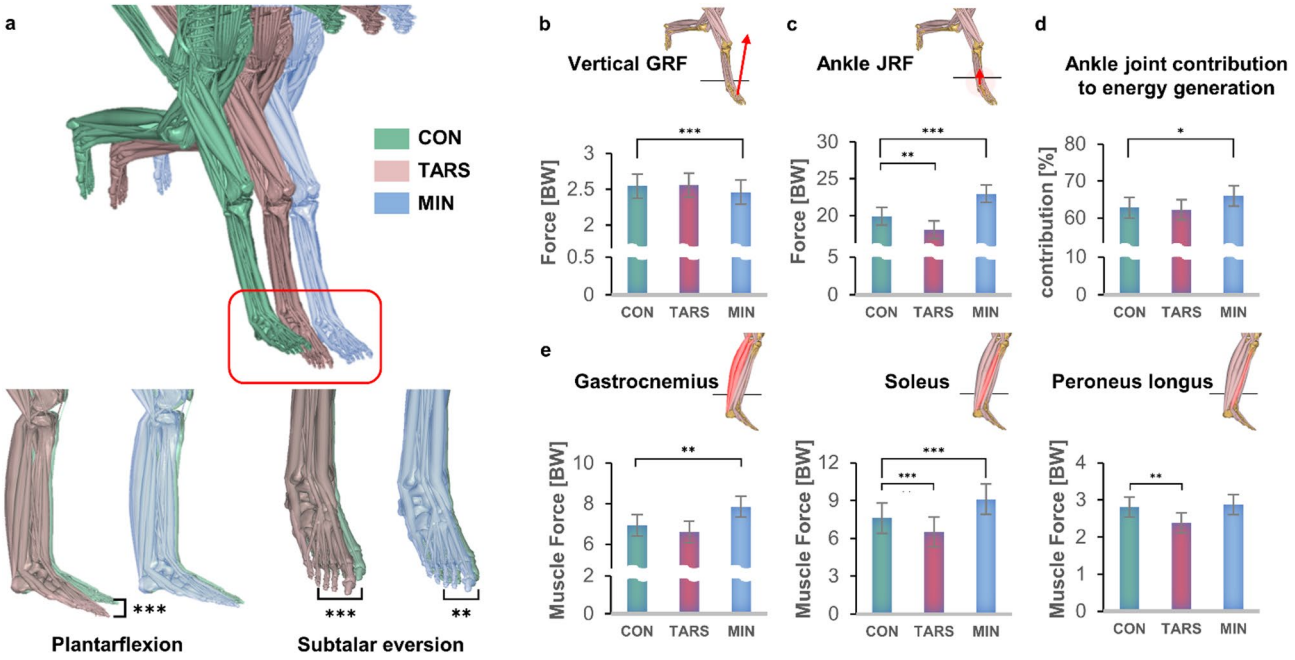


Fig. 3. Kinematics, kinetics, and internal forces across different shoe conditions. (**a**) The plantarflexion and subtalar eversion angles at initial contact. (**b**) Peak vertical ground reaction force (GRF). (**c**) Peak ankle joint reaction force (JRF). (**d**) The contribution of the ankle joint to the total energy generation. (**e**) Peak forces of the gastrocnemius, soleus, and peroneus longus. All the force data are normalized to the body weight (BW) of the runner. Asterisks indicate significant differences between shoe conditions; $^{*}p < 0.05$, $^{**}p < 0.01$, and $^{***}p < 0.001$. CON, Conventional cushioned shoes; TARS, Technologically advanced running shoe; MIN, Minimalist shoes.

reported that habitual FFS or MFS runners exhibit greater eversion than RFS runners during shod running, and this shoe-induced eversion can result in abnormal lower extremity loading[67]. We demonstrated that TARS eventually decreased eversion during the loading response. Given that the majority of stress to the joints and muscles develops during the loading response[4,38,68,69], the unique kinematic features resulting from TARS, which are observed in the present study, indicate that use of TARS does not necessarily increase the risk of RRI like medial tibial stress syndrome. MIN-induced spatiotemporal changes aligned with previous reports for runners wearing MIN or adopting an FFS or MFS pattern (Fig. 1)[13,30]. However, TARS achieved similar strike pattern changes without these spatiotemporal alterations, suggesting unique biomechanical effects.

Although we observed that MIN reduced peak vertical GRF compared to CON as in previous studies (Fig. 1)[2,8,9,12,26], the clinical significance of this metric in assessing RRI risk is questionable. The peak vertical GRF reflects loading only in a single direction, and GRF alone does not directly represent the forces applied to internal tissues[39,70]. Therefore, a comprehensive analysis of internal loading must be performed to assess biomechanical risk factor more thoroughly. Our inverse dynamic analysis revealed that TARS significantly reduced peak resultant ankle JRF and key muscle forces, including soleus and peroneus longus. This sharply contrasted with MIN, which are reported to increase these internal loads in this study as well as previous studies (Fig. 1)[27,28]. The average peak resultant ankle JRF induced by TARS was lower than those induced by CON and MIN by 1.84 and 4.91 BW, respectively (Fig. 3c). In addition, the average peak soleus force induced by TARS was lower than those induced by CON and MIN by 1.10 and 2.61 BW, respectively, and the reductions in the peak peroneus longus force were 0.43 and 0.50 BW, respectively (Fig. 3e). These reductions in internal loading by TARS have important biomechanical implications because the ankle JRF is the highest among all JRFs (Supplementary Table S6), and the affected muscles play crucial roles in running mechanics.

The peroneus longus stabilizes the ankle in the frontal plane during running. The soleus dissipates impact by eccentrically controlling ankle dorsiflexion moment and contributes to propulsion though plantarflexion during running. The activation of soleus also affects knee flexion velocity during stance phase[16,71]. A previous study showed that running with MIN or adopting FFS pattern significantly increases soleus usage, potentially elevating mechanical demand on the Achilles tendon and the triceps surae (Fig. 1)[12,13,28]. In this study, we observed that TARS significantly reduced ankle JRFs, peak soleus force, and peak peroneus longus force without significant differences in knee JRFs. Our results show that TARS induce FFS pattern without imposing additional loads on the ankle and other joints. This is also consistent with the results of prior studies showing that footwear with inserted carbon fiber plates and increased longitudinal bending stiffness does not necessarily increase Achilles tendon loading associated with tendinopathy risk[20,72].

Previous studies speculated that TARS may increase the risk of RRI[3,33], based partly on the potential performance benefits[15,17–20] and the assumed trade-off between performance and injury risk. Our biomechanical

findings challenge these concerns from a mechanical loading perspective. The observed TARS-induced decreases in muscle forces and JRF, which are considered as reliable indicators of RRI risk[6,39,43,68,70], suggest that this type of footwear may not necessarily increase mechanical loading associated with common injuries.

In contrast, the observed MIN-induced increase in soleus, gastrocnemius, and peroneus longus muscle forces are consistent with the findings of previous studies that reported higher Achilles tendon loading rates and frontal plane ankle torques, caused by MIN[27,28]. The MIN-induced increases in mechanical demands on the posterior lower leg and ankle align with findings that MIN elevate the requirement for positive ankle work in FFS or MFS runners[48], potentially increasing the risk of Achilles tendinopathies, and stress fractures of the metatarsals and other foot and ankle joint[13,48,49]. However, TARS, despite the fact that they promote an FFS or MFS pattern better than MIN, did not increase the mechanical demand on the ankle for energy generation compared with CON (Fig. 3d),

The unique biomechanical profile we observed with TARS suggests limitations in traditional footwear categorization methods. Previous study highlighted the distinctive characteristics of these technologically advanced shoes and how they differ from conventional footwear in performance aspects[73]. Our findings empirically support that these differences extend to biomechanical injury-related parameters as well. TARS demonstrated a distinct profile—promoting an FFS/MFS pattern without increasing ankle joint loading—that cannot be adequately captured by the conventional dichotomy between minimalist and cushioned shoes. Future footwear classification systems should integrate both structural and functional properties to better represent the complexity of modern running shoes. Furthermore, for a more comprehensive classification, it is necessary to consider not only simple indices derived from the shape and weight of the shoes but also the mechanical properties of the soles and their interaction with the shape.

Our results also indicate that the biomechanical response to footwear is not uniform across individuals. Interaction effects observed between footwear type and habitual FSA for several variables (Supplementary Table S6) suggest that individual running patterns substantially influence how different shoes affect joint and muscle loading. Notably, the reductions in ankle JRFs and plantarflexor muscle forces with TARS were more pronounced in runners with higher habitual FSA (i.e., habitual FFS runners), indicating a greater mechanical benefit in this subgroup. These findings underscore the importance of incorporating individual gait characteristics, such as foot strike pattern, when evaluating injury risk and making footwear recommendations.

Our findings were based on data from treadmill running which may not fully replicate outdoor running[74]. In addition, the study focused exclusively on male recreational runners, which limits generalizability to females and populations with different fitness levels or training histories. Although we intentionally selected only male participants to ensure the accuracy of the inverse analysis, future research should include female participants to explore the effects of running shoes on biomechanical indices of injury risk for female runners once the validity of musculoskeletal modeling-based analysis for females is further augmented through more studies in the field of modeling. Furthermore, the musculoskeletal modeling simulation, which was used in this study, is not always perfectly accurate though it has been well-validated and is currently considered one of the most reliable non-invasive methods for estimating internal loads in humans[44,52–54]. This type of musculoskeletal modeling does not account for time-varying subject-specific parameters such as fatigue, strength capacity, or individual tissue resilience. In addition, the current study examined relatively short-term biomechanical effects; future long-term studies are needed to determine how prolonged use of different footwear types influences adaptation, injury risk, or performance outcomes overtime. Lastly, higher mechanical loading is not necessarily injurious—such loading can also contribute to beneficial adaptations. Whether the observed magnitudes of biomechanical changes directly affect injury risk remains unclear and may vary depending on individual capacity and training context[20,75]. To eventually establish causal links between footwear use and injury incidence, prospective and retrospective clinical studies, which require methodological complexity and long-term observation, are inevitably necessary.

Conclusions

In this study, we demonstrated that TARS induced FFS or MFS pattern with increased ankle plantarflexion angle and subtalar eversion angle at initial contact, while reducing ankle JRFs and muscle forces in the soleus and peroneus longus during stance. These biomechanical changes suggest that TARS alter mechanical loading in ways that may reduce the risk of Achilles tendinopathy and metatarsal stress fractures. However, the magnitude and direction of these changes were modulated by individual habitual foot strike angles. Although our findings do not directly conclude injury outcomes, they offer biomechanical insights into how modern footwear technologies affect internal joint and muscle loading patterns. These results underscore the importance of considering both structural shoe features and runner-specific gait characteristics when evaluating footwear effects on performance and running biomechanics that potentially affects injury mechanisms.

Data availability

All data are available in the main text or the supplementary materials.

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Author contributions

H.K. designed the study, performed the experiments, analyzed the data, interpreted results, and created figures; J.A. supervised the study, and acquired funding. All authors (H.K. and J.A.) wrote the paper, edited the manuscript and approved the final version.

Declarations

Competing interests

The authors declare no competing interests.

Additional information

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Correspondence and requests for materials should be addressed to J.A.

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