Accepted Manuscript (Uncorrected Proof)

Title: Lower-Limb Muscle Activity During Running in Runners with Pronated Feet Wearing Carbon-Plated and Conventional Running Shoes: A Cross-Sectional Laboratory Study

Authors: Mohsen Barghamadi[1,*], Ladan Barzegarfar[1], Ebrahim Piri[1]

1. *Department of Sport Biomechanics, Faculty of Educational Sciences and Psychology, University of Mohaghegh Ardabili, Ardabil, Iran*

To appear in: ***Physical Treatments***

Received date: 2026/07/7

Revised date: 2026/09/22

Accepted date: 2026/09/27

First Online Published: 2026/11/08

Accepted Manuscript (Uncorrected Proof)

This is a “Just Accepted” manuscript, which has been examined by the peer-review process and has been accepted for publication. A “Just Accepted” manuscript is published online shortly after its acceptance, which is prior to technical editing and formatting and author proofing. ***Physical Treatments*** provides “Just Accepted” as an optional service which allows authors to make their results available to the research community as soon as possible after acceptance. After a manuscript has been technically edited and formatted, it will be removed from the “Just Accepted” Website and published as a published article. Please note that technical editing may introduce minor changes to the manuscript text and/or graphics which may affect the content, and all legal disclaimers that apply to the journal pertain.

Please cite this article as:

Barghamadi M, Barzegarfar L, Piri E. Lower-Limb Muscle Activity During Running in Runners with Pronated Feet Wearing Carbon-Plated and Conventional Running Shoes: A Cross-Sectional Laboratory Study. ***Physical Treatments.*** Forthcoming 2027. DOI: http://dx.doi.org/10.22034/ptj.2027.639.10

DOI: http://dx.doi.org/10.22034/ptj.2027.639.10

Accepted Manuscript (Uncorrected Proof)

Abstract

Background: Carbon-plated shoes may improve running performance by altering lower-limb biomechanics through increased longitudinal bending stiffness, compliant midsole foam, and rocker geometry. However, their neuromuscular effects remain inconsistent, and little is known about how runners with pronated feet, who exhibit distinct biomechanical and muscle activation patterns, respond to this footwear.

Objective: To compare lower-limb electromyographic activity during running between runners with pronated feet who habitually used carbon-plated or conventional running shoes.

Methods: In this pilot laboratory-based comparative study, 30 recreational runners with pronated feet were evaluated while running in their habitual footwear: carbon-plated shoes (n = 15) and conventional running shoes (n = 15). Surface electromyography (1000 Hz) was used to record the activity of the tibialis anterior, medial gastrocnemius, gluteus medius, biceps femoris, semitendinosus, rectus femoris, vastus medialis, and vastus lateralis during overground running. EMG signals were band-pass filtered (10–500 Hz), notch filtered at 60 Hz, normalized to maximum voluntary isometric contraction (%MVIC), and quantified using root mean square (RMS). Muscle activation during the push-off phase was compared between groups using independent-samples t-tests. All muscle-specific analyses were considered exploratory, and a Bonferroni-adjusted significance threshold of $p < 0.00625$ was applied for the eight muscle-specific comparisons.

Results: No significant between-group differences were observed in age, height, body mass, or body mass index (all $p > 0.05$). Unadjusted analyses showed higher EMG amplitudes in the medial gastrocnemius ($p = 0.035$, $d = 0.81$), vastus medialis ($p = 0.021$, $d = 0.931$), and rectus femoris ($p = 0.039$, $d = 0.795$) during push-off in the carbon-plated footwear group compared with the conventional footwear group. However, none of these between-group differences remained statistically significant after Bonferroni correction for the eight muscle-specific comparisons (adjusted significance threshold: $p < 0.00625$). No significant between-group differences were observed for the tibialis anterior, vastus lateralis, semitendinosus, biceps femoris, or gluteus medius (all $p > 0.05$).

Conclusion: Runners with pronated feet who habitually used carbon-plated shoes showed higher unadjusted EMG amplitudes in the medial gastrocnemius, vastus medialis, and rectus femoris during push-off than those who habitually used conventional footwear. However, none of these between-group differences remained statistically significant after Bonferroni correction. These findings should therefore be considered exploratory and interpreted cautiously because the study was observational and the footwear models differed in multiple design characteristics beyond the presence of a carbon-fiber plate.

Keywords: Footwear; Electromyography; Pronated feet; Lower-limb biomechanics; Gait analysis.

Accepted Manuscript (Uncorrected Proof)

Highlights

- Evaluated lower-limb EMG activity during running in runners with pronated feet.

-Compared muscle activation during the push-off phase across distal and proximal lower-limb muscle groups.

-Nominal between-group differences were observed in selected EMG amplitudes, but these differences did not remain statistically significant after correction for multiple comparisons.

-Provides preliminary biomechanical observations in a population underrepresented in carbon-plated footwear research.

-Emphasizes the need for cautious interpretation due to the observational design and habitual footwear-related variables.

Plain Language Summary

Carbon-plated running shoes are widely used by runners, but muscle activity patterns in runners with pronated feet who habitually use these shoes remain understudied. This laboratory study compared lower-limb muscle activity between recreational runners with pronated feet who habitually used carbon-plated running shoes and those who habitually used conventional running shoes. Surface electromyography (EMG) was used to assess activity in the calf, thigh, and hip muscles during the push-off phase of running.

In the unadjusted analyses, runners in the carbon-plated footwear group showed higher EMG amplitudes in the medial gastrocnemius, vastus medialis, and rectus femoris than those in the conventional-footwear group. However, none of these between-group differences remained statistically significant after Bonferroni correction for the eight muscle-specific comparisons.

These findings provide preliminary information on between-group EMG patterns in runners with pronated feet who habitually use different footwear configurations. Because the study was observational and participants were not randomly assigned to footwear conditions, the findings should not be interpreted as evidence that carbon-plated shoes caused changes in muscle activity. Differences in other footwear characteristics, such as midsole material, geometry, cushioning, and shoe mass, as well as participant-related factors, may also have contributed to the observed differences. Further randomized within-subject studies are needed to determine the specific contribution of different footwear characteristics to muscle activity during running.

Accepted Manuscript (Uncorrected Proof)

1. Introduction:

Running is one of the most popular forms of physical activity because of its accessibility and well-established health benefits. However, repetitive mechanical loading during running contributes to a high incidence of running-related musculoskeletal injuries, making the optimization of running biomechanics a major focus of sports medicine research (1, 2). Among the external factors influencing lower-limb biomechanics, footwear has attracted considerable attention because it directly alters the mechanical interaction between the foot and the ground (3, 4). Recent advances in footwear technology have led to the widespread use of carbon-plated shoes, which combine a compliant and resilient midsole foam with a curved carbon-fiber plate to enhance running performance (5-8). The biomechanical effects of carbon-plated shoes are primarily attributed to the interaction between three key structural characteristics: increased longitudinal bending stiffness, compliant energy-returning midsole foam, and rocker geometry, rather than to the carbon plate alone (6, 8-10). Together, these features modify metatarsophalangeal and ankle joint mechanics, reduce energy dissipation at the foot, redistribute mechanical work across the lower extremity, and improve running economy in many runners (6, 8-10). Accordingly, the term carbon-plated shoes in the present study refers to the complete shoe configuration incorporating a carbon-fiber plate, rather than to the isolated effect of the plate itself. Because the present study compared commercially available footwear models with multiple differing design characteristics, any observed between-group differences should be interpreted as responses to the overall footwear configurations rather than as effects attributable specifically to the carbon-fiber plate.

Foot posture is an important intrinsic determinant of lower-limb biomechanics because it influences how loads are transferred between the foot and the supporting surface during locomotion (3, 4, 11). Among different foot postures, excessive foot pronation has received

Accepted Manuscript (Uncorrected Proof)

considerable attention owing to its effects on lower-limb alignment, mechanical loading, and neuromuscular control (11, 12). Pronated foot posture is relatively common among recreational runners, with more than one-third of this population reported to exhibit pronated feet (13). Pronated foot posture has also been reported in association with lower-limb musculoskeletal complaints, particularly involving the knee and ankle; however, these associations do not establish a causal relationship. Furthermore, pronation distortion syndrome has been reported in approximately 35.9% of amateur athletes, further emphasizing the clinical relevance of abnormal foot posture in athletic populations (14).

Biomechanically, excessive pronation is characterized by prolonged rearfoot eversion, altered medial longitudinal arch mechanics, and increased tibial internal rotation during stance, resulting in changes in lower-limb alignment and load distribution throughout the kinetic chain rather than at the foot alone (12, 15-17). These mechanical alterations are accompanied by distinct neuromuscular activation strategies involving both distal and proximal muscles responsible for dynamic stability and propulsion, including the tibialis anterior, peroneus longus, gastrocnemius, quadriceps, and gluteus medius (4, 16, 18). Given that carbon-plated footwear incorporates distinct characteristics related to longitudinal bending stiffness, rollover mechanics, and distal joint function (6, 19), these characteristics may interact with the altered biomechanics associated with pronated feet and potentially influence neuromuscular responses. However, whether foot posture modifies the neuromuscular response to different footwear configurations remains a hypothesis requiring direct investigation in runners with different foot postures. Despite the rapid expansion of research on carbon-plated shoes, several important knowledge gaps remain. Most previous studies have focused on healthy runners with neutral or unspecified foot posture and have primarily evaluated global outcomes such as running economy, metabolic cost, and lower-limb

Accepted Manuscript (Uncorrected Proof)

biomechanics rather than neuromuscular adaptations (6, 8, 20, 21). Although electromyography has occasionally been incorporated, the available evidence remains inconsistent, with studies reporting reduced, unchanged, or muscle-specific alterations in activation depending on footwear configuration and experimental conditions (2, 10, 19, 20, 22). Consequently, the neuromuscular mechanisms through which carbon-plated shoes modifies running biomechanics remain incompletely understood.

These limitations are particularly relevant for runners with pronated feet, who exhibit distinct biomechanical and neuromuscular characteristics that may influence their response to footwear interventions (4, 16, 18, 23). To our knowledge, literature regarding the neuromuscular effects of carbon-plated shoes in runners with pronated feet is extremely limited, with prior work primarily focusing on running economy (24). Therefore, it remains unknown whether runners with pronated feet exhibit neuromuscular responses to carbon-plated shoes similar to those reported in healthy runners, or whether their altered foot biomechanics modifies these responses (4, 16, 18). Addressing this question is important from both mechanistic and clinical perspectives. Characterizing muscle activation patterns in this mechanically distinct population may improve understanding of how advanced footwear technology influences neuromuscular control during running and provide evidence to guide footwear selection, injury prevention, and rehabilitation strategies for runners with pronated feet (1, 4, 18). Therefore, it remains unclear how runners with pronated feet respond to different footwear designs, and whether foot posture contributes to variability in neuromuscular responses remains an important question for future research.

Accepted Manuscript (Uncorrected Proof)

2. Materials

2.1. Participants

This pilot, observational, cross-sectional laboratory study included 30 recreational runners with pronated feet. Participants were classified into one of two groups according to their habitual footwear: a carbon-plated running shoe group (n=15; mean age: 24.47±5.30 years) and a conventional running shoe group (n=15; mean age: 23.87±3.88 years, Table 1). No footwear condition was experimentally assigned or manipulated; participants performed all running trials exclusively in their own habitual running shoes. A priori sample size was estimated using G*Power software (version 3.1.9.7, Heinrich-Heine-Universität Düsseldorf, Düsseldorf, Germany). Based on an F-test (ANOVA: Repeated measures, within-between interaction), an effect size (f=0.82) reported in a recent study, a significance level (α) of 0.05, and a statistical power (1−β) of 0.95, the minimum required sample size was calculated to be 10 participants per group (N_t=20)(18). To account for potential electromyographic (EMG) signal artifacts, potential data loss, and the exploratory nature of this pilot investigation, a total of 30 participants (15 per group) were recruited and included in the final analysis. All participants had at least three years of recreational running experience and reported running at least three times per week. Following institutional approval, volunteers were recruited from local running clubs and attended the Health Center of the University of Mohaghegh Ardabili, where they underwent a medical examination performed by a sports medicine physician before participation.

Inclusion criteria were: (1) age between 18 and 35 years; (2) recreational running experience of at least three years; (3) a minimum training frequency of three sessions per week; (4) To ensure consistency in footwear size and habitual exposure, only participants who routinely wore EU size 41 running shoes and had used their habitual footwear for at least one year were included in the study; and (5) a pronated foot posture confirmed using the Foot Posture Index (FPI-6). Pronated foot posture was assessed using the six-

Accepted Manuscript (Uncorrected Proof)

item Foot Posture Index (FPI-6). A score of ≥+6 was used to classify a foot as pronated, and participants were required to meet this criterion bilaterally. The FPI-6 assessment was performed by a sports biomechanics specialist at the University Health Center. The assessor was blinded to participants' habitual footwear group and was responsible only for the foot-posture assessment. All participants met the pronated-foot criterion in both feet, with mean FPI-6 scores of 8.3 ± 1.4 for the right foot and 8.1 ± 1.2 for the left foot. The right lower limb was selected for EMG analysis because it was identified as the dominant limb using the soccer-ball kicking test. Assessor reliability was not independently assessed in the present study. Exclusion criteria included a history of lower-extremity injury during the previous year, previous lower-extremity surgery, severe trunk deformity, withdrawal from the study, or injury occurring during the testing session (25, 26). All study procedures were performed under the supervision of a medical specialist and in accordance with the principles of the Declaration of Helsinki. Participants' privacy and confidentiality were maintained throughout the study, and written informed consent was obtained from all participants prior to data collection. Ethical approval for the study was granted by the Ethics Committee of the University of Mohaghegh Ardabili (IR.UMA.REC.1401.085).

Table 1. Demographic and Anthropometric Characteristics of the Participants.

Variable	Carbon-Plated Shoes (Mean ± SD)	Conventional Shoes (Mean ± SD)	p-value
Age (years)	24.47±5.30	23.87±3.88	0.12
Height (cm)	174.60±9.17	177.27±7.68	0.39
Body Mass (kg)	69.93±10.91	74.32±9.95	0.26
Body Mass Index (kg/m^2)	22.71±2.82	23.10±2.95	0.61

Between-group comparisons were performed using independent samples t-tests. Statistical significance was set at $p < 0.05$.

Accepted Manuscript (Uncorrected Proof)

2.2. Footwear

Two commercially available running shoe models with distinct overall design characteristics were evaluated. The carbon-plated footwear was the Nike ZoomX Vaporfly Next% 2 (Nike Inc., Beaverton, OR, USA), a racing shoe featuring a full-length curved carbon-fiber plate embedded

within a ZoomX (Pebax-based) midsole foam. This configuration has a heel stack height of 38.6 mm, a forefoot stack height of 30.9 mm, a heel-to-toe drop of 7.7 mm, and an approximate weight of 196 g for a European size 41 (Figure 1). The conventional footwear was the ASICS Novablast 4 (ASICS Corporation, Kobe, Japan), a neutral daily running shoe without a carbon-fiber plate. This model features a full-length FF BLAST™ PLUS ECO midsole and has a heel stack height of 41.5 mm, a forefoot stack height of 33.5 mm, a heel-to-toe drop of 8 mm, and an approximate weight of 260 g for a European size 41 (Figure 2). The two footwear models differed in several design characteristics, including shoe category, midsole material, stack height, mass, cushioning properties, and mechanical characteristics. Therefore, the present study compared complete footwear configurations rather than isolating the contribution of the carbon-fiber plate. The detailed technical specifications of both shoe models are presented in Table 2. The technical specifications of both footwear models were obtained from the manufacturers' official product specifications and verified against the corresponding shoe models used in the study. Before testing, all shoes were inspected by the researcher to ensure they were free from visible wear or structural damage, verifying that the midsoles, outsoles, and insoles were completely intact. Although participants routinely used the shoes corresponding to their group, they completed a standardized 5-min familiarization run on the laboratory track before data collection to adapt to the testing environment and the overground running protocol (27). Because this was an observational comparison of two commercially available footwear models, the observed between-group differences reflect the combined characteristics of the footwear, including the presence or absence of a carbon-fiber plate, midsole material, shoe mass, stack dimensions, geometry, and rocker characteristics, rather than the isolated effect of the carbon-fiber plate. Longitudinal bending stiffness and midsole compliance were not experimentally measured in the present study.

Accepted Manuscript (Uncorrected Proof)

Table 2. Technical specifications of the carbon-plated and conventional running shoes.

Characteristic	Nike ZoomX Vaporfly Next% 2	ASICS Novablast 4
Shoe category	Racing shoe	Neutral daily running shoe
Manufacturer	Nike Inc, Beaverton, OR, USA	ASICS Corporation, Kobe, Japan
Carbon-fiber plate	Full-length curved carbon-fiber plate	None
Midsole material	ZoomX (Pebax-based foam)	FF BLAST™ PLUS ECO
Heel stack height	38.6 mm	41.5 mm
Forefoot stack height	30.9 mm	33.5 mm
Heel-to-toe drop	7.7 mm	8 mm
Shoe weight	Approximately 196 g	Approximately 260 g
Shoe size used in the study	EU 41	EU 41



Figure 1. Carbon-Plated shoe.



Figure 2. Control shoe.

Accepted Manuscript (Corrected Proof)

2.3. Experimental

Prior to the experiment, the dominant lower limb of each participant was determined using the soccer ball kicking test (28). and all participants were right-leg dominant; therefore, EMG recordings and analyses were performed on the right lower limb. To control for fatigue-related alterations in running biomechanics, participants were instructed to avoid vigorous physical training for 48 h before the laboratory session. All participants underwent a standardized orientation to familiarize themselves with the laboratory environment, running walkway, and instrumentation before data collection commenced (29). Running speed was monitored using a handheld stopwatch operated by the same investigator. Participants were instructed to complete the 18-m walkway in approximately 4.0 s, corresponding to a target average running speed of approximately 4.5 m/s. A practical acceptance range of 3.8–4.2 s (approximately 4.29–4.74 m/s) was used to identify trials with substantial deviation from the target running speed. Trials outside this range were discarded and repeated. Participants initiated running before entering the measurement area and continued beyond it to minimize the effects of acceleration and deceleration on the analyzed running phase. Three successful trials were recorded per participant, and their mean values were used for analysis. A trial was deemed successful if the participant maintained a continuous running pattern at the target speed without visible gait interruptions. A 30-s rest interval was provided between trials to prevent fatigue.

Ground reaction forces were recorded using a force plate during the running trials and processed according to previously published biomechanical procedures (18). Ground reaction force signals were low-pass filtered at 20 Hz using a fourth-order zero-lag Butterworth filter. Initial contact and toe-off were identified from the vertical ground reaction force signal using a 10-N threshold. The push-off phase was defined as the period from the instant of peak vertical ground reaction force to

Accepted Manuscript (Uncorrected Proof)

toe-off. These gait-event time points were synchronized with the EMG recordings and used to segment the EMG signals for calculation of normalized muscle activity during the push-off phase. Ground reaction force variables themselves were not analyzed or reported in the present study.

Muscle electrical activity was recorded from the dominant lower limb using an eight-channel wireless EMG system (Biometrics Ltd., UK) sampling at 1000 Hz. In accordance with the SENIAM guidelines, the skin at the electrode placement sites was shaved and cleansed with 70% ethanol to reduce skin impedance. Bipolar Ag/AgCl surface electrodes (11 mm diameter, 25 mm inter-electrode distance) were secured over the muscle bellies of the selected lower-extremity muscles (Table 3) (30). For signal normalization, maximum voluntary isometric contraction (MVIC) values were obtained for each muscle (30). A built-in reference electrode integrated into the Biometrics wireless EMG system was used according to the manufacturer's specifications. For the quadriceps muscles, participants performed maximal knee extension against resistance while seated with the knee flexed at 90°. Gastrocnemius activation was assessed during maximal plantarflexion in a standing position with the knee fully extended. Maximal dorsiflexion was performed to obtain reference values for the tibialis anterior. For the hamstring muscles, maximal knee flexion was performed in a prone position with the knee flexed to 90° against examiner resistance, whereas maximal hip abduction in a side-lying position was used as the reference contraction for the gluteus medius (31-33). The raw EMG signals were processed using a band-pass filter of 10–500 Hz together with a 60 Hz notch filter to remove power-line interference. To facilitate comparison across participants, dynamic EMG amplitudes were normalized to the highest RMS value obtained during the three MVIC trials for each muscle and expressed as %MVIC. Because MVIC was obtained during standardized static contractions whereas running involves dynamic, task-specific muscle activation, normalized EMG values exceeding 100%

Accepted Manuscript (Uncorrected Proof)

MVIC can occur. Such values may indicate that the static reference contraction did not fully capture the peak activation achieved during running and may also reflect differences in contraction mode, joint position, electrode placement, and task-specific neuromuscular recruitment. Therefore, %MVIC values greater than 100% were retained rather than truncated, but they were interpreted cautiously. For each muscle, three MVIC trials were performed, with each contraction maintained for 5 s and a 60-s rest interval between trials. Standardized verbal encouragement was provided during each maximal contraction. The RMS value was calculated over the central 3-s period of each 5-s MVIC trial to minimize the influence of the onset and termination phases of the contraction. The highest RMS value obtained across the three trials was used as the reference MVIC value for normalization. Signal processing and data analysis were conducted using Datalite and MATLAB software, and the Root Mean Square (RMS) values were calculated for further statistical analysis.

Table 3. Electrode Placement Sites According to the SENIAM Protocol

Muscle	Electrode Placement Site
TA	Upper 30% of the line connecting the tip of the fibula to the medial malleolus of the ankle
MG	Most prominent region of the medial muscle belly
GM	50% of the distance along the line connecting the anterior superior iliac spine to the greater trochanter
BF	50% of the line connecting the lateral condyle of the tibia to the ischial tuberosity
ST	50% of the line connecting the medial condyle of the tibia to the ischial tuberosity
RF	50% of the distance along the line connecting the anterior superior iliac spine to the superior border of the patella
VM	Distal 80% of the line connecting the anterior superior iliac spine to the patella at approximately 55° relative to the longitudinal axis of the thigh
VL	Upper 35% of the distance along the line connecting the anterior superior iliac spine to the lateral border of the patella

Table Note: Tibialis Anterior (TA), Medial Gastrocnemius (MG), Gluteus Medius (GM), Biceps Femoris (BF), Semitendinosus (ST), Rectus Femoris (RF), Vastus Medialis (VM), And Vastus Lateralis (VL).

2.4. Statistical Analysis

Accepted Manuscript (Uncorrected Proof)

Prior to parametric analyses, all underlying statistical assumptions were verified. Data distribution was assessed for normality using the Shapiro–Wilk test and visual inspection of Q–Q plots. Homogeneity of variance between the two groups was examined using Levene's test for equality of variances; in cases where the assumption of homoscedasticity was violated, degrees of freedom and p-values were adjusted using Welch's correction. The data were also screened for extreme outliers using boxplot distributions and studentized residuals. Differences in normalized EMG amplitudes (%MVIC) between the carbon-plated footwear group (n=15) and the conventional shoe group (n=15) were evaluated using independent-samples t-tests. Statistical analysis was performed using IBM SPSS Statistics (version 26; IBM Corp., Armonk, NY, USA), with the level of significance set at $p<0.05$. Cohen's d effect sizes were calculated to quantify the magnitude of the differences between the two footwear conditions (34). All muscle-specific analyses were exploratory; no muscle was prespecified as a primary outcome. To account for multiple comparisons across the eight muscle-specific analyses, a Bonferroni-adjusted significance threshold of $p<0.00625$ was applied (0.05/8).

3. Results

No statistically significant between-group differences were observed in age ($p = 0.120$), height ($p = 0.390$), body mass ($p = 0.260$), or body mass index ($p = 0.610$). These characteristics are presented descriptively in Table 1. The mean and standard deviation of normalized EMG activity (%MVIC) for the investigated muscles during the push-off phase of running are reported in Table 4. During the push-off phase, higher EMG amplitudes were observed in the carbon-plated footwear group in the unadjusted analyses for the medial gastrocnemius (MG, $p = 0.035$, $d = 0.81$, 95% CI: −65.23 to −2.58), vastus medialis (VM, $p = 0.021$, $d = 0.93$, 95% CI: −65.68 to −6.29), and rectus femoris (RF, $p = 0.039$, $d = 0.795$, 95% CI: −54.33 to −1.48) compared with the conventional

Accepted Manuscript (Uncorrected Proof)

footwear group. However, none of these between-group differences remained statistically significant after Bonferroni correction for the eight muscle-specific comparisons (adjusted significance threshold: $p < 0.00625$).

Table 4. Mean and Standard Deviation of Normalized EMG Activity (%MVIC) of Lower Extremity Muscles During Running Under the Two Footwear Conditions.

Muscle	Phase	Control Shoe Mean ± SD	CPS Mean ± SD	d-value	p-value
TA	Push-off	46.03 ± 13.38	40.76 ± 15.91	0.359	0.335
MG	Push-off	98.49 ± 34.55	132.39 ± 48.10	0.810	0.035
VM	Push-off	123.13 ± 13.98	159.12 ± 52.41	0.931	0.021
VL	Push-off	114.20 ± 26.21	141.23 ± 46.16	0.720	0.059
RF	Push-off	123.94 ± 27.48	151.85 ± 41.32	0.795	0.039
ST	Push-off	91.93 ± 31.70	103.09 ± 38.20	0.318	0.391
BF	Push-off	97.29 ± 13.34	105.70 ± 23.66	0.438	0.241
GM	Push-off	263.58± 44.26	237.94 ± 36.64	0.631	0.095

Table Note: CPS = carbon-plated shoes. TA, Tibialis Anterior; MG, Medial Gastrocnemius; GM, Gluteus Medius; BF, Biceps Femoris; ST, Semitendinosus; RF, Rectus Femoris; VM, Vastus Medialis; VL, Vastus Lateralis. Unadjusted p values are reported. The Bonferroni-adjusted significance threshold was $p < 0.00625$ for the eight muscle-specific comparisons; therefore, none of the muscle-specific comparisons reached the corrected significance threshold.

Discussion

The present study identified observed nominally higher EMG amplitudes in the medial gastrocnemius, vastus medialis, and rectus femoris in the carbon-plated shoesgroup; however, none of these muscle-specific differences remained statistically significant after Bonferroni correction for the eight comparisons. Specifically, higher unadjusted EMG amplitudes were observed for the MG ($p = 0.035$, $d = 0.81$, 95% CI: −65.23 to −2.58), VM ($p = 0.021$, $d = 0.931$, 95% CI: −65.68 to −6.29), and RF ($p = 0.039$, $d = 0.795$, 95% CI: −54.33 to −1.48) during push-

Accepted Manuscript (Uncorrected Proof)

off. Because all three p-values exceeded the prespecified Bonferroni-adjusted threshold of 0.00625, these findings should not be interpreted as statistically confirmed between-group differences. Rather, they represent nominal differences identified in the unadjusted analyses, with substantial uncertainty regarding whether they reflect reproducible differences in muscle activity. No nominal between-group differences were observed for the tibialis anterior (TA), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF), or gluteus medius (GM) (all $p > 0.05$). Thus, the overall pattern should be regarded as exploratory and hypothesis-generating rather than confirmatory. Because all participants had pronated feet and no neutral-foot comparison group was included, the present study cannot determine whether foot posture modifies the neuromuscular response to footwear. Accordingly, the nominal differences observed in selected muscles cannot be used to establish that pronated foot posture modifies the response to carbon-plated footwear. Nevertheless, the observed pattern may provide a basis for generating hypotheses may contribute to variability in EMG responses to different footwear designs, which warrants investigation in future studies including runners with different foot postures (10, 20). Unlike most previous studies conducted in healthy runners with neutral foot posture, the present findings suggest a possible area of interest rather than providing evidence that foot posture is a determinant of muscular responses advanced footwear technology, although this possibility cannot be established from the present study (16, 18).

The most notable nominal between-group difference was the higher medial gastrocnemius EMG amplitude during push-off in the carbon-plated footwear group. This pattern differs from previous studies reporting either reduced or unchanged plantar-flexor activity in healthy runners using carbon-plated shoes. For example, Hata et al. reported a substantial reduction in medial gastrocnemius EMG while participants ran in highly cushioned carbon-plated racing shoes (22),

Accepted Manuscript (Uncorrected Proof)

whereas Beck et al. and Ghanbari et al. observed little or no change in plantar-flexor activation despite improvements in running economy and alterations in distal joint mechanics (10, 20). These findings have generally been interpreted as evidence that carbon-plated shoes may influence mechanical efficiency without consistently increasing plantar-flexor muscle activation. However, because these studies primarily included runners with neutral or unspecified foot posture, their findings may not be directly applicable to runners with pronated feet (16, 18). Importantly, the nominal MG difference observed in the present study should also be interpreted cautiously because it did not survive correction for multiple comparisons. Therefore, the apparent discrepancy with previous findings may reflect sampling variability, differences in footwear configuration, participant characteristics, or other uncontrolled factors rather than a true difference in plantar-flexor response.

A possible explanation for the nominally greater medial gastrocnemius activationobserved in the carbon-plated footwear group relates to the interaction between footwear characteristics and the biomechanical features of pronated feet. Carbon-plated shoes are characterized by greater longitudinal bending stiffness and may influence mechanics at the metatarsophalangeal and ankle joints (6, 22). Reduced metatarsophalangeal motion with stiffer or intact carbon-plated constructions has been reported across biomechanical studies (9, 19, 35). Carbon-plate geometry and placement may also modify forefoot loading and load-transfer pathways, with curved or thicker configurations tending to reduce forefoot plantar pressure and alter medial load transfer (36, 37). These mechanisms could plausibly contribute to differences in plantar-flexor EMG; however, the present data do not establish that such mechanisms occurred in our participants. Because plantar pressure, joint kinematics, and kinetics were not measured in the present study,

Accepted Manuscript (Uncorrected Proof)

the proposed mechanistic explanation remains uncertain and should be considered hypothesis-generating rather than explanatory of the observed nominal difference(6, 16, 19, 35, 38).

In pronated feet, excessive pronation has been associated with altered lower-limb kinematics and kinetics, including greater knee and hip internal rotation and altered ankle mechanics (16). Pronated-foot populations have also been reported to exhibit greater medial gastrocnemius activation under certain running conditions (16, 38). Together, these findings provide a plausible context for the nominal MG difference observed in the present study. However, they do not establish that pronation modified the response to the footwear conditions in the present study, and the underlying mechanism cannot be confirmed without concurrent kinematic and kinetic measurements. The present study also observed nominally higher unadjusted EMG amplitudes of the vastus medialis and rectus femoris in the carbon-plated footwear group. Evidence regarding quadriceps activation in carbon-plated shoes is more mixed than for plantar-flexors. Beck et al. found no change in muscle activity across seven leg muscles with greater footwear bending stiffness (20), whereas Chen et al. reported that higher longitudinal bending stiffness significantly increased activation of vastus medialis and vastus lateralis (19). Excessive pronation has also been linked to altered proximal mechanics, including knee flexion and internal rotation changes, and some authors have emphasized the importance of knee extensor function in pronated-foot populations (16). Altered distal mechanics associated with carbon-plated shoes have been proposed to redistribute mechanical demand toward more proximal joints, particularly when metatarsophalangeal motion is constrained (6, 35). Although these mechanisms could provide a plausible explanation for the nominal differences in VM and RF activity, the absence of statistical significance after Bonferroni correction means that the present study cannot distinguish these possible mechanisms from random variation or residual confounding. Therefore, the higher

Accepted Manuscript (Uncorrected Proof)

unadjusted EMG amplitudes observed for the vastus medialis and rectus femoris should be interpreted as uncertain between-group patterns rather than evidence of altered neuromuscular or compensatory strategies (19, 35). Interestingly, no nominal between-group differences were observed in the tibialis anterior, semitendinosus, biceps femoris, vastus lateralis, or gluteus medius. This pattern is broadly consistent with prior work showing that carbon-plated or stiffness-related interventions do not uniformly change lower-limb EMG (20, 39). Tibialis anterior responses appear to depend on footwear configuration and testing context: Beck et al. found no effect with greater bending stiffness (20), whereas Xu et al. found tibialis anterior activation differed by curved versus flat plate shape and fatigue state (2). Hamstring and thigh-muscle responses during running can also remain relatively unchanged across biomechanical conditions; for example, rectus femoris and biceps femoris activity was similar between rearfoot and forefoot running techniques in one SPM analysis (40). Thus, the absence of nominal differences in several muscles, together with the lack of statistically significant findings after correction, suggests that there is currently insufficient evidence for a consistent or uniform effect of the compared footwear configurations on lower-limb EMG. The inconsistent electromyographic findings across studies likely reflect differences in both footwear design and participant characteristics. Reviews conclude that carbon-plated or high-stiffness shoes show mixed effects on running economy and biomechanics, and that plate geometry, placement, foam, and other shoe features all matter (6, 41). Carbon-plated shoes do not produce consistent changes in joint power or leg stiffness across healthy-adult crossover studies (21). Participant factors such as fatigue state, strike pattern, and foot posture also change lower-limb mechanics and EMG responses (16, 35, 40).

The present findings therefore do not provide definitive evidence of a footwear-related alteration in muscle activity. Instead, the nominal differences observed in three muscles may indicate

Accepted Manuscript (Uncorrected Proof)

potentially relevant patterns that require confirmation in adequately powered and better-controlled studies. EMG responses to different footwear configurations may vary between runners and may depend on the interaction between shoe construction and runner characteristics (21, 41). However, given the corrected statistical analysis, these possibilities remain uncertain and should not be interpreted as established effects.

However, because the present study compared complete footwear configurations, these observations cannot be attributed specifically to the carbon-fiber plate. Differences in midsole properties, geometry, cushioning, mass, and other design characteristics may also have contributed to the nominal EMG differences. Moreover, because the study was observational and participants were classified according to habitual footwear use, differences between groups may also reflect participant characteristics or prior exposure to the respective footwear rather than the footwear configurations themselves. From a clinical perspective, the present findings should be interpreted cautiously. The biomechanical responses to carbon-plated shoes should not be assumed to be identical across all runners. Although some carbon-plated designs have been reported to improve running economy, the magnitude and underlying mechanisms are inconsistent and are not explained by plate stiffness alone (9). In runners with pronated feet, distinct activation patterns during propulsion are plausible given known alterations in lower-limb alignment and muscle activity (16, 38). Nevertheless, the present study does not provide statistically significant evidence that carbon-plated footwear alters EMG activity in any of the eight muscles after correction for multiple comparisons. The observed EMG differences in the present study should not be interpreted as evidence of beneficial or harmful clinical effects. Higher or lower EMG amplitude alone does not establish changes in muscle force, injury risk, pain, running performance, rehabilitation outcomes, or long-term adaptation. Whether these acute muscle activation patterns

Accepted Manuscript (Uncorrected Proof)

have meaningful functional or clinical consequences over longer periods of footwear use remains unresolved (35, 41).

The present study has several strengths. To our knowledge, it is among the first studies to examine lower-limb EMG activity in runners with pronated feet who habitually use carbon-plated footwear, addressing a population that remains underrepresented in advanced footwear research. In addition, both distal and proximal lower-limb muscles were assessed during overground running, providing a broader assessment of muscle activity than studies examining isolated muscle groups (20, 35). Several limitations should, however, be considered when interpreting the findings. First, the cross-sectional between-group design based on habitual footwear use limits causal inference and introduces the possibility of selection bias. The absence of randomization and the use of habitual footwear groups may have resulted in systematic baseline differences between groups that could not be fully controlled in the present study. Differences in habitual footwear exposure and other unmeasured participant characteristics may have contributed to the observed between-group differences. Although all participants were male and the dominant right lower limb was assessed, several potentially relevant variables, including weekly running volume, habitual running pace, foot-strike pattern, duration of habitual shoe use, accumulated shoe mileage, and other training-related characteristics, were not systematically collected or controlled. Therefore, residual confounding cannot be excluded, and the observed between-group differences in EMG amplitude may partly reflect participant characteristics other than footwear type. These sources of uncertainty are particularly important because the three nominal differences did not remain statistically significant after Bonferroni correction. Randomized within-subject crossover studies are warranted to better determine footwear-related causal effects. Although a practical acceptance range of 3.8–4.2 s was used, running speed was monitored using a handheld stopwatch rather than

Accepted Manuscript (Uncorrected Proof)

an automated timing system. Because trial-level completion times were not retained, group-specific variability in running speed could not be retrospectively quantified. Therefore, small variations in instantaneous running speed and the effects of acceleration or deceleration cannot be completely excluded. Second, this was a pilot study with a relatively small sample size, which may have limited statistical power, particularly for muscles showing moderate effect sizes but non-significant between-group differences. Accordingly, the absence of statistically significant corrected findings should not be interpreted as proof of no effect; rather, the present study cannot reliably distinguish between the possibility of small or moderate footwear-related effects and sampling variability. Conversely, the nominal p-values observed for MG, VM, and RF should not be interpreted as evidence of confirmed effects because they did not meet the prespecified corrected threshold.An additional consideration concerns the normalized EMG values exceeding 100% MVIC observed in several muscles, particularly the gluteus medius. Although the highest RMS value from three standardized MVIC trials was used as the normalization reference, the selected static MVIC tasks may not have fully elicited the maximal activation achievable during dynamic running. Dynamic muscle activation can differ from static reference contractions because of differences in joint position, contraction mode, muscle length, force direction, and task-specific neuromuscular recruitment. In addition, surface EMG normalization is sensitive to electrode placement, cross-talk, and day-to-day variability. Therefore, values exceeding 100% MVIC should not be interpreted as indicating that the muscle generated more than a physiologically meaningful '100%' activation level. Rather, they represent EMG amplitudes that were greater than the reference RMS obtained during the standardized MVIC task. This issue may have introduced additional variability into the normalized EMG measurements and should be considered when interpreting between-group differences. Third, only one commercially available carbon-plated shoe model was evaluated; therefore, the findings cannot be generalized to shoes with different plate geometries, bending stiffness, midsole

Accepted Manuscript (Uncorrected Proof)

materials, or rocker profiles. Longitudinal bending stiffness and midsole compliance were not experimentally measured in the present study, limiting our ability to determine how these mechanical properties contributed to the observed EMG differences. Although all participants had used their habitual footwear for at least one year, the exact accumulated mileage of the shoes was not systematically recorded, which may have introduced additional variability in the observed EMG responses. Fourth, only surface electromyography was collected. Consequently, muscle force, tendon behavior, joint loading, and the biomechanical mechanisms underlying nominal EMG differences could not be assessed directly. The absence of synchronized three-dimensional kinematic, kinetic, plantar-pressure, and metabolic measurements further limits interpretation of the underlying mechanisms. Finally, the study assessed EMG activity during a single laboratory session, and the influence of long-term habituation or training with these footwear types remains unknown. Taken together, these limitations indicate that the present findings should be interpreted primarily as preliminary, hypothesis-generating observations. Confirmation of any footwear-related differences in muscle activity will require larger, randomized within-subject studies with mechanically characterized footwear, standardized running conditions, and appropriate control of potential confounding variables.

Conclusion

Although unadjusted EMG amplitudes in the medial gastrocnemius, vastus medialis, and rectus femoris were nominally higher in the carbon-plated footwear group, none of these differences reached statistical significance after Bonferroni correction, precluding definitive conclusions regarding neuromuscular adaptations in runners with pronated feet. Given the observational, cross-sectional design and habitual footwear usage, these findings do not establish a causal relationship and should be regarded as exploratory data. Future randomized, within-subject longitudinal

Accepted Manuscript (Uncorrected Proof)

studies, incorporating synchronized kinematic and kinetic measurements, are required to elucidate whether specific footwear configurations fundamentally alter neuromuscular recruitment strategies in this population.

Declarations

Authors' Contributions

Mohsen Barghamadi: Conceptualization, Methodology, Supervision, Writing, Review & Editing. Ladan Barzegarfar and Ebrahim Piri: Data Curation, Formal Analysis, Investigation, Writing, Original Draft. Ebrahim Piri: Methodology, Investigation, Writing, Review & Editing.

Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Consent to Participate

Written informed consent was obtained from all participants prior to their inclusion in the study.

Consent for Publication

Not applicable.

Ethical Approval

All procedures performed in this study involving human participants were conducted in accordance with the ethical standards of the institutional research committee and with the 1964 Declaration of Helsinki and its subsequent amendments. Ethical approval was obtained from the Ethics Committee of the University of Mohaghegh Ardabili (Ethics Code: IR.UMA.REC.1401.085).

Clinical Trial Registration

Not applicable.

Accepted Manuscript (Uncorrected Proof)

Data Availability Statement

The datasets generated and/or analyzed during the current study are available from the corresponding author upon reasonable request.

Conflicts of Interest

The authors declare that they have no conflicts of interest related to this work.

Acknowledgments

The authors would like to express their sincere appreciation to all participants who voluntarily took part in this study.

Accepted Manuscript (Uncorrected Proof)

Reference

1. Teng J, Pan M, Han Y, Diaz-Cidoncha Garcia J, Yi Q, Shen S. Biomechanical insights into carbon plate geometry in running shoes in male runners: implications for sprint performance and ankle stability. Frontiers in Bioengineering and Biotechnology. 2025;13:1735951.
2. Xu Y, Zhu C, Fang Y, Lu Z, Song Y, Hu C, et al. The effects of different carbon-fiber plate shapes in shoes on lower limb biomechanics following running-induced fatigue. Frontiers in bioengineering and biotechnology. 2025;13:1539976.
3. Agresta C, Giacomazzi C, Harrast M, Zendler J. Running injury paradigms and their influence on footwear design features and runner assessment methods: A focused review to advance evidence-based practice for running medicine clinicians. Frontiers in sports and active living. 2022;4:815675.
4. Murley GS, Landorf KB, Menz HB, Bird AR. Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review. Gait & posture. 2009;29(2):172–87.
5. Hunter I, Bradshaw C, McLeod A, Ward J, Standifird T. Energetics and biomechanics of uphill, downhill and level running in highly-cushioned carbon fiber midsole plated shoes. Journal of sports science & medicine. 2022;21(1):127.
6. Ortega JA, Healey LA, Swinnen W, Hoogkamer W. Energetics and biomechanics of running footwear with increased longitudinal bending stiffness: a narrative review. Sports Medicine. 2021;51(5):873–94.
7. Reynolds SR, Hastert LM, Nodland NM, Matthews IR, Wilkins BW, Gidley AD. The effect of carbon fiber plated shoes on submaximal running mechanics in non-elite runners. Footwear Science. 2023;15(3):171–7.
8. Kobayashi EN, de Toledo RRF, de Almeida MO, Sprey JWC, Jorge PB. Metabolic effects of carbon-plated shoes: a systematic review and meta-analysis. Frontiers in Sports and Active Living. 2025;7:1710224.
9. Healey LA, Hoogkamer W. Longitudinal bending stiffness does not affect running economy in Nike Vaporfly shoes. Journal of sport and health science. 2022;11(3):285–92.
10. Ghanbari A, Vienneau J, Nigg SR, Nigg BM. Effects of selected features of advanced footwear technology on lower limb joint work. Sports Biomechanics. 2025;24(9):2707–21.
11. Galloso-Lagos MJ, González-Elena ML, Pérez-Belloso AJ, Coheña-Jiménez M, Elena-Pérez M, Muriel-Sánchez JM, et al. A short-term evaluation of foot pronation tendency in healthy recreational runners. Life. 2023;13(11):2202.
12. Dodelin D, Tourny C, L'hermette M. The biomechanical effects of pronated foot function on gait. An experimental study. Scandinavian Journal of Medicine & Science in Sports. 2020;30(11):2167–77.
13. Umer A, Sukhera S, Ahmed T, Younas RMA, Qasim M, Mubeen M. Assessment of Foot Pronation and Lower Limb Injuries Among Recreational Runners: A Cross-Sectional Study: Foot Pronation and Injuries. THE THERAPIST (Journal of Therapies & Rehabilitation Sciences). 2024:23–8.
14. Minhas HS, Hassan A, Hussain G, Chohan AUH, Shah MA, Muzaffar A, et al. Prevalence of Pronation Distortion Syndrome and Anterior Pelvic Tilt in Relation to Body Stability among Amateur Athletes: Pronation and Pelvic Tilt in Athletes. Pakistan BioMedical Journal. 2024:9–14.
15. Mei Q, Gu Y, Xiang L, Baker JS, Fernandez J. Foot pronation contributes to altered lower extremity loading after long distance running. Frontiers in physiology. 2019;10:573.
16. Farahpour N, Jafarnezhadgero A, Allard P, Majlesi M. Muscle activity and kinetics of lower limbs during walking in pronated feet individuals with and without low back pain. Journal of Electromyography and Kinesiology. 2018;39:35–41.
17. Resende RA, Pinheiro LS, Ocarino JM. Effects of foot pronation on the lower limb sagittal plane biomechanics during gait. Gait & Posture. 2019;68:130–5.
18. Piri E, Sobhani V, Jafarnezhadgero A, Arabzadeh E, Shamsoddini A, Zago M, et al. Effect of double-density foot orthoses on ground reaction forces and lower limb muscle activities during running in adults with and without pronated feet. BMC Sports Science, Medicine and Rehabilitation. 2025;17(1):54.
19. Chen H, Shao E, Sun D, Xuan R, Baker JS, Gu Y. Effects of footwear with different longitudinal bending stiffness on biomechanical characteristics and muscular mechanics of lower limbs in adolescent runners. Frontiers in physiology. 2022;13:907016.
20. Beck ON, Golyski PR, Sawicki GS. Adding carbon fiber to shoe soles may not improve running economy: a muscle-level explanation. Scientific reports. 2020;10(1):17154.
21. Giachetti Martin S, Kobayashi EN, Amaral Coelho de Azevedo L, Bonini Vieira Campanhã D, Escudeiro de Oliveira D, Baches Jorge P. Carbon plates in running shoes biomechanics: a systematic review and meta-analysis. Frontiers in Sports and Active Living. 2026;8:1764338.

Accepted Manuscript Uncorrected Proof

22. Hata K, Hamamura Y, Noro H, Yamazaki Y, Nagato S, Kanosue K, et al. Plantar flexor muscle activity and fascicle behavior in gastrocnemius Medialis during running in highly cushioned shoes with carbon-fiber plates. Journal of applied biomechanics. 2024;40(3):192–200.
23. Hajilou B, Esmaeili H, Anbarian M. Effect of foot type on electromyography characteristics and synergy of lower limb muscles during running. Scientific Reports. 2024;14(1):25221.
24. Urushihara T, Haba T, Nakagawa Y, Kawamura D, Umezaki Y, Kumazawa I. Effect of wearing carbon plate shoes on the running economy of pronated feet. Journal of Internationl Exercise Sciences. 2022;1(1):40–9.
25. Keenan M, Peabody T, Gronley J, Perry J. Valgus deformities of the feet and characteristics of gait in patients who have rheumatoid arthritis. Journal of Bone and Joint surgery. 1991;73(2):237–47.
26. Shih Y-F, Chen C-Y, Chen W-Y, Lin H-C. Lower extremity kinematics in children with and without flexible flatfoot: a comparative study. BMC musculoskeletal disorders. 2012;13(1):31.
27. Lavcanska V, Taylor NF, Schache AG. Familiarization to treadmill running in young unimpaired adults. Human movement science. 2005;24(4):544–57.
28. Shimotashiro Y, Shinya M. Quantification in shooting precision for preferred and non-preferred foot in college soccer players using the 95% equal confidence ellipse. Frontiers in Sports and Active Living. 2024;6:1434096.
29. McWalter EJ, Cibere J, MacIntyre NJ, Nicolaou S, Schulzer M, Wilson DR. Relationship between varus-valgus alignment and patellar kinematics in individuals with knee osteoarthritis. JBJS. 2007;89(12):2723–31.
30. Albertus-Kajee Y, Tucker R, Derman W, Lamberts RP, Lambert MI. Alternative methods of normalising EMG during running. Journal of Electromyography and Kinesiology. 2011;21(4):579–86.
31. Craig CE, Goble DJ, Doumas M. Proprioceptive acuity predicts muscle co-contraction of the tibialis anterior and gastrocnemius medialis in older adults' dynamic postural control. Neuroscience. 2016;322:251–61.
32. OMURO K, SHIBA Y, OBUCHI S, TAKAHIRA N. Effect of Ankle Weights on EMG Activity of the Semitendinosus and Knee Stability During Walking by Community-Dwelling Elderly. Rigakuryoho Kagaku. 2011;26(1).
33. Mercer VS, Gross MT, Sharma S, Weeks E. Comparison of gluteus medius muscle electromyographic activity during forward and lateral step-up exercises in older adults. Physical Therapy. 2009;89(11):1205–14.
34. Cohen J. A Power Primer. Pychological Bulletin, 112 (1), 155–159. 1992.
35. Luo X, Liu R, Li B. Influence of carbon-plated shoes and fatigue on lower limb biomechanics. Journal of Biomechanics. 2025:113102.
36. Song Y, Cen X, Chen H, Sun D, Munivrana G, Bálint K, et al. The influence of running shoe with different carbon-fiber plate designs on internal foot mechanics: A pilot computational analysis. Journal of biomechanics. 2023;153:111597.
37. Song Y, Cen X, Sun D, Bálint K, Wang Y, Chen H, et al. Curved carbon-plated shoe may further reduce forefoot loads compared to flat plate during running. Scientific Reports. 2024;14(1):13215.
38. Jafarnezhadgero A, Fatollahi A, Sheykholeslami A, Dionisio VC, Akrami M. Long-term training on sand changes lower limb muscle activities during running in runners with over-pronated feet. Biomedical engineering online. 2021;20(1):118.
39. Roy J-PR, Stefanyshyn DJ. Shoe midsole longitudinal bending stiffness and running economy, joint energy, and EMG. Medicine & Science in Sports & Exercise. 2006;38(3):562–9.
40. Valencia O, Weinstein A, Salas R, Guzmán-Venegas R, Arvanitidis M, Martinez-Valdes E. Temporal differences in the myoelectric activity of lower limb muscles during rearfoot and forefoot running: A statistical parametric mapping approach. European Journal of Sport Science. 2023;23(6):983–91.
41. Hill K. Advancements in Running Shoe Technology: The Impact of Carbon Fiber Plates. Research in Strength and Performance. 2025;5(1).

Accepted Manuscript (Uncorrected Proof)