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Title: Electromyographic Mean Frequency During Walking in Runners with Pronated Feet Evaluated in Carbon-Plated and Conventional Running Shoes: A Pilot Cross-Sectional Study

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Abstract

Background: Carbon-plated shoes (CPS) incorporate structural features intended to modify footwear mechanics and have been associated with differences in lower-extremity biomechanics. However, their neuromuscular characteristics during walking remain insufficiently characterized, particularly in runners with pronated feet, who may present different lower-extremity mechanical and neuromuscular patterns.

Objective: To compare the electromyographic mean frequency (MNF) of selected lower-limb muscles during walking between runners with pronated feet evaluated in carbon-plated and conventional running shoes.

Methods: Thirty recreational runners with pronated feet participated in this pilot cross-sectional study and were evaluated in two independent footwear groups: carbon-plated shoes (n=15) and conventional running shoes (n=15). Surface electromyography (EMG; 1000 Hz) was recorded from the tibialis anterior, medial gastrocnemius, gluteus medius, biceps femoris, semitendinosus, rectus femoris, vastus medialis, and vastus lateralis. EMG signals were band-pass filtered at 10–500 Hz and notch filtered at 60 Hz. MNF was calculated from the power spectrum during the loading response and push-off phases. Normality was assessed using the Shapiro–Wilk test, and homogeneity of variance was evaluated using Levene's test. Student's independent-samples t-test was used when the equal-variance assumption was satisfied; otherwise, Welch's t-test was used.

Results: No statistically significant between-group differences were detected in the reported demographic and anthropometric variables. In the principal comparison, semitendinosus MNF during loading response was higher in the CPS group than in the conventional-shoe group (p=.001, Cohen's d=1.457). Exploratory unadjusted analyses also showed higher biceps femoris MNF in the CPS group during loading response (p=.020, d=0.932) and push-off (p=.017, d=0.924). These secondary findings should be interpreted cautiously because no multiplicity correction was applied. No statistically significant between-group differences were detected for the other examined muscles.

Conclusion: Higher EMG mean frequency was observed for selected hamstring muscles in the carbon-plated shoe group during specific gait phases. These findings indicate muscle-specific differences in EMG spectral characteristics between the footwear groups rather than a generalized difference across the lower limb. Given the cross-sectional design, post hoc outcome hierarchy, absence of multiplicity correction, and differences in the overall structural characteristics of the footwear systems, the findings should be considered preliminary and require confirmation in prospectively designed studies using mechanically characterized or matched footwear conditions.

Keywords: Carbon-fiber plate footwear; Electromyography; Flat foot; Gait analysis; Hamstrings; Neuromuscular control

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Highlights

-Higher semitendinosus EMG mean frequency was observed in the carbon-plated shoe group during loading response.

-Exploratory unadjusted analyses showed higher biceps femoris MNF during loading response and push-off.

-The biceps femoris findings should be interpreted cautiously because no multiplicity correction was applied.

-No statistically significant between-group differences were detected for the other examined muscles.

-The findings provide preliminary evidence of muscle-specific differences in EMG spectral characteristics between footwear groups.

Plain Language Summary

Carbon-plated running shoes have structural characteristics that distinguish them from conventional running shoes. However, differences in the frequency characteristics of muscle electrical signals between these footwear conditions are not well understood.
This study compared the frequency characteristics of lower-limb EMG signals during walking between recreational runners with pronated feet evaluated in carbon-plated and conventional running shoes.
Thirty runners participated in the study, with 15 evaluated in each footwear group. EMG signals were collected from eight lower-limb muscles and analyzed using mean frequency, a measure describing the frequency characteristics of the EMG signal.
The carbon-plated shoe group showed higher semitendinosus mean frequency during loading response. Exploratory analyses also showed higher biceps femoris mean frequency during loading response and push-off.
These findings indicate differences in EMG spectral characteristics between the footwear groups, particularly for selected hamstring muscles. However, the study cannot determine whether these differences were caused by the carbon plate because the footwear systems differed in several structural characteristics and participants were not randomly allocated.

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1. Introduction

Human locomotion depends on the interaction between foot structure, neuromuscular function, and external mechanical constraints such as footwear (1). As the primary interface between the body and the ground, the foot plays an important role in shock absorption, adaptation to the supporting surface, and propulsion during gait (2). Consequently, foot posture and footwear characteristics have been associated with differences in lower-extremity biomechanics and muscle function during locomotion (3, 4).

Among different foot postures, excessive foot pronation has received considerable attention because of its association with lower-limb alignment, mechanical loading, and neuromuscular function (1). Pronated foot posture is common among recreational runners and has been associated with differences in lower-extremity biomechanics and lower-limb musculoskeletal complaints (5, 6). Pronation distortion syndrome has also been reported in approximately 35.9% of amateur athletes (7). Studies using the Foot Posture Index have reported associations between increased pronation scores and differences in knee and ankle joint mechanics, including joint moments, contact forces, and tibial and knee mechanics during running (8, 9).

Experimental studies have also reported associations between excessive pronation and differences in ankle and knee mechanics, tibial and hip internal rotation, joint moments, and power generation during walking (10, 11). In addition, previous EMG studies have reported differences in the activity of ankle and proximal stabilizing muscles, including the gastrocnemius, quadriceps, and gluteus medius, in relation to altered foot mechanics (10, 12, 13). Recent developments in athletic footwear have introduced carbon-plated shoes (CPS), which commonly combine compliant, energy-returning foam midsoles with embedded carbon-fiber plates and are designed with relatively high longitudinal bending stiffness (4). Previous research has reported differences in

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metatarsophalangeal joint motion, shoe deformation, and lower-limb mechanical function across footwear conditions with different structural characteristics during stance (14). However, neuromuscular findings across footwear conditions have not been consistent. Differences in activation of the vastus medialis, vastus lateralis, and intrinsic foot muscles have been reported in adolescent runners using footwear with different stiffness characteristics (15) , whereas other studies have reported no significant differences in EMG amplitude of major lower-limb muscles (16), or differences in gastrocnemius medialis activation during push-off in carbon-plated footwear (17). These findings indicate that EMG responses may vary according to muscle, footwear characteristics, locomotor task, and participant characteristics (13, 15, 16). Moreover, most previous studies have examined running rather than walking and have included runners with neutral or unspecified foot posture. Therefore, differences in neuromuscular characteristics between carbon-plated and conventional footwear in runners with pronated feet, particularly during walking, remain insufficiently characterized (4, 14-16).

Conventional EMG amplitude analysis provides information about the magnitude of electrical activity but does not fully characterize the frequency-domain properties of the EMG signal. Mean frequency (MNF) describes the central tendency of the frequency distribution of the EMG power spectrum and may be related to motor-unit conduction properties and recruitment characteristics (10, 12, 18-22). MNF therefore provides complementary information regarding EMG spectral characteristics beyond amplitude-based measures.

Despite increasing interest in the neuromuscular characteristics of advanced footwear, it remains unclear whether EMG spectral characteristics differ between carbon-plated and conventional footwear during walking in runners with pronated feet. To our knowledge, no previous study has specifically compared EMG MNF between these footwear conditions in this population.

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Therefore, the present study aimed to compare EMG mean frequency between two footwear conditions during walking in runners with pronated feet. Specifically, EMG MNF was compared during the loading response and push-off phases of gait. We hypothesized that EMG mean frequency would differ between the footwear groups in selected lower-limb muscles rather than showing generalized differences across all examined muscles.

2. Materials and Methods

2.1. Participants

This study was designed as a pilot cross-sectional laboratory study comparing lower-limb EMG mean frequency during walking between two independent footwear groups: carbon-plated shoes and conventional running shoes. An a priori power analysis was performed using G*Power version 3.1.9.7. Because directly comparable data for semitendinosus MNF during loading response were unavailable, a standardized between-group effect size of d=0.70 was assumed as a planning value, informed by the magnitude of effects reported in previous footwear-related electromyographic research (12). Under these parameters (α=0.05, $1-\beta$=0.80), a sample size of 34 participants per group (68 in total) would be required. However, because this was a pilot investigation, a convenience sample of 30 participants (15 per group) was recruited, which provided an estimated prospective power of approximately 45.7% (df=28). Therefore, this work should be considered an exploratory pilot study. Thirty recreational runners with pronated feet were ultimately recruited (mean age: 23.22±2.93 years; body mass: 72.29±10.68 kg; height: 176.43±9.61 cm). All participants had a minimum of three years of running experience and participated in running training at least three times per week. Following institutional approval, participants attended the Health Center of the University of Mohaghegh Ardabili, where a sports medicine physician conducted medical screening to confirm eligibility. Participants comprised two independent

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footwear groups: a carbon-plated shoe group (n=15) and a conventional running shoe group (n=15). Participants were evaluated in investigator-provided footwear rather than their own shoes. No random allocation or matching procedure was performed, and participants did not select the footwear condition. Pronated foot posture was confirmed using the Foot Posture Index-6 (FPI-6). Participants were classified as having pronated feet when both feet had an FPI-6 score ≥+6. Assessments were performed by an experienced biomechanical specialist. Study-specific intra-rater or inter-rater reliability was not assessed; therefore, previously published reliability of the FPI-6 is cited only as evidence regarding the instrument and not the reliability of the present assessor (25, 26). Additional inclusion criteria were age between 18 and 35 years and no history of lower-extremity injury during the preceding 12 months. The FPI-6 is a validated clinical instrument for assessment of static foot posture (23-26). All procedures adhered to the ethical principles of the Declaration of Helsinki. Written informed consent was obtained from all participants. Ethical approval was granted by the Ethics Committee of the University of Mohaghegh Ardabili (IR.UMA.REC.1401.081).

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 an independent samples t-test. Statistical significance was set at $p < 0.05$.

2.2. Footwear

In this study, two types of footwear with distinct structural characteristics were utilized to evaluate lower-limb electromyographic activity during walking. The carbon-plated shoe was the Nike

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ZoomX Vaporfly NEXT% 2, a commercially available racing shoe featuring a full-length curved carbon-fiber plate embedded within a ZoomX foam midsole based on Pebax. The carbon-fiber plate is a structural component of the shoe that has been designed to increase longitudinal bending stiffness and support propulsion. The shoe had 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 mass of 196 g in a standard size (Figure 1). The conventional footwear condition was provided using the PUMA Morphic (PUMA SE, Herzogenaurach, Germany), a commercially available neutral running shoe without a carbon-fiber plate. The PUMA Morphic featured a compression-molded EVA (CMEVA) midsole, a 100% rubber outsole, and a mesh upper, with an approximate sole height of 45.7 mm (1.8 in) according to the manufacturer. To control potential confounding factors, both pairs of shoes were inspected by the researcher before the experimental protocol to ensure insole integrity, absence of wear, and structural soundness. To ensure a standardized footwear fit, all participants wore European shoe size 41 in the footwear condition in which they were evaluated. In addition, participants completed a 5-minute familiarization period while walking in the footwear used for their evaluation before data collection to minimize the potential influence of footwear unfamiliarity (27). Walking trials were performed overground along a 10-m walkway in the biomechanics laboratory. Participants were instructed to walk at a controlled speed of approximately 1.43 m/s (10 m in 7 s) while maintaining their natural gait pattern. Three successful walking trials were recorded for each participant. Trials were considered acceptable when the EMG signals were free of substantial noise or signal artifacts and the participant demonstrated a normal walking pattern. If a trial showed substantial EMG noise, signal artifacts, or an abnormal walking pattern, the trial was repeated until three acceptable trials were obtained. The mean value across the three acceptable trials was used for subsequent statistical analysis. Because each

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participant was evaluated in only one footwear condition, footwear order randomization and counterbalancing were not applicable to the study design. EMG signals and force plate signals were synchronously recorded throughout all walking trials. Force plate signals were used only to identify gait events (initial contact and toe-off) and to segment the loading response and push-off phases. No force plate variables were included as outcome measures in the statistical analyses. Following gait-event identification, the corresponding EMG signals from each gait phase were extracted for frequency-domain analysis.

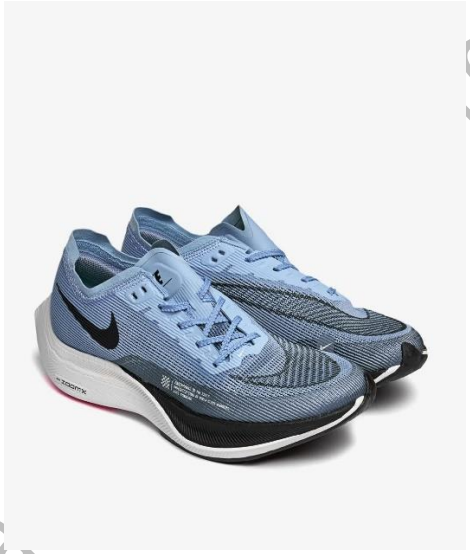


Figure 1. The carbon-plated shoe.

2.3. Experimental

Participants in the carbon-plated shoe group (n = 15) and the conventional shoe group (n = 15) completed the same experimental protocol. The experimental protocol was identical for both groups. All participants presented with pronated foot posture, and the dominant lower limb was determined using the soccer ball kicking test (28). To minimize fatigue-related effects that could influence gait patterns, participants were instructed to refrain from vigorous physical activity for

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48 hours prior to testing. Before data collection, all participants were familiarized with the laboratory environment, testing procedures, and equipment operation (29). Complete blinding during data collection was not feasible because the footwear conditions were visually distinguishable. The same investigator performed the data collection and EMG processing. However, participant data were coded before EMG processing and statistical analysis, and the EMG files were analyzed using anonymous identification codes without access to group labels during the analysis stage. Muscle electrical activity was recorded using an eight channel wireless EMG system (Biometrics Ltd., UK) with a sampling frequency of 1000 Hz. In accordance with the recommendations of the SENIAM protocol, excess hair at electrode placement sites was shaved and the skin was cleaned with 70% ethanol to reduce skin impedance. Bipolar surface electrodes (Ag/AgCl), with a diameter of 11 mm and an inter-electrode distance of 25 mm, were placed over the bellies of selected lower extremity muscles following standardized SENIAM guidelines (Table, 2) (30). MNF values were calculated from the power spectrum of the EMG signals and were reported as raw frequency values in Hertz (Hz). 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. The spectral outcome variable was the electromyographic Mean Frequency (MNF), which was calculated from the power spectrum of the processed EMG signals for each muscle during the loading response and push-off phases of gait. Signal processing was performed using DataLITE software (Biometrics Ltd., UK). Cross-talk between adjacent muscles was minimized by adhering to the SENIAM recommendations for electrode placement, positioning the electrodes over the muscle belly, and maintaining the recommended inter-electrode distance. The extracted MNF values were subsequently exported for statistical analysis using SPSS software (version 26). Because the primary outcome variable was the frequency-domain parameter MNF

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rather than EMG amplitude, no normalization to maximal voluntary contraction (MVC) or another reference condition was applied, and MNF values were analyzed as raw frequency values in Hertz (Hz).

Table 2. 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

In the revised analytical framework, semitendinosus MNF during the loading response phase was designated as the principal outcome for interpretation. The remaining muscle- and phase-specific MNF comparisons were considered exploratory. Because this hierarchy was not prospectively registered, the findings should be interpreted as exploratory and require confirmation in a prospectively specified study.Because this was a pilot and exploratory study, no formal correction for multiple comparisons, such as Bonferroni adjustment or false discovery rate control, was applied. Therefore, the exploratory findings should be interpreted cautiously because of the potential for inflated Type I error.

The normality of the data distribution was assessed using the Shapiro–Wilk test. Homogeneity of variance between the carbon-plated and conventional shoe groups was evaluated using Levene's test. Student's independent-samples t-test was used when the assumption of equal variances was

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satisfied, whereas Welch's t-test was used for comparisons in which the equal-variance assumption was not satisfied. Because each participant was evaluated in only one footwear condition, the two footwear groups were treated as independent groups, and a repeated-measures analysis was not applicable. All statistical analyses were conducted at a significance level of $p < 0.05$ using SPSS software (version 26). Additionally, effect sizes were calculated using Cohen's d index (31). To facilitate interpretation, effect sizes (Cohen's d) are reported alongside p-values.

2. Results

No statistically significant between-group differences were detected in the reported demographic and anthropometric variables, including age (p=0.12), height (p=0.39), body mass (p=0.26), and BMI (p=0.61). The mean±SD values and between-group statistical comparisons for EMG MNF are presented in Table 3. During loading response, the principal comparison showed higher ST MNF in the CPS group than in the conventional-shoe group (p=0.001, Cohen's d=1.457, mean difference = −42.81 Hz, 95% CI = −65.33 to −20.29 Hz). Exploratory unadjusted analysis also showed higher BF MNF in the CPS group during loading response (p=0.020, d=0.932, mean difference = −31.38 Hz, 95% CI = −57.20 to −5.55 Hz). This result was obtained using Welch's t-test because the equal-variance assumption was not supported by Levene's test (p<0.001). No statistically significant between-group differences were detected during loading response for TA (p=0.897), MG (p=0.164), VL (p=0.110), VM (p=0.599), RF (p=0.395), or GM (p=0.639). During push-off, exploratory unadjusted analysis showed higher BF MNF in the CPS group (p=0.017, d=0.924, mean difference = −29.34 Hz, 95% CI = −53.09 to −5.58 Hz). The equal-variance assumption was not statistically rejected for this comparison (Levene's p=0.076); therefore, Student's independent-samples t-test was used. No statistically significant between-group differences were detected during push-off for TA (p=0.799), MG (p=0.434), VM (p=0.341), VL

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(p=0.210), RF (p=0.591), ST (p=0.872), or GM (p=0.283). The secondary BF findings should be interpreted cautiously because no formal multiplicity correction was applied.

Table 3. Mean and Standard Deviation of EMG Mean Frequency (MNF) Values of Lower Extremity Muscles During Walking Under Two Footwear Conditions (Hz)

Muscle	**Phase**	**Control Shoe Mean ± SD**	**CPS Shoe Mean ± SD**	**p-value**	**Cohen's d**
TA	Loading response	93.20± 17.09	93.90±11.49	0.897	0.048
MG	Loading response	93.02± 18.46	84.09± 15.63	0.164	0.522
VM	Loading response	62.52 ± 11.73	70.85 ± 15.62	0.110	0.603
VL	Loading response	66.73 ± 13.90	70.00 ± 19.38	0.599	0.194
RF	Loading response	68.19 ± 17.69	76.26 ± 31.55	0.395	0.316
ST	Loading response	81.46 ± 15.44	124.27 ± 38.59	**0.001***	**1.457**
BF	Loading response	81.13 ± 17.11	112.50± 44.42	**0.020 ***	**0.932**
GM	Loading response	91.41 ±19.10	87.96 ± 20.72	0.639	0.173
TA	Push-off	87.19 ± 15.66	88.80 ± 18.55	0.799	0.094
MG	Push-off	111.06 ± 93.86	91.58 ± 14.50	0.434	0.29
VM	Push-off	70.33 ± 20.88	64.31 ± 11.99	0.341	0.354
VL	Push-off	66.88 ± 23.50	57.70 ± 14.65	0.210	0.469
RF	Push-off	74.43 ± 27.79	79.36 ± 21.51	0.591	0.198
ST	Push-off	91.92 ± 27.10	90.43 ± 22.97	0.872	0.059
BF	Push-off	76.58 ± 18.60	105.92 ± 40.87	**0.017 ***	**0.924**
GM	Push-off	80.79 ± 19.44	88.39 ± 18.50	0.283	0.401

Table Note: CPS = carbon plated shoes. Tibialis Anterior (TA), Medial Gastrocnemius (MG), Gluteus Medius (GM), Biceps Femoris (BF), Semitendinosus (ST), Rectus Femoris (RF), Vastus Medialis (VM), And Vastus Lateralis (VL). *: $p<0.05$.

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3. Discussion

Between-group comparisons showed higher EMG mean frequency in the semitendinosus during loading response and in the biceps femoris during selected gait phases among runners with pronated feet evaluated in carbon-plated footwear compared with conventional running footwear.

The findings showed a selective between-group difference in MNF for the semitendinosus and biceps femoris rather than a generalized difference across the lower limb. Specifically, in the principal comparison, the semitendinosus exhibited higher MNF during the loading response phase in the CPS group ($p = 0.001$, approximately 53%). In exploratory unadjusted analyses, the biceps femoris showed higher MNF in the CPS group during loading response ($p = 0.020$, approximately 39%) and push-off ($p = 0.017$, approximately 38%). Because no formal multiplicity correction was applied, these BF findings should be interpreted cautiously as secondary exploratory findings rather than as confirmatory evidence. In contrast, no statistically significant between-group differences in MNF were detected for the tibialis anterior, medial gastrocnemius, vastus medialis, vastus lateralis, rectus femoris, or gluteus medius. This muscle-specific pattern indicates differences in EMG spectral characteristics between footwear groups that were not generalized across all examined muscles (32, 33). The observed higher MNF should be interpreted as differences between two complete footwear systems rather than as a direct indicator of greater muscle activation, because MNF is influenced by motor-unit conduction properties and recruitment characteristics (34, 35). Differences in MNF may reflect changes in the electrophysiological characteristics of the EMG signal and may be associated with motor-unit recruitment, discharge behavior, and muscle fiber conduction velocity (36). However, MNF alone cannot determine whether the observed difference resulted from recruitment of faster-conducting motor units, changes in motor-unit firing behavior, or other physiological factors. Therefore, the higher hamstring MNF observed in the CPS group should be interpreted as a difference in EMG spectral characteristics, while the underlying motor-unit mechanisms remain uncertain. The higher hamstring MNF observed during the loading response phase may reflect differences in the mechanical and neuromuscular demands associated with the two footwear conditions in

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individuals with pronated feet. During this phase, the lower limb must rapidly accept body weight while simultaneously controlling movements occurring at the ankle, knee, and hip joints (3, 10). Previous research has shown that footwear characteristics, including bending stiffness, can modify the interaction between the foot and ground, shoe deformation, and force transmission during stance (37). Previous studies have shown that footwear mechanical characteristics, including bending stiffness, may influence lower-limb muscle–tendon behavior during running (37). However, because longitudinal bending stiffness and other mechanical properties were not directly measured in the present study, these findings provide only contextual support and cannot establish that stiffness was responsible for the observed differences in EMG MNF. In individuals with pronated feet, these mechanical considerations may be particularly relevant because excessive foot pronation has been associated with increased tibial internal rotation, and a strong coupling between rearfoot motion and tibial rotation has been demonstrated during gait-related tasks (38). Given the established contribution of the hamstrings to knee and limb function during gait, one possible explanation is that the observed difference in MNF may be related to differences in the neuromuscular demands associated with the two footwear conditions. Within the revised analytical framework, semitendinosus MNF during loading response was treated as the principal outcome for interpretation. However, because this hierarchy was not prospectively registered, the finding should be regarded as exploratory and requires confirmation in a prospectively specified study. Specifically, the semitendinosus exhibited higher MNF during the loading response phase in the CPS group ($p = 0.001$, approximately 53%). In exploratory unadjusted analyses, the biceps femoris showed higher MNF during loading response ($p = 0.020$, approximately 39%) and push-off ($p = 0.017$, approximately 38%). Because no formal multiplicity correction was applied, these BF findings should be interpreted cautiously as secondary exploratory findings rather than as

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confirmatory evidence. However, the mechanical demands of the lower limb were not directly measured in the present study. Nevertheless, the proposed relationship between footwear condition, pronation-related tibial rotation, and hamstring MNF was not directly tested. Tibial rotation, joint kinematics, joint kinetics, and ankle–foot mechanics were not measured; therefore, these biomechanical pathways should be regarded as hypothetical explanations rather than demonstrated mechanisms (33). This interpretation is consistent with the biomechanical role of the hamstrings in resisting rotational knee movements (33); however, tibial rotation, knee kinematics, and joint moments were not directly measured in the present study. Therefore, this mechanism should be considered a plausible interpretation rather than a confirmed pathway.

During the push-off phase, the biceps femoris showed higher MNF in an exploratory unadjusted analysis. Because no formal multiplicity correction was applied, this finding should be interpreted cautiously and considered preliminary. This finding may be related to the functional roles of the biceps femoris in knee stabilization, hip extension assistance, and coordination of stance-limb stability during gait (39, 40). muscle–tendon behavior, including gastrocnemius medialis fascicle shortening and Achilles tendon energy behavior (17, 37). Previous research has also reported reduced gastrocnemius EMG activity during push-off when using carbon-plated footwear (17). Previous findings suggest that footwear characteristics may be associated with differences in distal muscle–tendon function. However, the previous studies primarily focused on running conditions and plantar-flexor muscles; therefore, their findings cannot be directly extrapolated to the higher biceps femoris MNF observed during walking in the present study. Accordingly, the exploratory BF finding observed here should not be interpreted as direct evidence of a specific proximal neuromuscular adaptation caused by the carbon plate. The higher biceps femoris MNF observed in the present study may therefore represent a difference in EMG spectral characteristics between

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footwear groups that is compatible with differences in proximal neuromuscular demands during walking, rather than direct evidence of a change in motor-control strategy or energy utilization attributable to the carbon plate (41). Because joint kinematics and kinetics were not measured, the specific biomechanical pathway underlying this spectral difference cannot be established from the present data.

An important characteristic of the present findings is the selective nature of the between-group differences in EMG spectral characteristics. The absence of statistically significant differences in MNF for the tibialis anterior, medial gastrocnemius, quadriceps muscles, and gluteus medius does not establish that these muscles were unaffected, particularly given the relatively small sample size and the multiple comparisons performed. This pattern may be considered in relation to differences in the functional roles of the examined muscles. The tibialis anterior and medial gastrocnemius primarily contribute to sagittal-plane ankle mechanics, while the quadriceps muscles contribute mainly to knee extension and weight acceptance (10, 35). The gluteus medius primarily contributes to pelvic stability in the frontal plane (41). In contrast, the hamstrings have important roles in knee stability and control of tibial rotational movements. Therefore, the higher ST MNF observed in the principal comparison, together with the exploratory unadjusted BF findings, may indicate a possible muscle-specific difference in EMG spectral characteristics between the footwear groups. However, the BF findings require cautious interpretation because they were obtained from secondary analyses without multiplicity correction. Whether these differences were related to pronation-associated mechanical demands cannot be determined from the present data (33). This interpretation remains tentative because the present study did not directly assess the mechanical demands or neuromuscular mechanisms underlying the observed spectral differences.

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These findings provide preliminary evidence of muscle-specific differences in EMG mean frequency between the two footwear conditions rather than a generalized difference across the examined muscles. Comparison of the present findings with previous carbon-plated footwear studies indicates that reported neuromuscular responses to advanced footwear may differ according to muscle, locomotor task, and assessment method. The current results are consistent with Hata et al. (2024), who reported muscle-specific effects of carbon-plated footwear, particularly changes in gastrocnemius function during push-off, supporting the concept that footwear modifications may selectively influence muscles according to their mechanical roles (17). Similarly, Xu et al. (2025) demonstrated that variations in carbon-fiber plate design were associated with differences in tibialis anterior activation and lower-limb biomechanics, supporting the potential relevance of plate characteristics to neuromuscular coordination (42). In contrast, Beck et al. (2020) found no significant changes in EMG amplitude across seven lower-limb muscles after increasing footwear stiffness (16), and Lee et al. (2026) reported no interaction between advanced footwear technology and thigh-muscle metabolic responses. These discrepancies may be related to differences in locomotor task, because previous carbon-plated footwear studies have predominantly investigated running rather than walking (16, 17, 42, 43). Furthermore, differences in participant characteristics, because previous studies did not specifically investigate individuals with pronated feet, and differences in outcome measures, as previous research mainly focused on EMG amplitude, kinetics, or metabolic responses rather than MNF, may also contribute to variations among studies. Therefore, the current findings should not be considered contradictory to previous literature but rather as preliminary evidence extending the literature by describing muscle-specific differences in EMG mean frequency during walking (16, 17, 42).

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Several limitations should be considered when interpreting the present findings. First, blinding of both participants and the investigator responsible for EMG data collection was not feasible because the carbon-plated and conventional running shoes were visually and structurally distinguishable. Consequently, the possibility of performance and observer bias cannot be completely excluded. Second, the proposed mechanisms are based on indirect evidence because tibial rotation, knee kinematics, joint moments, and foot–ankle mechanics were not directly measured. Third, MNF is an indirect neuromuscular indicator and cannot independently determine muscle fiber type or specific motor-unit recruitment strategies (35, 36). In particular, the observed upward shift in MNF should not be interpreted as direct evidence of faster motor-unit recruitment, increased recruitment of fast-twitch fibers, or a specific motor-control strategy, because these mechanisms cannot be distinguished using MNF alone. Although the walking protocol was relatively short and substantial localized hamstring fatigue was unlikely, fatigue-related effects cannot be completely excluded without direct physiological assessment.

Fourth, because this was a pilot cross-sectional laboratory study, the findings should be interpreted with caution and confirmed in larger studies with more comprehensive biomechanical assessments. Although achieving 80% statistical power for an assumed effect size of d=0.70 would require 34 participants per group (68 total), our pilot sample was limited to 15 participants per group (N=30), yielding an estimated prospective power of only 45.7%. Consequently, the study was underpowered to detect small-to-moderate effects across the examined muscles and gait phases, substantially increasing the risk of Type II errors. The absence of statistically significant differences in the remaining muscles should therefore be interpreted with caution and not as definitive evidence of an absence of effect. No statistically significant between-group differences were detected in the measured demographic and anthropometric variables; however, these

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nonsignificant findings do not establish baseline equivalence or exclude residual confounding. The possibility of residual confounding from unmeasured factors, such as individual training characteristics, pronation severity, gait variability, or previous experience with carbon-plated footwear, cannot be completely excluded.

Fifth, the carbon-plated and conventional shoes were not fully matched for mass, stack height, midsole composition, cushioning characteristics, and longitudinal bending stiffness. Consequently, the observed between-group differences cannot be attributed specifically to the carbon-fiber plate or to longitudinal bending stiffness and may reflect the combined characteristics of the overall footwear systems. Future studies should use mechanically matched footwear conditions that differ primarily in plate characteristics and should directly measure relevant mechanical properties such as longitudinal bending stiffness.

Finally, because multiple statistical comparisons were performed across several muscles and gait phases without formal adjustment for multiplicity, the possibility of Type I error cannot be completely excluded. Accordingly, the BF loading-response and push-off findings should be regarded as exploratory and unadjusted secondary findings rather than confirmatory evidence. In addition, the distinction between the principal and exploratory outcomes was introduced in the revised analytical framework rather than prospectively registered before data analysis. Therefore, the principal ST finding should also be interpreted as exploratory and requires confirmation in a future study with a prospectively specified outcome hierarchy. The relatively small sample size may have limited the statistical power to detect small or moderate differences in muscles that did not reach statistical significance. Comparisons with previous studies should also be made cautiously because most research on carbon-plated footwear has been conducted during running rather than walking.

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From a practical perspective, the present findings may provide preliminary information for understanding differences in muscle-specific EMG spectral characteristics between footwear conditions during walking in runners with pronated feet. However, these findings are not sufficient to support clinical footwear prescription or performance recommendations, and such applications require confirmation through studies incorporating kinematic, kinetic, and functional outcomes. Nevertheless, the main novelty of the present study is that it examined frequency-domain EMG characteristics of the hamstring muscles during walking in runners with pronated feet evaluated in carbon-plated and conventional running shoes. Taken together, these preliminary findings suggest a possible muscle-specific difference in EMG mean frequency between footwear groups, primarily involving the hamstrings, rather than a generalized difference across lower-limb muscles.

Conclusion

Between-group comparisons showed higher EMG mean frequency in the semitendinosus during loading response and in the biceps femoris during selected gait phases among runners with pronated feet evaluated in carbon-plated footwear compared with conventional running footwear. The findings indicate muscle-specific differences in EMG spectral characteristics rather than a generalized difference across the lower limb. The biceps femoris findings were obtained from exploratory unadjusted comparisons and should therefore be interpreted cautiously. Because the study was cross-sectional, participants were not randomly allocated, the outcome hierarchy was introduced during manuscript revision rather than prospectively registered, and the footwear conditions differed in several structural characteristics, the findings cannot establish a causal effect of carbon plating or longitudinal bending stiffness. Future studies should use prospective outcome specification, randomized or crossover designs, mechanically characterized or matched footwear

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conditions, larger samples, and direct assessment of kinematic, kinetic, and footwear-mechanical variables.

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 and Aysan Ghahremani: Methodology, Investigation, Writing, Review & Editing.

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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.1401.081).

Clinical Trial Registration

Not applicable.

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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.

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