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Research Article

Long-term Effect of Training on Different Surfaces on Knee Muscular Co-Contraction During Running in Individuals with Over-Pronated Feet

AmirAli Jafarnezhadgero[1*], Fariborz Imani[1], Raha Noroozi[1], Hamed Sheikhalizadeh[1]

[1] Department of Sports Biomechanics, Faculty of Educational Sciences and Psychology, University of Mohaghegh Ardabili, Ardabil, Iran

***Corresponding Author**: AmirAli Jafarnezhadgero, University of Mohaghegh Ardabili, Ardabil, P.O-Box 56199-11367, Tel: +98 9105146214, Email: amiralijafarnezhad@gmail.com.

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Abstract

Introduction: Over-pronated feet (OPF) are associated with altered lower-limb mechanics and elevated injury risk during running. Long-term training on different sport surfaces may modulate neuromuscular control, including knee muscular co-contraction; however, surface-specific adaptations in individuals with OPF remain insufficiently characterized. To determine the long-term effects of training on natural grass, artificial turf, and synthetic surfaces on knee muscular co-contraction during running in individuals with OPF.

Method: In this randomized controlled trial (IRCT20170806035517N5), thirty-two adults (aged 18–30 years) with clinically diagnosed OPF were randomly allocated to one of three intervention groups (natural grass, artificial turf, or synthetic surface) or a no-training control group. The intervention groups completed a supervised, eight-week running program with standardized frequency and progression. Surface electromyography (sEMG) was used to quantify knee muscular co-contraction during treadmill running at a controlled speed. General co-contraction and directed flexor-extensor co-contraction indices were computed over discrete stance sub-phases (heel-contact, mid-stance, and push-off) using established processing pipelines. Assessments were performed pre- and post-intervention by the same blinded operators. Group-by-time effects were examined for primary outcomes, with alpha set at 0.05.

Result: No significant changes emerged for general knee co-contraction across time or between groups. In contrast, significant group-by-time interactions were observed for directed flexor-extensor co-contraction during mid-stance ($p = 0.035$) and push-off ($p = 0.020$), indicating that training induced surface-specific neuromuscular adaptations rather than uniform joint stiffening. A main effect of time was also noted at heel-contact ($p = 0.050$), consistent with a generalized training-related modulation early in the stance phase. Collectively, these patterns suggest that extended exposure to distinct surfaces selectively re-tunes knee muscle coordination strategies in individuals with OPF without globally increasing co-contraction.

Conclusion : Eight weeks of running training on different surfaces prompted targeted adjustments in knee flexor-extensor coordination in runners with OPF, while general co-contraction remained unchanged. Although such adaptations may be beneficial for control, they might not fully mitigate injury risk in this at-risk population. Individualized surface selection alongside integrative neuromuscular conditioning should be considered in rehabilitation and return-to-running planning for individuals with OPF.

Keywords: Artificial Grass, Co-Contraction, Electromyography, Natural Grass, Pronation, Synthetic Surface.

Introduction

Over-pronation of the foot (OPF) is one of the most common problems affecting the ankle joint, stemming from biomechanical changes and inefficiencies that can contribute to injury in the ankle area.[1, 2] The prevalence of OPF in adults is between 20% and 23%.[3] Any biomechanical changes in the structure of the ankle, including OPF, can contribute to injury of the ankle joint and other body joints, including the spine.[4] OPF is an abnormality that causes a decrease in the height of the medial longitudinal arch during weight-bearing. This problem creates a deformity in the structure of the ankle. In this abnormality, the head of the talus and navicular bones are inclined inward, which can eventually lead to OPF. This complication imposes increased mechanical load on the ankle, knee, and pelvic girdle joints.[5]

In OPF, the medial longitudinal arches collapse (navicular bone loss) because these arches play an important role in absorbing shocks from activities such as walking, running, and jumping.[6] Therefore, the collapse of these arches can pose a serious risk to athletes, especially runners and field athletes. Based on the evidence, lower limb abnormalities such as OPF can negatively affect the biomechanics of human movements such as running and lead to muscle co-contraction, potentially as a compensatory mechanism to enhance joint stability, as well as unstable symptoms in lower limb joints (such as the knee).[7] In addition, foot deformities can lead to muscle imbalances in the lower limbs, which in turn can affect walking and running patterns. These effects include changes in stride length, step width, and walking speed.[8, 9] These changes often occur as adaptations or adjustments of the body to compensate for the deformity and maintain stability and function during movement.[8] These gait alterations may necessitate changes in muscular stabilization strategies at the knee, including modulations in co-contraction. The type and severity of foot deformity determine the magnitude of change in walking and running patterns.[10]

Researchers have reported that in individuals with OPF, due to faster contact of the inner part of the foot compared to the normal state, the necessary functions of the foot are not performed properly.[11] Additionally, people with OPF often exhibit altered electrical activity in the lower limb muscles. According to the kinematic chain theory, weakness in the performance of one movement segment can affect other movement segments as well.[12] One possible cause of changes in muscle function and activity during different phases of the gait cycle in people with OPF is the alteration in the length-tension relationship in the lower limb area, which can lead to changes in muscle function.

Running is one of the most popular physical activities worldwide and is typically performed on various surfaces, which may be associated with overuse injuries.[13] However, all available evidence supporting this association remains inconclusive.[14, 15] Running is often performed on various surfaces, including grass, woods, dirt roads, city streets and sidewalks, synthetic surfaces, stadium tracks, and parquet floors in sports halls. These surfaces differ in smoothness, stiffness, elasticity, and other properties, and may induce specific neuromuscular responses to adjust appropriate stiffness, potentially through modulation of muscle co-contraction.[16] Incidence rates of running-related injuries (RRIs) vary depending on the population, ranging from 3% to 85% and 2.5 to 33 injuries per 1,000 hours of activity.[17, 18] Approximately 87% of RRIs occur in the knee, leg, foot, and ankle areas, and abnormal foot positioning is one of the main reasons for these injuries.[19]

As mentioned, OPF has been reported as one of the main risk factors for injuries in runners, such as medial tibial stress syndrome, Achilles tendon injuries, and plantar fasciitis.[20] Dynamic stabilization of the medial longitudinal arch depends on the activity of several muscles, including the triceps surae, peroneal, tibialis posterior, and tibialis anterior during walking and running.[21] Evidence shows that individuals with OPF exhibit increased activity of some leg muscles (tibialis posterior, tibialis anterior, toe flexors, gastrocnemius medialis, and gluteus medius) and decreased activation of the evertor muscles compared to individuals with normal feet.[22] Research has shown that as surface stiffness increases, adaptive changes occur in lower limb kinematics.[23] These changes include a reduction in hip and knee flexion at initial contact and a reduction in maximum hip flexion. Additionally, it has been observed that runners who strike the ground with their rearfoot show greater pronation and plantarflexion when running on a harder surface such as concrete or asphalt than on a softer surface such as grass or synthetic rubber.[13, 24, 25]

Farahpour et al. reported that OPF increases pressure and load on the lower limb joints from the ground level during walking.[26, 27] Their research showed that during walking, the calf muscles of individuals with OPF are more active than in those with normal foot structure.[28] Additionally, increased activity of the evertor muscles has been reported in individuals with OPF. These alterations in distal muscle activity may, in turn, influence proximal joint mechanics and stabilization strategies, such as knee muscle co-contraction.[29]

In recent years, the popularity of artificial turf surfaces has increased, with an estimated 6,000 installations in North America and 1,000–1,500 new installations occurring annually.[30] Appropriate sports surfaces are among the most important pieces of equipment.[31, 32] Various factors such as impact absorption, friction, and energy loss are considered when selecting playing surfaces.[32] Among these factors, shock absorption is regarded as a key factor in injury prevention.[32] Some mechanical properties of artificial sports surfaces may be related to acute and chronic sports injuries. Potential mechanisms for different injury patterns on artificial turf compared to natural grass include torque, rotational stiffness, shoe-surface interaction, and shock absorption.[32]

Feehery et al. conducted a study comparing running on asphalt, concrete, and natural grass.[33] They found that running on concrete resulted in a shorter time to reach the first peak vertical force compared to grass and asphalt.[33] However, a higher first peak vertical force was observed on grass. The researchers suggested that injury may occur in people running on hard surfaces due to the rapid transmission of shock waves through the body, which can limit the body's ability to neutralize high-frequency shock waves as speed increases.[34] Wang et al. studied lower extremity muscle activity while running on a treadmill and compared it to other surfaces such as cement, natural grass, and artificial surfaces.[35] Their findings showed that lower limb muscle activity during running on different surfaces exhibited significant changes, which were attributed to the kinematic adaptation of the body to the running surfaces.[36]

Murley et al.[37], using electromyography, showed that individuals with OPF had a higher percentage of maximum EMG amplitude for the tibialis anterior muscle during ground contact (during walking) and for the tibialis posterior muscle during the mid-stance phase compared to the normal group. On the other hand, this group exhibited less activity in the peroneus longus muscle during the stance phase, both in walking and running.[10] This reduced activity in the peroneus longus muscle, which is crucial for controlling and stabilizing foot inversion during walking—especially on uneven surfaces—could exacerbate instability in individuals with OPF, potentially necessitating compensatory strategies at more proximal joints. The peroneal muscles are responsible for controlling and stabilizing foot inversion during walking, particularly on uneven surfaces.[38] Indeed, different surface characteristics are expected to alter the activation patterns of calf muscles during running.[13] The role of muscle activation patterns during running on different surfaces has been mostly investigated in controlled laboratory settings or on treadmills, and there is a lack of data for running on outdoor surfaces.[10]

The researchers did not identify any study that determined the control mechanisms of foot and knee joint movement through agonist-antagonist muscle co-contraction during training on artificial grass, natural grass, and synthetic surfaces in individuals with OPF. Therefore, this study aimed to compare the effects of eight weeks of training on artificial grass, natural grass, and synthetic surfaces on knee joint co-contraction during running in individuals with OPF.

Methods

Study Design

This study was conducted as an assessor-blinded randomized controlled trial using an envelope concealment method for participant allocation. Both participants and examiners were unaware of the group assignments, ensuring a blinded setup for assessments; however, participants could not be blinded to the training surface intervention itself. A power analysis using G*Power, based on expected variance in ground reaction force (GRF) variables from pilot data or relevant literature, indicated that 28 participants were necessary to achieve a statistical power of 0.80, with an effect size of 0.80 and an alpha level of 0.05. This sample size was also anticipated to be adequate for detecting meaningful changes in the primary EMG-derived co-contraction variables.

Participants

Thirty-two male participants aged 18–30 years with diagnosed OPF were recruited from the University of Mohaghegh Ardabili in Ardabil, Iran. They were randomly assigned to four groups (natural grass, artificial grass, synthetic surface, and control) (Figure 1). To determine the dominant limb, a kicking ball test was used. Inclusion criteria included: age 18–30 years, navicular drop greater than 10 mm, rearfoot eversion greater than 4°, and a Foot Posture Index over 10. Navicular drop was measured from non-weight-bearing compared to static standing. Exclusion criteria included: history of trunk and/or lower limb surgery, orthopedic conditions (excluding OPF), history of fractures, and limb length difference greater than 5 mm. The study procedures were explained to all participants, and written informed consent was obtained. The study protocol was approved by the local ethics committee of Ardabil Medical Sciences University (IR.SSRC.REC.1400.08) and registered with the Iranian clinical trial organization (IRCT20170806035517N5). The study adhered to the latest version of the Declaration of Helsinki.

Intervention

The intervention groups participated in a supervised training program for eight weeks, with three sessions per week conducted on their allocated surfaces (natural grass, artificial turf, or synthetic). Each training session, lasting approximately 60 minutes, began with a 10-minute warm-up that included dynamic stretches and submaximal running. The main segment of the protocol was designed to simulate athletic demands and consisted of several components: An Illinois agility test (3 sets), 3 sets of a 40-meter sprint run, and an agility T-test (3 sets), all interspersed with structured rest periods of 2 to 5 minutes. Additionally, participants performed a beep test consisting of 20–25 shuttles and a 6-minute continuous running bout consistent with the Cooper test protocol. Each session concluded with a 5-minute cool-down period involving slow jogging and static stretching for the lower limb muscles. Participants in the control group were instructed to maintain their habitual physical activity levels and refrain from any new structured exercise programs during the study period.

Training Surfaces

The intervention groups trained on three distinct surface types: fourth-generation artificial turf (20 mm pile height), natural grass (20 mm blade height), and synthetic rubber flooring (20 mm thickness). All surfaces were installed over a standard concrete base and were regularly maintained to ensure uniformity throughout the eight-week intervention period. The control group did not engage in surface-specific training and continued their habitual daily physical activities without any structured surface exposure.

Electromyography (EMG) Data Collection

Knee muscle activity was evaluated using an 8-channel electromyography (EMG) system (Biometrics DataLITE System, UK) with surface electrodes. The muscles assessed included knee extensors (quadriceps: rectus femoris, vastus medialis [VM], vastus lateralis [VL]) and knee flexors (hamstrings: biceps femoris, semitendinosus). Co-contraction between VM and VL was also specifically analyzed. Following SENIAM recommendations, the skin over the selected muscles was shaved and cleaned with 70% ethanol. The skin was then gently abraded before electrode placement to reduce impedance. Electrodes (Ag/AgCl, pre-gelled, 20 mm inter-electrode distance) were placed parallel to the muscle fibers. Raw EMG signals were digitized at a sampling rate of 1000 Hz. Noise was filtered out using a 450 Hz low-pass filter, a 10 Hz high-pass filter, and a 50 Hz notch filter. Data were transmitted via Bluetooth to a computer for analysis.

Gait Event Detection

EMG signals were synchronized with footswitches for the recognition of gait events. Footswitches placed under the heel and forefoot were used to determine gait cycle events: heel contact (HC), mid-stance (MS), and push-off (PO). The third stride signal after running initiation was typically analyzed to ensure steady-state gait, based on the quality and consistency of signals from the footswitches.

Testing Protocol

Before data collection, participants ran a few steps in the laboratory to identify and address any limitations or discomfort caused by the electrodes. All participants then performed three running trials during pre-test and post-test assessments. These running tests were conducted barefoot. The average of these three trials was used for data analysis.

EMG Data Processing

For EMG data analysis, Biometrics DataLITE software was used to process raw EMG signals. To normalize the EMG signals, root mean square (RMS) values for each muscle during running were calculated for specific gait phases and expressed as a percentage of the value obtained during a maximum voluntary isometric contraction (MVIC). MVICs were performed for each muscle group before the running trials. For quadriceps MVIC, participants performed maximal isometric knee extension against fixed resistance. Each MVIC was held for approximately 5 seconds. The highest average RMS EMG over a 1-second window during these MVIC trials was used as the MVIC value for normalization.

Co-Contraction Calculations

The following equations were used to determine both directional co-contraction and general co-contraction values during different running phases.

For directional co-contraction (which can be positive or negative, as shown in Tables 1 and 3):

If agonist mean EMG > antagonist mean EMG:

$$Directed\ co-contraction\ =\ 1\ _{-}\frac{(antagonist\ mean\ EMG)}{(agonist\ mean\ EMG)}$$

Else

$$Directed\ co-contraction\ =\ 1-\frac{(agonist\ mean\ EMG)}{(antagonist\ mean\ EMG)}$$

For General Co-contraction:

$$General\ co-contraction$$
$$=\ The\ sum\ of\ the\ mean\ activity\ of\ all\ muscles$$

For directional co-contraction, the closer the value is to zero, the greater the co-contraction; conversely, the closer the value is to 1 or -1, the lesser the co-contraction.[39]

Statistical Analysis

The normal distribution of the data was confirmed using the Shapiro–Wilk test. To identify baseline differences between groups, a one-way ANOVA was conducted. If significant baseline differences were found for a particular variable, an analysis of covariance (ANCOVA), using the pre-test score as a covariate, was planned to assess intervention effects on that variable. A two-way repeated measures ANOVA was then used to assess the effects of the interventions across the four groups (three intervention groups: natural grass, artificial turf, and synthetic surface; and one control group) over two time points (pre- and post-intervention).

When significant group-by-time interactions were found, or significant main effects were observed in the absence of an interaction, group-specific post-hoc tests with Bonferroni adjustments were performed. Additionally, effect sizes were calculated by converting partial eta-squared ($\eta^2 p$) from the ANOVA output to Cohen's d values. According to Cohen's guidelines (40), a d value of less than 0.50 indicates a small effect, 0.50 to 0.79 indicates a medium effect, and 0.80 or higher indicates a large effect. Statistical significance was set at $p < 0.05$. All statistical analyses were carried out using the Statistical Package for Social Sciences (SPSS) version 26.0.

Results

Baseline Characteristics

Baseline demographic and clinical characteristics of the participants across all four groups are summarized in Table 1. No statistically significant differences were found among the groups in terms of age, height, or body mass ($p > 0.05$), confirming the comparability of participants at baseline. Additionally, key clinical variables, including navicular drop, foot posture index (FPI), and rearfoot eversion angle, were also statistically similar between groups. These findings suggest that the randomization process was successful in establishing equivalent groups prior to the intervention phase.

Pre-Test Comparisons

Statistically significant differences were observed for general co-contraction at the push-off phase ($p = 0.032$) among the four groups during pre-tests. Furthermore, statistically significant differences were found for directed co-contraction between the vastus lateralis (VL) and vastus medialis (VM) muscles at the heel-contact phase ($p = 0.048$). Other variables did not show statistically significant differences during pre-test ($p > 0.05$) (Table 2).

General Co-Contraction

There were no statistically significant main effects of group, main effects of time, or group-by-time interactions for general co-contraction at any phase (HC, MS, PO) among the four groups ($p > 0.05$) (Table 3).

Directed Flexor-Extensor Co-Contraction

A marginally significant main effect of time was observed for directed flexor-extensor co-contraction in the heel-contact phase ($p = 0.050$). Additionally, statistically significant group-by-time interactions were found for directed flexor-extensor co-contraction at the mid-stance ($p = 0.035$) and push-off phases ($p = 0.020$) (Table 4).

Post-Hoc Analysis

To explore the significant group-by-time interactions, within-group changes from pre-test to post-test were assessed for the directed flexor-extensor co-contraction variables using paired-samples t-tests. For the mid-stance phase, the change was statistically significant only for the natural grass group ($p = 0.017$, Cohen's $d = -0.32$). The changes for the control group ($p = 0.457$, Cohen's $d = -0.73$), synthetic surface group ($p = 0.291$, Cohen's $d = 1.19$), and artificial turf group ($p = 0.183$, Cohen's $d = -0.04$) were not statistically significant.

For the push-off phase, none of the within-group changes reached statistical significance: control group ($p = 0.085$, Cohen's $d = 0.84$), synthetic surface group ($p = 0.271$, Cohen's $d = -0.93$), natural grass group ($p = 0.653$, Cohen's $d = -0.41$), and artificial turf group ($p = 0.484$, Cohen's $d = 0.76$).

Analysis of Covariance (ANCOVA)

After adjusting for pre-test values, ANCOVA did not demonstrate statistically significant differences for general co-contraction at the push-off phase ($p > 0.05$) or for directed co-contraction between the VL and VM muscles in the heel-contact phase ($p > 0.05$) among the four groups (Table 5).

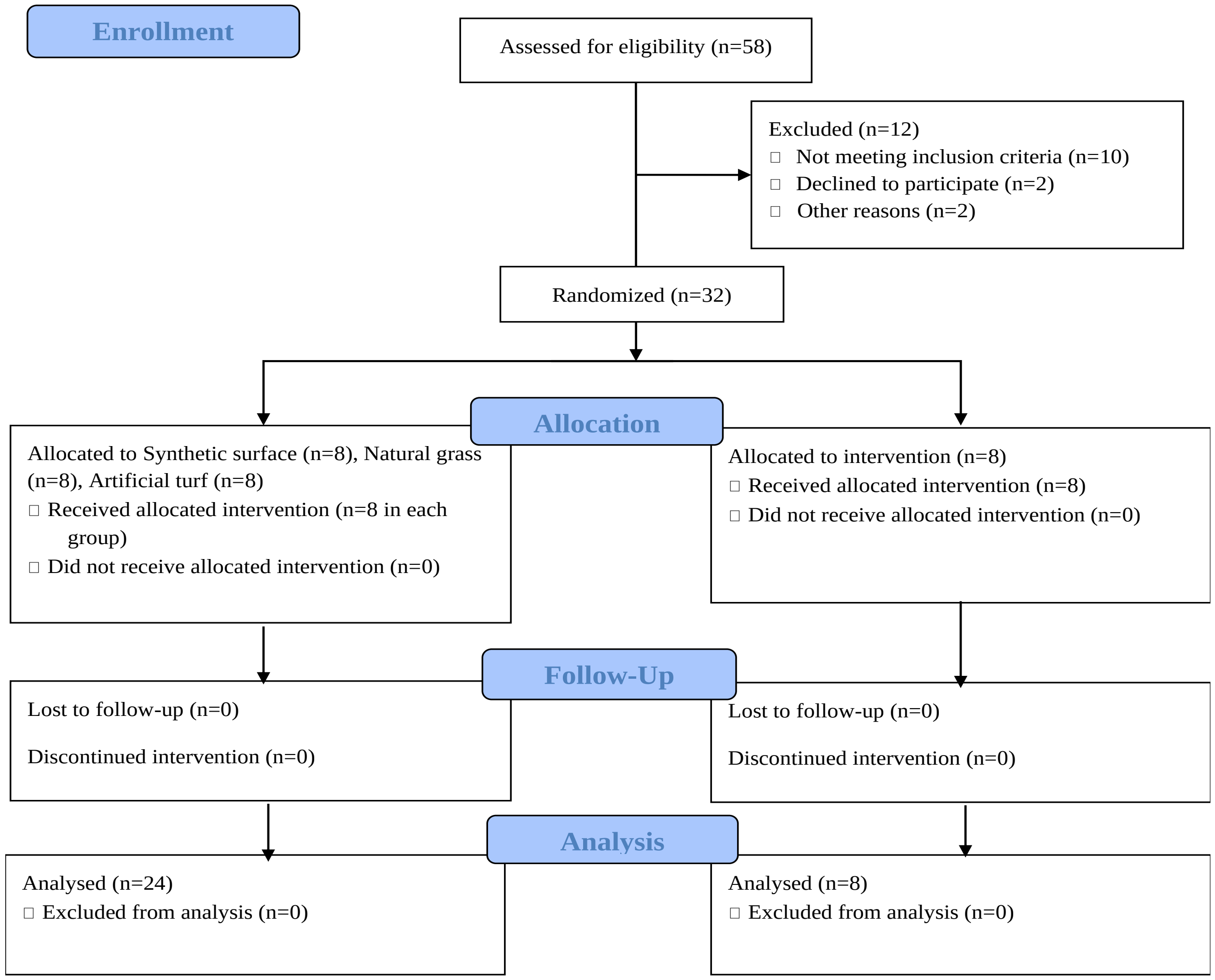

Figure 1 Consort flow diagram.

Table 1. Demographic characteristics of all four groups (mean ± SD).

Variable	Control	Synthetic surface	Natural grass	Artificial turf	Sig. (p-value)
Age (years)	24.82 ± 1.13	24.08 ± 1.39	23.5 ± 0.84	23.98 ± 1.22	0.174
Height (m)	1.76 ± 0.03	1.77 ± 0.02	1.78 ± 0.03	1.78 ± 0.03	0.389
Body mass (kg)	73.45 ± 4.0	77.35 ± 2.95	76.66 ± 4.31	77.12 ± 3.92	0.169
Navicular drop (mm)	11.44 ± 0.65	11.31 ± 0.97	11.6 ± 0.9	11.38 ± 0.44	0.891
Foot posture index	10.72 ± 0.49	11.06 ± 0.4	10.86 ± 0.5	11.2 ± 0.62	0.273
Rearfoot eversion (°)	6.2 ± 0.47	6.18 ± 0.46	6.19 ± 0.61	6.22 ± 0.48	0.998

Table 2. Comparison of the general and directed co-contraction between the four groups (pre-tests)

Muscle (sub-phase)	Control	Synthetic surface	Natural grass	Artificial turf	Sig.
General (HC)	243.25±178.29	142.85±69.14	221.05±154.19	218.76±79.99	0.435
General (MS)	352.60±170.06	218.74±122.10	319.33±174.12	359.28±148.34	0.257
General (PO)	540.83±188.82	317.16±117.93	412.48±194.18	515.39±80.39	0.032*
Directed flexor-extensor (HC)	0.33±0.44	-0.08±0.58	0.05±0.70	0.45±0.20	0.241
Directed flexor-extensor (MS)	-0.97±1.05	-0.03±0.65	-0.80±1.31	-1.20±1.06	0.166
Directed flexor-extensor (PO)	0.25±0.28	-0.81±1.52	-1.00±0.95	0.31±0.39	0.104
Directed mediolateral (HC)	-0.00±0.46	-0.51±0.66	-0.12±1.10	-0.26±0.39	0.522
Directed mediolateral (MS)	-0.11±0.52	0.26±0.30	-0.06±0.37	0.27±0.37	0.133
Directed mediolateral (PO)	0.07±0.38	0.16±0.36	-0.03±0.40	0.20±0.19	0.580
Directed VL-VM (HC)	-0.09±0.62	-2.13±2.59	-0.13±0.99	-1.37±1.99	0.048*
Directed VL-VM (MS)	-0.07±0.58	0.36±0.39	-0.04±0.41	0.27±0.57	0.195
Directed VL-VM (PO)	0.18±0.38	0.28±0.38	-0.25±0.85	0.11±0.50	0.230

Notes: HC: Heel Contact; MS: Mid-Stance; PO: Push-Off; VL: Vastus Lateralis; VM: Vastus Medialis; Sig.: Significance level (p-value). Data are presented as Mean ± Standard Deviation. *Differences were considered statistically significant at p<0.05.

Table 3. General co-contraction values in the four groups

Surface / Condition	Variable	Pre-test	Post-test	Main effect of Group (Eta square)	Main effect of Time	Group*Time interaction
Control	General (HC)	243.25±178.29	148.99±71.77	0.224 (0.142)	0.499 (0.017)	0.053 (0.237)
Control	General (MS)	352.60±170.06	289.29±117.24	0.968 (0.009)	0.550 (0.013)	0.326 (0.114)
Control	General (PO)	199.88±134.69	183.82±83.68	0.764 (0.040)	0.461 (0.020)	0.622 (0.060)
Synthetic surface	General (HC)	142.85±69.14	130.77±59.62	0.224 (0.142)	0.499 (0.017)	0.053 (0.237)
Synthetic surface	General (MS)	218.74±122.10	291.56±120.99	0.968 (0.009)	0.550 (0.013)	0.326 (0.114)
Synthetic surface	General (PO)	160.52±54.69	168.49±57.65	0.764 (0.040)	0.461 (0.020)	0.622 (0.060)
Natural grass	General (HC)	221.05±154.19	187.36±99.21	0.224 (0.142)	0.499 (0.017)	0.053 (0.237)
Natural grass	General (MS)	319.33±174.12	331.75±232.93	0.968 (0.009)	0.550 (0.013)	0.326 (0.114)
Natural grass	General (PO)	155.98±55.17	163.60±54.49	0.764 (0.040)	0.461 (0.020)	0.622 (0.060)
Artificial turf	General (HC)	218.76±79.99	303.60±193.64	0.224 (0.142)	0.499 (0.017)	0.053 (0.237)
Artificial turf	General (MS)	359.28±148.34	407.41±191.31	0.968 (0.009)	0.550 (0.013)	0.326 (0.114)
Artificial turf	General (PO)	147.98±44.08	193.37±109.98	0.764 (0.040)	0.461 (0.020)	0.622 (0.060)

Table 4. Directed flexor-extensor and mediolateral co-contractions in the four groups

Surface / Condition	Variable	Pre-test	Post-test	Main effect of Group (Eta square)	Main effect of Time	Group*Time interaction
Control	Directed flexor-extensor (HC)	0.33±0.44	-0.06±0.44	0.275 (0.127)	*0.050 (0.130)	0.955 (0.011)
Control	Directed flexor-extensor (MS)	-0.97±1.05	-0.56±0.07	0.714 (0.047)	0.565 (0.012)	0.035* (0.260)
Control	Directed flexor-extensor (PO)	0.25±0.28	-0.28±0.99	0.803 (0.034)	0.580 (0.011)	0.020* (0.291)
Control	Directed mediolateral (HC)	-0.00±0.46	-0.00±0.43	0.149 (0.171)	0.431 (0.022)	0.967 (0.009)
Control	Directed mediolateral (MS)	-0.11±0.52	-0.08±0.49	0.631 (0.059)	0.171 (0.066)	0.617 (0.061)
Control	Directed mediolateral (PO)	0.07±0.38	-0.01±0.52	0.586 (0.066)	0.942 (0.000)	0.860 (0.026)
Control	Directed VL-VM (MS)	-0.07±0.58	-0.28±0.86	0.569 (0.068)	0.120 (0.084)	0.916 (0.018)
Control	Directed VL-VM (PO)	0.18±0.38	0.01±0.49	0.426 (0.093)	0.724 (0.005)	0.507 (0.079)
Synthetic surface	Directed flexor-extensor (HC)	-0.08±0.58	-0.39±1.42	0.275 (0.127)	*0.050 (0.130)	0.955 (0.011)
Synthetic surface	Directed flexor-extensor (MS)	-0.03±0.65	-1.40±1.65	0.714 (0.047)	0.565 (0.012)	0.035* (0.260)
Synthetic surface	Directed flexor-extensor (PO)	-0.81±1.52	0.16±0.56	0.803 (0.034)	0.580 (0.011)	0.020* (0.291)
Synthetic surface	Directed mediolateral (HC)	-0.51±0.66	-0.32±0.64	0.149 (0.171)	0.431 (0.022)	0.967 (0.009)
Synthetic surface	Directed mediolateral (MS)	0.26±0.30	-0.07±0.74	0.631 (0.059)	0.171 (0.066)	0.617 (0.061)
Synthetic surface	Directed mediolateral (PO)	0.16±0.36	0.15±0.46	0.586 (0.066)	0.942 (0.000)	0.860 (0.026)
Synthetic surface	Directed VL-VM (MS)	0.36±0.39	-0.20±1.08	0.569 (0.068)	0.120 (0.084)	0.916 (0.018)
Synthetic surface	Directed VL-VM (PO)	0.28±0.38	0.11±0.79	0.426 (0.093)	0.724 (0.005)	0.507 (0.079)
Natural grass	Directed flexor-extensor (HC)	0.05±0.70	-0.42±0.86	0.275 (0.127)	*0.050 (0.130)	0.955 (0.011)
Natural grass	Directed flexor-extensor (MS)	-0.80±1.31	-0.48±0.71	0.714 (0.047)	0.565 (0.012)	0.035* (0.260)
Natural grass	Directed flexor-extensor (PO)	-1.00±0.95	-0.47±1.67	0.803 (0.034)	0.580 (0.011)	0.020* (0.291)
Natural grass	Directed mediolateral (HC)	-0.12±1.10	0.06±0.42	0.149 (0.171)	0.431 (0.022)	0.967 (0.009)
Natural grass	Directed mediolateral (MS)	-0.06±0.37	-0.11±0.58	0.631 (0.059)	0.171 (0.066)	0.617 (0.061)
Natural grass	Directed mediolateral (PO)	-0.03±0.40	0.09±0.44	0.586 (0.066)	0.942 (0.000)	0.860 (0.026)
Natural grass	Directed VL-VM (MS)	-0.04±0.41	-0.38±1.41	0.569 (0.068)	0.120 (0.084)	0.916 (0.018)
Natural grass	Directed VL-VM (PO)	-0.25±0.85	0.06±0.72	0.426 (0.093)	0.724 (0.005)	0.507 (0.079)
Artificial turf	Directed flexor-extensor (HC)	0.45±0.20	0.24±0.64	0.275 (0.127)	*0.050 (0.130)	0.955 (0.011)
Artificial turf	Directed flexor-extensor (MS)	-1.20±1.06	-1.15±1.29	0.714 (0.047)	0.565 (0.012)	0.035* (0.260)
Artificial turf	Directed flexor-extensor (PO)	0.31±0.39	-0.19±0.93	0.803 (0.034)	0.580 (0.011)	0.020* (0.291)
Artificial turf	Directed mediolateral (HC)	-0.26±0.39	-0.06±0.50	0.149 (0.171)	0.431 (0.022)	0.967 (0.009)
Artificial turf	Directed mediolateral (MS)	0.27±0.37	-0.13±0.74	0.631 (0.059)	0.171 (0.066)	0.617 (0.061)
Artificial turf	Directed mediolateral (PO)	0.20±0.19	0.20±0.22	0.586 (0.066)	0.942 (0.000)	0.860 (0.026)
Artificial turf	Directed VL-VM (MS)	0.27±0.57	0.04±0.53	0.569 (0.068)	0.120 (0.084)	0.916 (0.018)
Artificial turf	Directed VL-VM (PO)	0.11±0.50	0.36±0.23	0.426 (0.093)	0.724 (0.005)	0.507 (0.079)

Notes: HC: Heel Contact; MS: Mid-Stance; PO: Push-Off; VL: Vastus Lateralis; VM: Vastus Medialis. Data are presented as Mean ± Standard Deviation. Eta square values (effect size) are presented in parentheses. * Indicates a statistically significant difference ($p<0.05$).

Table 5. Co-variate co-contractions between the four groups

Muscle (sub-phase)	Control	Synthetic surface	Natural grass	Artificial turf	Sig.
General (PO)	413.26±137.64	404.62±118.51	370.84±193.86	382.57±99.91	0.085 (0.214)
Directed VL-VM (HC)	-0.31±1.03	-0.49±1.06	-0.43±1.17	-0.39±0.56	0.991 (0.004)

Notes: HC: Heel Contact; PO: Push-Off; VL: Vastus Lateralis; VM: Vastus Medialis; Sig.: Significance level. Data are presented as Mean ± Standard Deviation. Values in the 'Sig.' column represent p-value followed by Eta square (effect size) in parentheses. Differences were considered statistically significant at $p<0.05$.

Discussion

The present study aimed to investigate the long-term effects of training on different surfaces on knee muscular co-contraction in individuals with over-pronated feet (OPF). Our findings indicate that while an eight-week training program did not significantly alter general knee co-contraction, it appeared to induce surface-specific modulations in directed flexor-extensor co-contraction, as evidenced by the significant group-by-time interactions during the mid-stance ($p = 0.035$) and push-off ($p = 0.020$) phases of running. This distinction between the stability of general co-contraction and the adaptability of directed co-contraction suggests a sophisticated re-tuning of motor control strategies, rather than a simple increase in joint stiffening, in response to new mechanical environments.

The absence of significant changes in general knee co-contraction is a noteworthy finding. It is plausible that this reflects a homeostatic mechanism, whereby the neuromuscular system attempts to maintain a consistent level of overall joint stiffness during a given locomotor task. This concept is consistent with the work of Fu and colleagues, who reported that runners appear to make biomechanical adjustments to maintain similar tibial impact forces across surfaces of varying hardness. The stability of general co-contraction in our study could represent a neuromuscular component of such a compensatory strategy. In contrast, the significant alterations in directed co-contraction may reveal the underlying mechanism of this "fine-tuning." This suggests that the central nervous system strategically modulates the balance between agonist and antagonist muscle groups to meet specific mechanical demands, a principle of remote compensation that has been effectively modeled in simulation studies. For instance, the work by Wang and Gutierrez-Farewik demonstrated that distal perturbations at the ankle necessitate coordinated compensatory actions from proximal musculature at the knee and hip to preserve gait integrity.[35, 40]

The strategic use of specific co-contraction patterns to enhance joint stability is a well-documented neuromuscular response to mechanical perturbation. For example, in studies of unstable footwear, Horsak and colleagues[41] reported increased co-contraction around the knee as a stabilizing response. In a clinical context, a more targeted adaptation was observed by Heiden et al.[39] in patients with knee osteoarthritis, who employed a distinct, laterally directed co-contraction strategy, presumably to dynamically stabilize the joint against an external adduction moment. Our findings in an OPF population, which is characterized by altered lower limb mechanics, appear to be consistent with this paradigm of functional adaptation. This altered mechanical profile in OPF is well theorized; as Powers (2003) described,[42] excessive foot pronation can initiate a cascade up the kinetic chain, leading to compensatory femoral internal rotation and subsequent alterations in patellofemoral joint mechanics. Our findings of altered knee co-contraction can therefore be interpreted as a direct neuromuscular response to this perturbed mechanical environment. The surface-specific nature of these directed strategies suggests that the neuromuscular system may be highly attuned to the unique mechanical properties of the training environment, including factors such as surface compliance and friction. This is further highlighted by related findings from Jafarnezhadgero et al. (2024),[43] who observed that running on artificial turf specifically increased vastus medialis activity, a key stabilizer of the patellofemoral joint.

It is important to consider that such neuromuscular adaptations likely have metabolic consequences. As demonstrated by Sassi and colleagues,[44] the energetic cost of running was found to be significantly higher on grass and artificial turf compared to a hard surface, suggesting a potential trade-off between enhanced stability and metabolic efficiency. However, it is important to consider that this adaptive response may be specific to long-term training rather than acute stressors. In a study on acute fatigue, for instance, Fasihi et al. (2021) found that individuals with pronated feet demonstrated lower general knee co-contraction than healthy controls.[7] This contrasts with our findings and suggests that the nature of the perturbation (chronic training vs. acute fatigue) may dictate the neuromuscular strategy.

A more advanced interpretation of our findings can be framed within the context of inter-joint stiffness regulation. Recent work by Apps and colleagues using instability footwear has shown that the locomotor system can dynamically reorganize joint stiffness, reporting a decrease in ankle stiffness that occurred concurrently with an increase in knee stiffness. This suggests a compensatory strategy whereby the burden of stabilization is shifted proximally. This concept of kinetic shifting is further corroborated by the work of Willwacher et al. (2022),[25] who found that running barefoot on harder surfaces tended to increase external ankle moments while simultaneously decreasing knee moments, highlighting the dynamic interplay between these joints in managing ground reaction forces. It is plausible that our OPF participants employ a similar strategy, whereby the mechanical challenge at the foot leads to a neuromuscular response that prioritizes stability at the knee. However, the relationship between muscle activation and mechanical stiffness is complex. As the work by Apps et al. also highlighted, increased ankle co-contraction can be associated with reduced ankle quasi-stiffness, underscoring that these two metrics of stability are not always directly proportional.

From a clinical and practical standpoint, these data suggest caution in the prescription of training surfaces for individuals with OPF. The notion that soft surfaces are universally protective is challenged, particularly when our findings are contextualized with companion studies from the same research protocol. First, Jafarnezhadgero et al. (2024) found that this same eight-week training program induced no significant changes in ankle joint co-contraction, which strongly suggests that the primary site of neuromuscular adaptation was indeed the knee.[43] More critically, a second companion study by Jafarnezhadgero et al. (2024) demonstrated that long-term training on compliant surfaces such as natural grass and artificial turf resulted in significantly higher vertical loading rates upon returning to a hard surface.[45] This suggests that a mixed-surface training approach might be a more effective strategy for developing a robust and versatile neuromuscular system.

Limitations

Notwithstanding these findings, several limitations of this study must be acknowledged. First, the use of barefoot testing, while effective for isolating the foot-surface interaction, may limit the direct applicability of the results to the more common shod running condition. Second, the absence of concurrent kinematic or kinetic data precludes a complete analysis of the underlying movement strategies. Third, our cohort consisted of recreationally active individuals, and the findings may not be generalizable to elite athletes. Finally, as highlighted in the work of García-Pérez et al. [46](47), significant biomechanical differences exist between overground and treadmill running; therefore, our findings should be interpreted within the context of overground locomotion.

Future Directions

This work opens several avenues for future inquiry. A critical next step will be to conduct longitudinal studies to ascertain whether these neuromuscular adaptations have long-term clinical consequences, such as influencing injury rates or contributing to the progression of joint degeneration over time. It would also be highly beneficial to integrate EMG analysis with 3D motion capture and kinetic measurements to create a comprehensive biomechanical profile of these adaptations. Such an approach would allow for a more direct examination of the relationship between muscle activation strategies, joint moments, and mechanical stiffness. Finally, extending these investigations to competitive athletic populations and assessing neuromuscular function directly on the training surfaces themselves will be essential for a deeper understanding of the intricate interplay between foot biomechanics, the training environment, and injury prevention.

Conclusion

This study demonstrates that long-term surface-specific training in runners with OPF prompts a sophisticated re-tuning of knee muscular control, rather than a global change in joint stabilization. While the overall magnitude of knee co-contraction remained unchanged following the eight-week program, we observed significant, surface-dependent adaptations in the directed co-contraction between flexor and extensor muscle groups. This adaptive plasticity, however, may not be wholly protective. Our findings suggest a potential disconnect between neuromuscular adaptation and mechanical risk mitigation, particularly when considering evidence that such training can increase subsequent loading rates during surface transitions. Consequently, the selection of training surfaces for individuals with biomechanical abnormalities such as OPF warrants a more nuanced approach than simply favoring compliant surfaces. An emphasis on integrated conditioning that prepares the neuromuscular system for varied mechanical demands appears to be a more prudent strategy for injury prevention.

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Conflict of Interest Disclosures

Authors declare no conflict of interests.

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Authors' Contributions

Conceptualization: AJ, FI, SE, HS; Methodology: AJ, FI, SE, HS; Validation: FI, SE, HS; Formal analysis: FI; Investigation: AJ, FI; Data curation: AJ, FI; Supervision: AJ; Project administration: AJ; Writing original draft: AJ, FI, SE, HS; Writing review & editing: AJ, FI, SE, HS. All authors have read and approved the final version of the manuscript and agree with the order of authorship.

Ethical Statement

The study protocol was reviewed and approved by the Local Ethics Committee of Ardabil Medical Sciences University, Ardabil, Iran (Ethics Approval No.: IR.SSRC.REC.1400.08). The trial was prospectively registered in the Iranian Registry of Clinical Trials (IRCT) under the identifier IRCT20170806035517N5. All procedures complied with the latest version of the Declaration of Helsinki and relevant institutional/national guidelines. Written informed consent was obtained from all participants prior to enrollment. The ethics approval code and IRCT registration ID are reported in the Materials and Methods section.

Declaration of Generative AI and AI-assisted technologies

The authors declare that no generative AI or AI-assisted technologies were used in the writing or analysis of this manuscript.

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