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Effect of Eight Weeks of Training on Artificial Grass, Natural Grass, and Synthetic Surface on Ankle Joint Co-Contraction during Running in Individuals with Over-Pronation

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Abstract

Running is one of the most popular physical activities in the world and is usually done on different surfaces. Different levels of running are associated with overuse injuries. Therefore, this study aimed to evaluate the effect of eight weeks of training on artificial grass, natural grass, and synthetic surface on ankle joint co-contraction during running in individuals with over-pronation. This study was designed as a double-blinded randomized controlled trial. Sixty participants aged 18–30 years with diagnosed excessive pronation of foot were randomly allocated into three intervention groups (natural grass, artificial grass, and synthetic surface) and a control group. Electromyography data during pre and posttest was collected using surface electromyography system. Results did not demonstrated and statistically significant between group differences in in directed and general ankle joint co-contraction (P>0.05). The results of the present study showed that the ankle joint co-contraction during training on three types of artificial grass, natural grass, and synthetic surfaces was not statistically different in individuals with over-pronation.

Key Words: Over-pronation, Co-contraction, Electromyography, Running

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INTRODUCTION

Running is one of the most popular physical activities in the world and is usually done on different surfaces. Different levels of running are associated with overuse injuries. However, all the evidence supporting this statement is inconclusive [1, 2]. Research has shown that the most commonly used road surfaces for distance runners are synthetic rubber, concrete, and asphalt [3, 4]. On the other hand, recreational runners tend to prefer artificial grass surfaces [5, 6]. It is important to note that each road surface may have varying levels of stiffness and elasticity which can impact the biomechanics of running [1]. The characteristics of different surfaces cause biomechanical changes in gait [3] and make the running surface an essential aspect in the design of gait analysis methodology [7].

A previous study has indicated that when the surface stiffness increases, adaptive changes occur in the lower limb kinematics [8]. It has also been observed that runners who strike with their rearfoot exhibit more pronation feet and a more plantarflexed foot when running on a harder surface such as concrete or asphalt, as compared to a softer surface like grass or synthetic rubber [1, 9, 10]. The popularity of artificial grass surfaces has increased in recent years, with an estimated 6,000 installations in North America and 1000 to 1,500 new installations annually [11]. These surfaces consist of various components that can be manipulated independently to achieve desired mechanical properties [11, 12]. These components include an underlayment or shock pad, a synthetic fiber carpet, and infill material made up of sand, rubber, and possibly other organic materials [11, 13].

Athletic performance, including jumping and running economy, has been shown to be altered by changes in surface stiffness, with performance improvements as high as 12% reported [14, 15]. Short-duration activities like sprinting and vertical jumping and long-duration activities like running can be influenced by changes in stiffness [11]. However, it should be noted that many of the studies utilized extreme changes in stiffness or controlled non-sport-specific movements [11]. Varying stiffness in in filled artificial grass has been shown to affect peak vertical accelerations during running, as well as athlete contact times and step lengths [16, 17]. The underlying mechanism of how changes in surface stiffness affect athlete performance and biomechanics remains unknown, making it challenging for sports surface manufacturers to optimize surface stiffness [11].

Proper sporting surfaces are among the most important pieces of equipment [18, 19]. Various factors, including shock absorbance, friction, and energy loss are considered for the selection of playing surfaces [19]. Among these factors, shock absorbance is considered a key factor in preventing injuries [19]. Potential mechanisms for different patterns of injuries on artificial turf compared to natural grass include torque, rotational stiffness, interaction of surface and shoes, and shock absorption [19].

People who have experienced lower-leg injuries due to exercise were found to have increased pronation and eversion, higher plantar pressure on the medial side of the foot, and increased eversion velocity with lateral roll-off [20]. This suggests that changes in running biomechanics, such as plantar pressure distribution, can cause injury [20]. Identifying these changes can aid in preventing injuries [3]. Over-pronation foot is one of the most important causes of musculoskeletal injuries in the lower limb, which leads to an increase in mechanical loads on the lower limb structure [21]. Over-pronation of the foot can lead to various injuries in the lower back and lower limbs [21]. People with excessive foot pronation suffer from many injuries including Achilles tendonitis, leg pain due to muscle strain, and patellar femoral pain [21]. Also, over-pronation of the foot causes disturbance in posture control, disturbance in the pressure on the plantar, changes in the excitability of the ankle joint, and changes in the activity of the ankle joint muscles [22]. Farahpour et al reported that over-pronation of the foot increases the pressure and load on the joints of the lower limb from the ground surface during gait [23, 24]. In research, they showed that during walking, the leg muscles of people with over-pronation feet are more active than the normal structure of the foot [25]. Also, an increase in the activity of the evertor muscles has been reported in people with over-pronation of the foot [21].

Feehery et.al conducted a study comparing running on asphalt, concrete, and natural grass [26]. They discovered that running on concrete led to a shorter time to reach the first vertical force peak compared to grass and asphalt [26]. However, there was a higher first vertical force peak on grass. The researcher suggested that injury may occur in individuals running on hard surfaces due to the rapid transmission of shock waves through the body. This can limit the body's ability to dampen high-frequency shock waves as speed increases [3].

Many running coaches advise their athletes to use natural grass surfaces due to the lower risk of musculoskeletal injuries [27]. However, it is still unclear how the musculoskeletal system adapts to repetitive cyclical loads during running and how different surfaces affect these adaptations [8]. Research in this area has yet to yield definitive conclusions on the relationship between surface type, load on the locomotor apparatus, and injuries [3].

To identify and compare muscle coordination, movement patterns, and muscle activity level, surface electromyography can be used [28]. Electromyography activity is influenced by the forces acting on the foot, which are considered sensory inputs that affect muscle tone [28]. Lower limb muscle activity while running on a treadmill was studied by Wang et al. in comparison to other surfaces such as cement, natural grass, and synthetic surface [29]. Their findings showed significant changes in lower limb muscle activity during running on different surfaces, which was attributed to the body's kinematic adaptability to running surfaces [28].

Co-contraction is a phenomenon where agonist and antagonist muscles (antagonistic pairs) are activated simultaneously during activities such as postural control, walking, and running [30-32]. In normal gait, the antagonistic muscles contract alternatively with low durations of concurrent activity to produce enough joint moments [33, 34]. Falconer and Winter observed the highest ankle plantar- and dorsiflexor co-contraction during weight-acceptance and the lowest during push-off and swing phases, which suggests the stabilizing function of muscle co-contraction [32, 34].

Some neuromuscular pathologies and high locomotive energy costs can cause inefficient or abnormal movement due to increased or decreased co-contraction [34]. Electromyography (EMG) is frequently used to measure muscle activity and quantify co-contraction using indices based on the overlap area of EMG activity of antagonistic muscle pairs, but the number of muscles that can be feasibly recorded is limited, and compensation strategies to overcome excessive co-contraction cannot be assessed [34]. Pinnington et.al studied the kinematic and electromyography aspects of running on a firm surface and on soft, dry sand to elucidate mechanisms contributing to the higher energy cost of sand running [35]. They found the increased energy cost of running on sand can be attributed in part to the increased EMG activation associated with greater hip and knee range of motion compared with firm surface running [35].

Co-contraction can distribute internal forces more evenly and may contribute to injury prevention [36]. For the plantar-flexors and dorsi-flexors, separately, it has been demonstrated that an increase in muscle activation leads to increased mechanical joint stiffness, which results in increased joint stability. Therefore, the co-contractions of plantar-flexors and dorsi-flexors may increase ankle mechanical joint stiffness [37-40].

The researchers did not find a study that determined the control mechanisms of feet and ankle joint movement through the co-contraction of the agonist and antagonist muscles during training on artificial grass, natural grass, and synthetic surfaces in individuals with over-pronation feet, so this study aimed to compare the effect of eight weeks of training on artificial grass, natural grass, and synthetic surfaces on ankle joint co-contraction during running in individuals with over-pronation.

METHODOLOGY

This study was designed as a double-blinded randomized controlled trial. An envelope concealment method was used to allocate study participants. Participants and examiners were unaware of group allocation. In other words, participants and assessors were blinded. A power analysis (G*Power [41]) determined that 60 participants were required to achieve a statistical power of 0.80 at an effect size of 0.80 with an alpha level of 0.05 in GRF variables.

Sixty participants aged 18–30 years with diagnosed excessive pronation of foot were recruited from the students of the University of Mohaghegh Ardabili in Ardabil City, Iran, and were randomly allocated into three intervention groups (natural grass, artificial grass, and synthetic surface) and a control group. A kicking ball test was used to determine the dominant limp of the participants [42]. Inclusion criteria to participate in this study were: he/she had to be aged between 18-30, show a navicular drop of >10 mm [43], rear foot eversion of >4° [44], and a Foot Posture Index of >10 [43]. The amount of navicular drop was examined during non-weight bearing in comparison to static standing [7, 45]. The exclusion criteria, for this research were: (i) history of trunk and/or lower limbs surgery, (ii) orthopedic conditions (except for PF), having a history of fracture, and (iii) greater than 5 mm limb length differences. Before starting the study, the study procedures were described to all participants. Thereafter, written informed consent was obtained from the participants. The study protocol was approved by the local ethics committee of the Ardabil Medical Sciences University (IR.SSRC.REC.1400.08) and registered by the Iranian clinical trial organization (IRCT20170806035517N5). The study was conducted in agreement with the latest version of the Declaration of Helsinki.

Training on natural grass surfaces, artificial grass surfaces, and synthetic surfaces was applied to the subjects of the intervention group for eight weeks (three sessions per week), which included slow running, long strides, bounding, and short sprints. Each session started with warming up for 5 minutes and ended with cooling down for 5 minutes. The duration of each training session was 50 minutes. For the subjects of the control group, no exercise and sports activities were applied during eight weeks. Tibialis anterior and gastrocnemius medialis muscle activity was evaluated by an 8-channel electromyography system with a surface electrode during running. The raw EMG signals were digitized at 2000 Hz and streamed via Bluetooth to a computer for further analysis. According to the European recommendations for surface electromyography (SENIAM), the skin surface was shaved and cleaned with alcohol (70% Ethanol–C2H5 OH) over the selected muscles. After that, the skin was scratched gently prior to electrode placement. Electromyography (EMG) data were synchronized using Nexus software (Oxford Metrics, Oxford, UK).

Using an 8-channel electromyography device (Biosystem, UK) and a surface electrode, tibialis anterior and gastrocnemius medialis muscle activity was investigated. In order to record the surface electromyography waves, first the hairs of the desired surfaces were shaved and the skin was prepared for electrode placement with cotton and alcohol. The distance from the center to the center of the electrodes was 20 mm. The electrical signals were recorded with a frequency of 1000 Hz, and a bandwidth of 500 Hz, and then the existing noises were removed with 500 Hz low-pass and 10 Hz high-pass filters and a 50 Hz notch filter. Since the activity of lower limb muscles is related to the structure of the foot and lower limb damage, the electrical activity of the anterior tibial muscles was recorded at a sampling frequency of 1000 Hz to calculate the co-contraction of the ankle joint [46]. Electrode placement was done with the SENIAM method [46]. After the completion of the electrode placement process, the subject was asked to run a few steps in the laboratory environment, and in this way, possible limitations through the electrodes that could be created for the subject were identified and removed. Then the subjects of the two groups performed three barefoot running tests naturally during the pre-test and post-test, and the average of these three trials was used for data analysis [47].

To analyze the data obtained from electromyography, Biometrics DataLITE software and a 10-450 Hz low-pass filter were used. To normalize the electromyography signals, the RMS (Root Mean Square) information of each muscle was divided by the maximum isometric voluntary contraction (MVIC) value of that muscle and then multiplied by one hundred. For this purpose, the maximum electrical activity of each muscle was recorded in a period of 1 second and it was used as a baseline for comparisons. Muscle activity in each stage was expressed as a percentage of the baseline. According to the quality of the signals obtained from the foot switches, the third stride signal after running was studied. The following relationships were used to determine the values of both directional co-contraction and general co-contraction in different phases of running [47].

If agonist mean EMG > antagonist mean EMG;

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

Else

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

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

In directional co-contraction, the closer the number is to zero, the greater the co-contraction, and the closer the number is to 1 and -1, the less co-contraction [47].

The normal distribution of data was confirmed through the Shapiro–Wilk test. One-way ANOVA was applied to detect between-group differences at baseline. four groups (three interventions: Natural grass, Artificial turf and synthetic surface, and control) by two-time (pre, post) ANOVA for repeated measures was used to evaluate potential intervention effects. In the case of statistically significant group-by-time interactions, group-specific and Bonferroni-adjusted post-hoc tests were applied. Moreover, effect sizes were calculated by converting partial eta-squared (η2p) from ANOVA output to Cohen’s d. In accordance with Cohen [48], $d < 0.50$ demonstrate small effects, $0.50 \leq d < 0.80$ demonstrate medium effects and $d \geq 0.80$ demonstrate large effects. Statistical significance was set at $p < 0.05$. The Statistical Package for Social Sciences (SPSS) version 20.0 was used for all statistical analyses.

RESULTS

Based on the results obtained, there were no statistically significant differences in one of the variables (P>0.05) (Table 1).

Table1. Comparison of electrical activity of the muscles of between the three groups

Muscle (sub phase)	Control	Synthetic surface	Natural grass	Artificial grass	Sig.
General HC	73.50 ± 62.75	64.62 ± 32.00	64.01 ± 39.72	40.68 ± 12.43	0.608
General MS	110.84 ± 63.84	107.23 ± 36.97	86.08 ± 37.75	74.39 ± 24.20	0.492
General PO	199.88 ± 134.69	160.52 ± 54.69	155.98 ± 55.17	147.98 ± 44.08	0.620
Directed plantar-dorsal flexor HC	0.77 ± 0.15	0.81 ± 0.09	0.73 ± 0.17	0.72 ± 0.07	0.632
Directed plantar-dorsal flexor MS	0.49 ± 0.33	0.23 ± 0.40	0.46 ± 1.46	0.37 ± 1.24	0.596
Directed plantar-dorsal flexor PO	0.23 ± 0.69	0.41 ± 0.40	0.58 ± 0.21	0.68 ± 0.13	0.565

Based on the results obtained, there were no statistically significant differences in one of the variables (P>0.05) (Table 2).

Table 2. General co-contraction contrast between the three groups

Variable	Control		Synthetic surface		Natural grass		Artificial grass		Main effect of Group (Eta square)	Main effect of Time	Group by Time interaction
Co-contraction	Pre-test	Post-test	Pre-test	Post-test	Pre-test	Post-test	Pre-test	Post-test			
General HC	73.50 ± 62.75	55.92 ± 21.56	64.62 ± 32.00	58.07 ± 21.41	64.01 ± 39.72	59.00 ± 29.00	40.68 ± 12.43	71.44 ± 30.21	0.960 (0.010)	0.958 (0.000)	0.250 (0.134)
General MS	110.84 ± 63.84	96.48 ± 41.59	107.23 ± 36.97	94.51 ± 27.01	86.08 ± 37.75	102.74 ± 44.44	74.39 ± 24.20	115.48 ± 44.16	0.968 (0.009)	0.317 (0.036)	0.196 (0.152)
General PO	199.88 ± 134.69	183.82 ± 83.68	160.52 ± 54.69	168.49 ± 57.65	155.98 ± 55.17	163.60 ± 54.49	147.98 ± 44.08	193.37 ± 109.98	0.764 (0.040)	0.461 (0.020)	0.622 (0.060)

Based on the results obtained, there were no statistically significant differences in one of the variables (P>0.05) (Table 3).

Table 3. Directed plantar-dorsal flexor co-contraction between the three groups

Variable	Control		Synthetic surface		Natural grass		Artificial grass		Main effect of Group (Eta square)	Main effect of Time	Group by Time interaction
Co-contraction	Pre-test	Post-test	Pre-test	Post-test	Pre-test	Post-test	Pre-test	Post-test			
Directed plantar-dorsal flexor HC	0.77 ± 0.15	0.80 ± 0.20	0.81 ± 0.09	0.69 ± 0.24	0.73 ± 0.17	0.39 ± 0.50	0.72 ± 0.07	0.73 ± 0.2	0.205 (0.148)	0.176 (0.064)	0.297 (0.121)
Directed plantar-dorsal flexor MS	0.49 ± 0.33	0.26 ± 0.79	0.23 ± 0.40	0.36 ± 0.31	0.46 ± 1.46	0.27 ± 0.71	0.37 ± 1.24	0.09 ± 0.45	0.644 (0.057)	0.556 (0.013)	0.540 (0.073)

Directed plantar-dorsal flexor PO	0.23 ± 0.69	0.44 ± 0.40	0.41 ± 0.40	0.14 ± 0.73	0.58 ± 0.21	0.64 ± 0.20	0.68 ± 0.13	0.32 ± 0.52	0.914 (0.018)	0.746 (0.004)	0.473 (0.084)

DISCUSSION

The aim of the present study was to compare the effect of eight weeks of training on artificial grass, natural grass, and synthetic surface on ankle joint co-contraction during running in individuals with over-pronation. The findings showed that there was no statistically significant difference in the general co-contraction of the ankle joint in individuals with over-pronation between the three surfaces of artificial grass, natural grass, and synthetic surface in running practice. Powell et al. [49] reported that individuals with over-pronation feet have instability and high mobility in the push-off phase. On the other hand, increasing muscle co-contraction creates stability in the joint [50]. As a result, to maintain more stability in the feet and prevent extra movements of the ankle joint, the shank muscles of people with over-pronation foot need a greater amount of co-contraction, which is not consistent with the results of the present study. Antagonist muscles work simultaneously with agonist muscles, in the meantime, co-contraction with higher levels of constant net torque is created [51, 52]. Zeni et.al (2009) have recently shown that the general co-contraction increases even when walking speed is increased in control and experimental groups with different severities of knee osteoarthritis [53]. They also showed in a similar article [54] that basically all the kinetic and kinematic differences between the control and experimental groups can be statistically calculated with different walking speeds of the subject. Based on this, the lack of significant difference between the groups of the present study can probably be due to the same speed of the subjects.

The results of the present study showed that there was no statistically significant difference in the directional co-contraction of the ankle joint in individuals with over-pronation between the three surfaces of artificial grass, natural grass, and synthetic surface in running training. The closer the directional co-contraction is to zero, the greater the co-contraction rate, and the closer it is to 1 and -1, the less the co-contraction rate [47]. The numerical value of the directional co-contraction of the ankle joint was changed on the surfaces of artificial grass, natural grass, and synthetic surfaces, but these changes did not show any statistical difference. It has been reported that an increase in the directional co-contraction of the ankle joint can be beneficial for the instability and mobility of the ankle joint of individuals with over-pronation feet [52]. It has also been shown that running and walking on a soft surface requires much more effort than on a hard surface, as a result of which muscle activity increases, and this strengthens the muscles, endurance, and stability [55, 56].

Conclusion

The results of the present study showed that the co-contraction of the muscles of the lower limbs during training on three types of artificial grass, natural grass, and hall surfaces was not statistically different during running in individuals with over-pronation.

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مقایسه تاثیر هشت هفته تمرین بر روی چمن مصنوعی، چمن طبیعی و سطح مصنوعی بر هم انقباضی مفصل مچ پا در حین دویدن در افراد دارای پرونیشن بیش از حد پا

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چکیده:

دویدن یکی از پرطرفدارترین فعالیت های بدنی در جهان است. سطوح مختلف دویدن با آسیب‌های ناشی از استفاده بیش از حد همراه است. بنابراین، این مطالعه با هدف بررسی مقایسه تاثیر هشت هفته تمرین بر روی چمن مصنوعی، چمن طبیعی و سطح مصنوعی بر هم‌انقباضی مفصل مچ پا در حین دویدن در افراد دارای پرونیشن بیش از حد پا انجام شد. این مطالعه به صورت کارآزمایی بالینی تصادفی دوسوکور طراحی شد. ۶۰ شرکت کننده ۱۸ تا ۳۰ ساله با تشخیص پرونیشن بیش از حد پا از بین دانشجویان دانشگاه محقق اردبیلی شهر اردبیل انتخاب و به طور تصادفی در سه گروه مداخله (چمن طبیعی، چمن مصنوعی، و سطح مصنوعی) و یک گروه کنترل گروه‌بندی شدند. بر اساس نتایج به‌دست‌آمده تفاوت آماری معنی‌دار بین گروهی در مقادیر هم‌انقباضی مچ پا وجود نداشت ($P>۰/۰۵$). نتایج مطالعه حاضر نشان داد که انقباض همزمان عضلات اندام تحتانی در حین تمرین بر روی سه نوع چمن مصنوعی، چمن طبیعی و سطوح سالن در حین دویدن در افراد دارای پرونیشن بیش از حد تفاوت آماری نداشت.

کلمات کلیدی: پرونیشن بیش از حد، هم انقباضی، الکترومایوگرافی، دویدن