*Journal of Advanced Sport Technology 3(1):8 – 21* Received: Jan. 13, 2019; Accepted: Apr. 23, 2019

**Original Research**

# Efficacy of motion control shoes for reducing the frequency response of ground reaction forces in fatigued runners

**AmirAli Jafarnezhadgero[1]*, Elham Sorkhe[2] & Abbas Meamarbashi [3]**

*1. Department of Physical Education and Sport Sciences, Faculty of Educational Sciences and Psychology, University of Mohaghegh Ardabili, Ardebil, Iran*

*2. Department of Physical Education and Sport Sciences, Faculty of Educational Sciences and Psychology, University of Mohaghegh Ardabili, Ardebil, Iran*

*3. Department of Sport Sciences, Faculty of Sport Sciences, Ferdowsi University of Mashhad, Mashhad, Iran*

**ABSTRACT**

Fatigue along with pronated feet during running are possible risk factors for injuries. Motion control shoes were designed to limit excessive foot motion in runners, but its clinical efficacy has not been well reported. This study investigated the ground reaction force characteristics in novice female runners with pronated feet during running with different footwear before and after fatigue of the lower limb muscles. 22 novice female runners with pronated feet were volunteered to participate in this study. A Kistler force plate and a Vicon three-dimensional motion analysis system were used to record the ground reaction forces and gait pattern of participants, when running with motion control shoes and control shoes before and after fatigue of the lower limb muscles.
Results: The results of this study showed that the frequency with a power of 99.5% of in the vertical ground reaction force with the motion control shoes were lower than that control shoes during post-test. the frequency with a power of 99.5% of in the anterior-posterior Center of pressure ground reaction force with the motion control shoes were greater than that motion control shoes during pre-test. the frequency with a power of 99.5% of in the Medio-lateral Center of pressure ground reaction force with the motion control shoes were greater than that control shoes during pre-test. the frequency with a power of 99.5% of in the free moment ground reaction force with the control shoes were greater than that control shoes during post-test.

**Keywords:** Fatigue, Motion control shoes, Ground reaction force, Frequency response of ground reaction forces

## Introduction

During running, repetitive loading at footstrike generates intermittent impulse forces which are propagated through the musculosketetal system [1]. These impulse forces are gradually attenuated as they travel toward the head by the body's natural shock absorbers such as the heel pad, subchondral bone, articular cartilage, intervertebral discs and menisci [2]. Some researchers have suggested the repetitive nature and magnitude of these impulse forces are associated with musculoskeletal injury [3-5].The effect of physical fatigue on running style also remains elusive. Fatigue due to training and extended bouts of running has been theorized as a potential mechanism of injury [6-8]. Researchers have looked into identifying biomechanical differences between measurements taken before and after a fatiguing running bout, yielding conflicting results for vertical loading rate [6, 9], A shorter stride length [7, 10] and higher stride frequency [10, 11] for example. fatigued runner may alter the biomechanics of their running by decreasing the angle of the foot with the running surface at initial contact [6], changing the plantar pressure distribution [12, 13], and modifying reaction forces [10, 14].

One biomechanical risk factor thought to predispose a runner to this injury is ‘‘excessive’’ pronation [15, 16]. Foot pronation, defined as a combined movement of calcaneal eversion, forefoot abduction and dorsiflexion, is the natural adaptive movement in the subtalar and midtarsal joints of the foot for shock absorption during walking and running (15, 16). Among the foot abnormalities, excessive foot pronation has received much clinical attention due to its high incidence [17, 18].result in abnormal lower limb kinematics [19, 20], ground reaction forces (GRF) [21], muscle activities [22, 23], and free moments [24, 25] during gait 12 leading to LBP [26]. Footwear is essential equipment for runners and there are different shoe technologies developed to prevent running injuries [27]. One mainstream shoe technology is the ‘motion control’ design, which targets to reduce excessive rearfoot movement during running, and thus, the incidence of injuries. The mechanism of motion control shoes was based on different deformation rates between lateral and medial midsoles [28, 29] and wedging heel counter [30, 31] so as to control the relative rates of mid-foot and rearfoot motion. Shoe type has been thought to affect the running style, and a handful of studies comparing motion control and conventional running shoes have been published [27, 32, 33].

These biomechanical parameters are typically analyzed through the identification of local discrete points and their respective times of occurrence [34].Ground reaction force (GRF) frequency content may be representative of frequency domain characteristics of many anatomical components such as joints, muscles, and nerves during walking [35]. This approach is referred to as time domain analysis, where the data are examined as a function of time. However, time domain analysis examines discrete points rather than the global make-up of the data [34].Furthermore, it seems that the frequency content of GRFs during gait can provide an important role in clinical treatment [36]. For example, Giakas et al. (1996) examined the frequency domain of GRF in scoliosis patients. Their study was important as they also examined common time-dependent measures. They found significantly higher frequency domain content in all three planes with the largest effect in the medial–lateral direction. They found no significant differences in the mode pendent measures. This is valuable considering scoliosis is a tri-planar spinal deformity that seems to affect balance and walking in all three planes and yet common time-dependent measures could not detect any differences [37].

Wurdeman et al. (2011), in their study on multiple sclerosis patients, suggested that frequency domain analysis of GRF could potentially provide earlier insights into the progression of the disease [38]. Furthermore, the frequency corresponding to 99% or 99.5% of the GRF total power spectral density (PSD) was found to be lower in clinical populations than in healthy controls [34, 39]. Also, it is reported that preadolescents with down syndrome had a similar fundamental frequency as their healthy peers even though the down syndrome group walked at slower speeds [35]. In light of a lack of information on motion control shoe design for controlling pronation foot during muscle fatigue, the present study was conducted. The aims were to: (1) examine the (GRF) frequency content of the pronation foot during running with motion control shoes and control shoes before lower leg muscle fatigue, and (2) compare the pronation foot (GRF) frequency content after lower leg muscle fatigue in the two footwear testing conditions.

## Material and Methods

Twenty-two female recreational runners with foot pronation (age: 24.1±5.6 years; height: 165.6±10.4 cm; body mass: 63.21±12.05 kg) and no other no known neural, musculoskeletal injuries in the previous 6 months or cardiopulmonary problems participated voluntarily in this study. Inclusion criteria for the experiment were running training volume of 2-3 times/week for 45 min or 10 km each work out. A participant’s foot was considered pronated if there was a navicular drop >10 mm [40], maximum rearfoot angle higher than 6○ during running with neutral shoe compared with offset angle measured in standing [41, 42], and a foot posture index of greater than 10 [42]. Exclusion criteria were the support from running clubs and training for any competitive races in the period of the experiment. Prior to participation, written informed consent was obtained from all participants. Approval for all procedures was obtained from the Research Ethics Committee of the Ardabil University of Medical Sciences (code number: IR.ARUMS.REC.1396.135).

participants were asked to run while wearing either motion control shoes designed for over-pronators (ASICS Women's GEL-Kayano 24 Running Shoe; Figure 1A) or regular neutral regular running shoes (ASICS Women's GEL-Nimbus 19 Running Shoe; Figure 1B). The order of the shoes across days was randomized. These shoe models were selected based on their availability in the local market and comparable design. According to the manufacturer, the major structural difference between the selected shoes was the composite materials in the midsole. After arriving in the laboratory, a specific set of retro-reflexive markers were placed on specific landmarks and segments of the participant (see session 2.4 for further details). The next step was an overground running protocol designed to record kinematic running parameters. Subsequently, the participant performed a fatiguing treadmill running protocol on a closely located laboratory. Finally, the same overground running protocol was conducted immediately after finishing the treadmill fatiguing protocol.


**Figure 1.** Control neutral shoes (A) and Motion control (B) used in this study.

Ground reaction forces was recorded using Force plates (Kistler AG, Winterthur, Switzerland) embedded in the middle of an 18-m walkway. These force plates were connected to a Vicon MX system (Vicon Motion Systems, Oxford, UK) that recorded data at a 1000 Hz sampling frequency. Participants were familiarized with the laboratory environment and walkway area and at least five practice trials were performed to ensure subjects were able to strike the force plate without consciously changing their running cadence. Afterwards, each participant identified with heel strike pattern during running (kinematic analysis) performed five acceptable running trials at running speed of 3.3 m/s. A trial was discarded if the dominant foot did not land on the force plates, if the participant targeted the platforms, lost balance during the trial, ran with mid or forefoot strike pattern, or even fell during running. A 10 cm visual analog scale was used to assess the level of comfort of the footwear. Participants were asked to mark their responses on a 10 cm visual analog scale after each testing condition.

The fatiguing protocol consisted of running on a motorized treadmill with no inclination (Horizon Fitness, Omega GT, USA), while heart rate was monitored continuously (Polar RS100, Polar Electro Oy, Woodbury, NY). Participants started the test at 6 km/h, and the treadmill speed was increased 1 km/h every 2 minutes. The perceived exertion was collected from participants at every speed increase through a 15-point Borg scale (6–20) [43]. When a participant reported a perceived exertion of 13 or higher, the treadmill speed was fixed to allow for a steady-state running period. If a participant reported perceived exertion equal or abo17, or if his/her heart rate was above 80% of maximum, the fatiguing protocol was terminated after two minutes from this point [44].

We performed a frequency analysis by first converting the GRFs ($F_x$, $F_y$, $F_z$), free moment and $COP_x$ , $COP_y$ of each trial to the frequency domain using fast. Force data were normalized by expressing the results as a percentage of the subject's body weight. The Matlab program used a fast Fourier transform to extract frequency content of the GRF data [45]. The detailed presentation of Fourier series of GRF can be found elsewhere [46, 47]. Based on previous studies, five frequency-domain indexes were used for further analysis for the vertical, anterior–posterior, and medio-lateral GRF of each trial [34, 38]. From the power spectrum curve, it is possible to calculate different dependent variables of interest. Based on previous studies [39, 48] and our pilot work, we chose the dependent variables to be investigated in this study as: The frequency component of the greatest magnitude contained in 99.5% of the PSD plot was chosen as the criterion to represent frequency content. This choice was made in order to identify subtle features of the GRFs patterns, such as the heel strike impulsive load in the vertical force, usually described by high-frequency harmonics.

The first index was frequency with power of 99.5% (F99.5%), which indicates the frequency that has 99.5% of the signal power or, in other words 99.5% of signal power is lower than that such frequency (Equation 1) [34].

$$\int_0^{f99.5} p(f)df = 0.995 \times \int_0^{f_{max}} p(f)df \tag{1}$$

The second index was median frequency ($F_{med}$) [34]. Median frequency is the point where half of the total power is above and below that frequency (Equation 2) [34, 49].

$$\int_0^{f_{med}} p(f)df = \int_{f_{mea}}^{f_{max}} p(f)\, df \qquad (2)$$

The third index is frequency bandwidth ($F_{band}$) The frequency band width ($F_{band}$) is a difference between maximum and minimum frequencies when the power is at greater than half the maximum power (Equation 3) [34, 39, 48].

$$f_{band} = f_{max} - f_{min}(when\ p > 1/2 \times p_{max}) \qquad (3)$$

According to the method described by Schneider and Chao (1983), the fourth index was the essential number of harmonics (ne) that required for 99.5% degree of data reconstruction.
This variable is defined as the number of harmonics satisfying the condition that the sum of the relative amplitudes of each harmonic over the total amplitude is less than or equal to
0.995 (Equation 4) [50].

$$\sum_{n=1}^{n_e} \frac{\sqrt{A_n^2 + B_n^2}}{\sum_{n=1}^{m} \sqrt{A_n^2 + B_n^2}} \leq 0.95 \qquad (4)$$

Where n is the harmonic number; $A_n$ and $B_n$ are Fourier coefficients. The fifth index was the amplitude of each harmonic (Hi), where i is the number of each harmonic. Normal distribution of data was confirmed by Shapiro-Wilk test. Repeated measures ANOVA (general linear model), with significance level at 0.025 (Bonferroni adjustment), was used to test the main factors of fatigue (Pre vs. Post) and shoe type (motion control vs. neutral). Paired t-test was used to detect difference between subjects' feedback on footwear comfort.

## Results

No significant differences in reported comfort were observed between control (5.4±2.7) and motion control shoes (5.8±2.9)(p>0.05).

The ANOVA results revealed that in frequency content of the GRFs in the three directions (vertical, anterior-posterior, medio-lateral, $Cop_x$, $Cop_y$, free moment). vertical frequency at 99.5% on the motion control shoes after muscle fatigue were lower than that control shoes after muscle fatigue protocol (by 20%; P=0.00; small effect size). Other frequency content of vertical did not show any significant differences between four conditions (P>0.05; Table 1). (Table 1). However, Amplitude of H4 (by 60%; P=0.04; small effect size), H5 (by 30%; P=0.00; small effect size) and H6 (by 45%; P=0.00; medium effect size) in vertical component on the motion control shoes after muscle fatigue were lower than that control shoes after muscle fatigue protocol (Figure 2).

**Table 1.** Mean and standard deviation of the frequency of vertical ($F_z$), ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Fz | Frequency with power of 99.5% | 9.42±1.02 | 9.727±1.23 | 8.931±1.62 | 8.07±0.87 | 0.140 (0.101) | 0.000* (0.559) | 0.063 (0.155) |
| | Medium frequency | 2.00±0.00 | 2.000±0.00 | 2.000±0.00 | 2.00±0.00 | 1.000 (0.000) | 1.000 (0.000) | 1.000 (0.000) |
| | Frequency bandwidth | 1.01±0.05 | 1.000±0.00 | 1.000±0.00 | 1.00±0.00 | 0.329 (0.045) | 0.329 (0.045) | 0.329 (0.045) |
| | essential number of harmonics | 26.16±6.04 | 28.31±5.62 | 28.75±4.86 | 27.62±6.87 | 0.722 (0.006) | 0.541 (0.018) | 0.124 (0.109) |

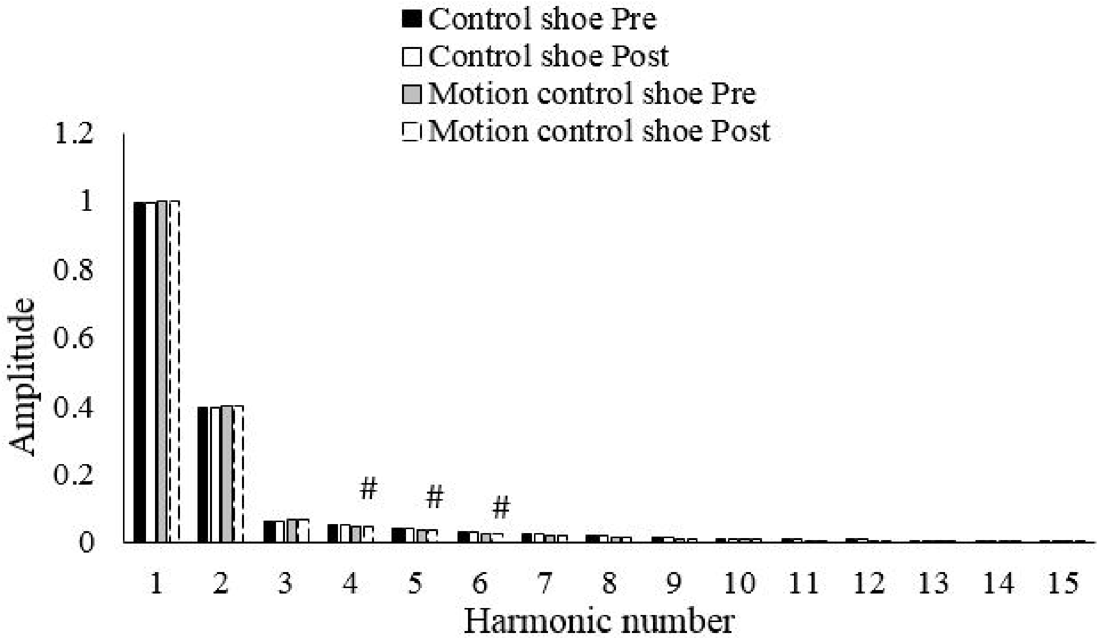


**Figure 2.** Amplitude versus harmonic number plot of the vertical ground reaction force.
# Significant difference between the control shoes and the motion control shoes testing groups during post-test.

In the anterior-posterior frequency at 99.5%, medium frequency, frequency bandwidth, essential number of harmonics did not show any significant differences between four conditions (P>0.05; Table 2).

Amplitude of H2 (by 80%; P=0.00; medium effect size), H3 (by 70%; P=0.01; medium effect size), H4 (by 43%; P=0.00; medium effect size) and H5 (by 40%; P=0.00; medium effect size) in anterior-posterior component on the motion control shoes after muscle fatigue were lower than that motion control shoes after muscle fatigue protocol. However, Amplitude of H2 (by 40%; P=0.00; medium effect size), H3 (by 65%; P=0.02; medium effect size), H4 (by 49%; P=0.00; high effect size) on the motion control shoes after muscle fatigue were greater than that control shoes after muscle fatigue protocol. Other Amplitude of H7 (by 95%; P=0.03; high effect size), H8 (by 97%; P=0.00; high effect size), H9 (by 87%; P=0.04; high effect size) and H11 (by 78%; P=0.03; high effect size), on the motion control shoes after muscle fatigue were greater than that motion control shoes after muscle fatigue protocol, and Amplitude of H7 (by 56%; P=0.03; high effect size), H8 (by 99%; P=0.00; high effect size), H10 (by 91%; P=0.02; high effect size) and H11(by 91%; P=0.03; high effect size) on the control shoes after muscle fatigue were greater than that motion control shoes after muscle fatigue protocol (Figure 3).

**Table 2.** Mean and standard deviation of the frequency of anterior-posterior ($F_x$), ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Fy | Frequency with power of 99.5% | 14.00±2.75 | 14.65±3.49 | 14.25±3.32 | 14.19±3.22 | 0.521 (0.020) | 0.857 (0.002) | 0.686 (0.008) |
| | Medium frequency | 2.62±4.11 | 1.78±0.33 | 1.68±0.26 | 1.75±0.28 | 0.388 (0.036) | 0.277 (0.056) | 0.341 (0.043) |
| | Frequency bandwidth | 1.03±0.11 | 1.00±0.00 | 1.00±0.00 | 1.02±0.10 | 0.825 (0.002) | 0.825 (0.002) | 0.110 (0.117) |
| | essential number of harmonics | 20.23±4.42 | 20.00±3.96 | 23.07±6.89 | 20.47±4.52 | 0.227 (0.069) | 0.070 (0.148) | 0.322 (0.047) |

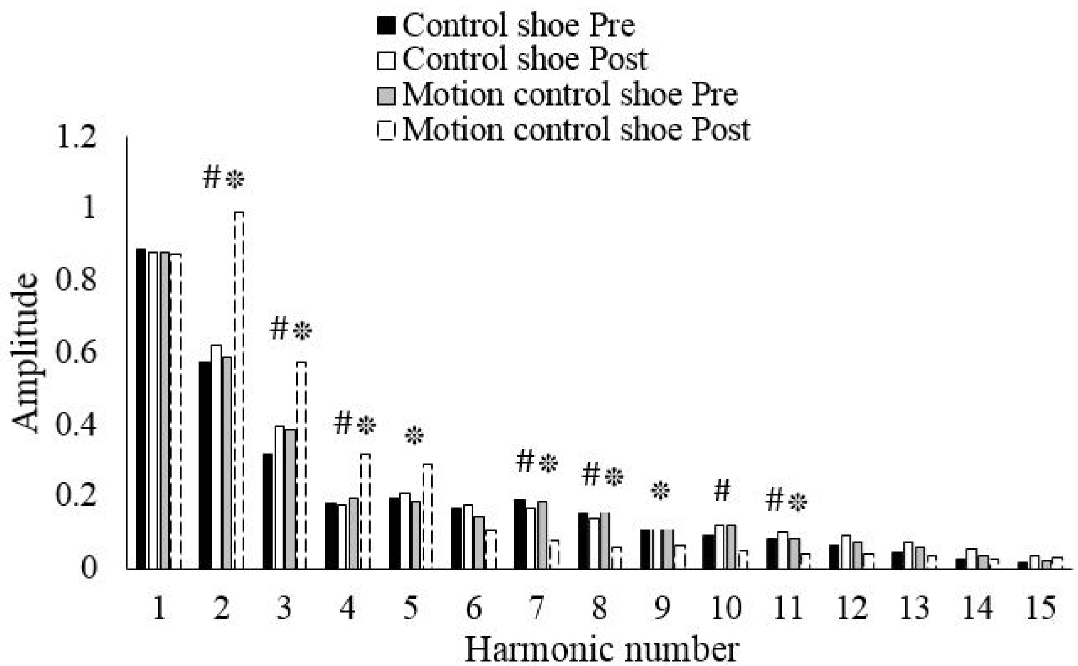


**Figure 3.** Amplitude versus harmonic number plot of the anterior-posterior ground reaction force. # Significant difference between the control shoes and the motion control shoes testing groups during post-test.* Significant difference between the pre and post-fatigue conditions of the motion control shoes testing condition.

In the Medio-lateral frequency at 99.5%, medium frequency, frequency bandwidth, essential number of harmonics of Medio-lateral did not show any significant differences between four conditions (P>0.05; Table 3).

In the Medio-lateral harmonic amplitudes did not show any significant differences between the four conditions (P>0.05; Figure 4).

**Table 3.** Mean and standard deviation of the frequency of Medio-lateral ($F_y$), ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Fx | Frequency with power of 99.5% | 13.63±2.19 | 16.04±4.43 | 12.29±2.17 | 14.02±1.70 | 0.001* | 0.013* | 0.499 |
| | | | | | | (0.433) | (0.261) | (0.022) |
| | Medium frequency | 2.95±1.50 | 3.46±2.47 | 3.55±2.54 | 2.23±0.29 | 0.055 | 0.340 | 0.050* |
| | | | | | | (0.165) | (0.043) | (0.170) |
| | Frequency bandwidth | 1.96±1.40 | 2.17±2.21 | 2.57±2.49 | 1.19±0.29 | 0.030* | 0.482 | 0.075 |
| | | | | | | (0.206) | (0.024) | (0.143) |
| | essential number of harmonics | 20.40±6.70 | 17.92±6.40 | 20.10±4.84 | 19.56±6.05 | 0.222 | 0.570 | 0.500 |
| | | | | | | (0.070) | (0.016) | (0.022) |

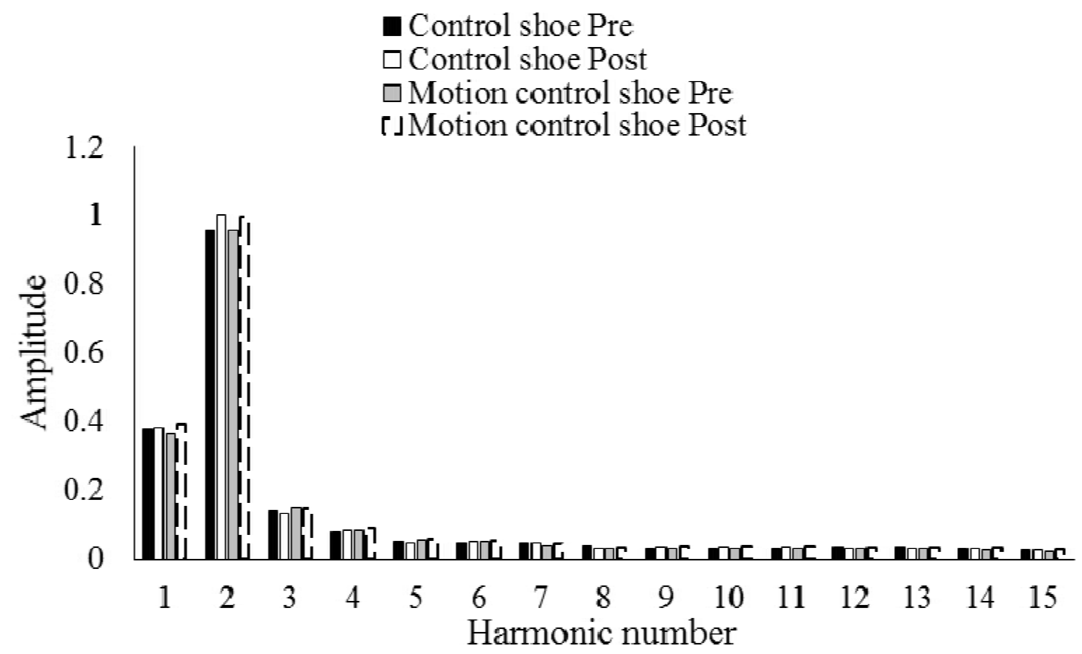


**Figure 4.** Amplitude versus harmonic number plot of the Medio-lateral ground reaction force.

In the $Cop_x$ frequency at 99.5% on the motion control shoes after muscle fatigue were greater than that motion control shoes after muscle fatigue protocol (by 97%; P=0.03; medium effect size).Other frequency content of Copx did not show any significant differences between four conditions (P>0.05; Table 4).

In the $Cop_x$ harmonic amplitudes did not show any significant differences between the four conditions (P>0.05; Figure 5).

**Table 4.** Mean and standard deviation of the frequency of the ($Cop_x$), ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Copx | Frequency with power of 99.5% | 23.95±8.57 | 24.37±7.77 | 18.25±5.66 | 18.27±13.40 | 0.900 | 0.003* | 0.906 |
| | | | | | | (0.001) | (0.357) | (0.001) |
| | Medium frequency | 2.18±0.44 | 2.09±0.25 | 2.00±0.15 | 2.03±0.08 | 0.637 | 0.036* | 0.342 |
| | | | | | | (0.011) | (0.192) | (0.043) |
| | Frequency bandwidth | 1.13±0.35 | 1.08±0.17 | 1.09±0.42 | 1.01±0.04 | 0.245 | 0.375 | 0.887 |
| | | | | | | (0.064) | (0.038) | (0.001) |
| | essential number of harmonics | 13.59±7.46 | 15.49±7.96 | 16.99±6.53 | 15.36±8.66 | 0.933 | 0.278 | 0.376 |
| | | | | | | (0.000) | (0.056) | (0.038) |

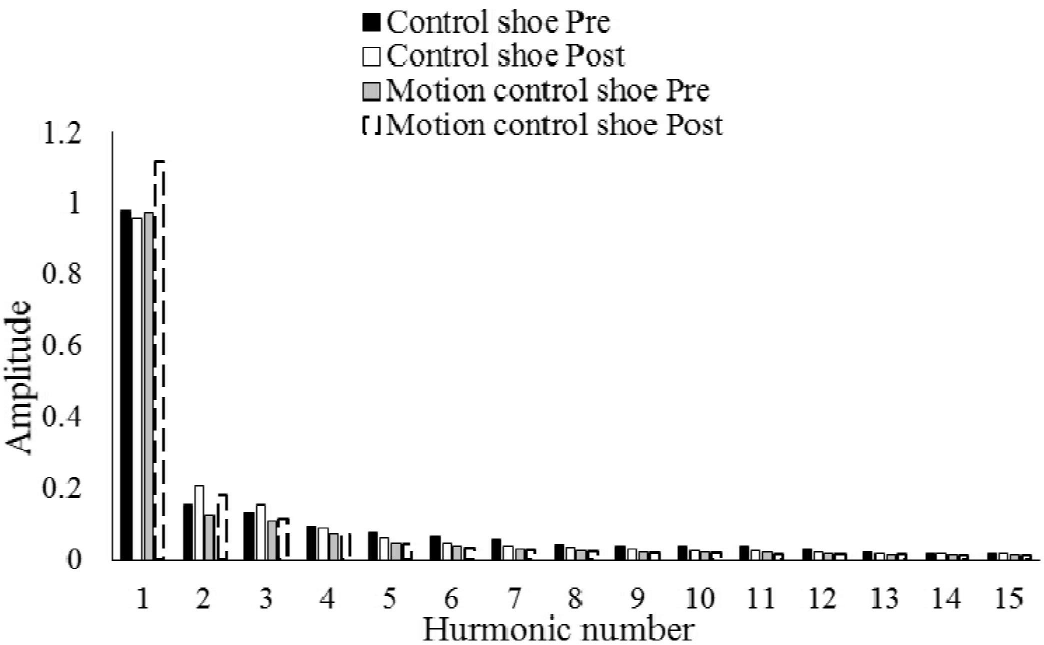


**Figure 5.** Amplitude versus harmonic number plot of the ($Cop_x$) ground reaction force.

In the ($Cop_y$) frequency at 99.5% on the motion control shoes before muscle fatigue were greater than that control shoes before muscle fatigue protocol (by 81%; P=0.04; small effect size). Other frequency content of Copy did not show any significant differences between four conditions (P>0.05; Table 5).

In the $Cop_y$ harmonic amplitudes did not show any significant differences between the four conditions ($P>0.05$; Figure 6).

**Table 5.** Mean and standard deviation of the frequency of the ($Cop_y$), ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Copy | Frequency with power of 99.5% | 25.13±3.43 | 24.71±4.47 | 28.75±13.57 | 24.68±2.47 | 0.202 (0.076) | 0.299 (0.051) | 0.208 (0.075) |
| | Medium frequency | 2.31±0.53 | 2.28±0.45 | 2.31±0.33 | 2.32±0.36 | 0.917 (0.001) | 0.820 (0.003) | 0.761 (0.005) |
| | Frequency bandwidth | 1.19±0.43 | 1.18±0.57 | 1.18±0.28 | 1.12±0.18 | 0.690 (0.008) | 0.735 (0.005) | 0.722 (0.006) |
| | essential number of harmonics | 16.51±2.33 | 16.16±1.88 | 17.04±2.03 | 16.62±1.72 | 0.387 (0.036) | 0.316 (0.048) | 0.911 (0.001) |

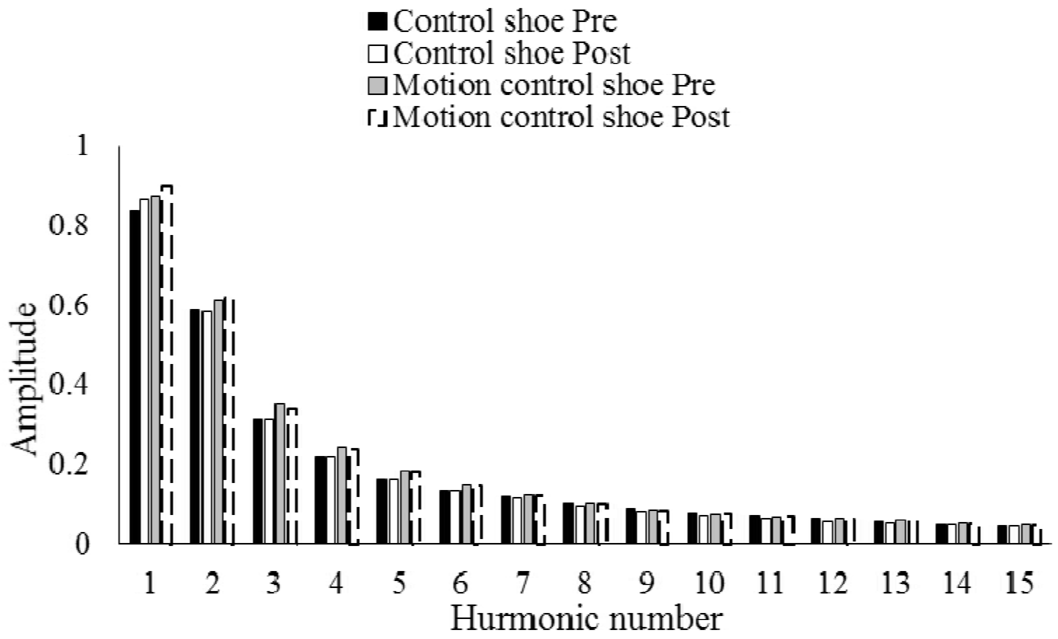


**Figure 6.** Amplitude versus harmonic number plot of the ($Cop_y$) ground reaction force.

In the Free moment frequency at 99.5% on the control shoes before muscle fatigue were greater than that control shoes before muscle fatigue protocol (by 56%; P=0.00; medium effect size). Other frequency content of free moment did not show any significant differences between four conditions ($P>0.05$; Table 6). However, Amplitude of H11 (by 57%; P=0.04; medium effect size) on the motion control shoes after muscle fatigue were greater than that motion control shoes before muscle fatigue protocol (Figure 7).

**Table 6.** Mean and standard deviation of the frequency of the free moments, ground reaction force during in different stance phases for four conditions are presented.

| Direction | Component | Control shoe | | Motion control shoe | | shoe | fatigue | Shoe* fatigue |
|---|---|---|---|---|---|---|---|---|
| | | Pre | Post | Pre | Post | | | |
| Free moment | Frequency with power of 99.5% | 19.71±3.32 | 22.46±0.18 | 15.01±3.64 | 16.09±3.50 | 0.117 | 0.000* | 0.533 |
| | | | | | | (0.113) | (0.475) | (0.019) |
| | Medium frequency | 1.82±0.24 | 2.12±1.17 | 1.77±0.27 | 1.78±0.28 | 0.253 | 0.210 | 0.291 |
| | | | | | | (0.062) | (0.074) | (0.053) |
| | Frequency bandwidth | 1.00±0.00 | 1.00±0.00 | 1.00±0.00 | 1.00±0.00 | 1.000 | 1.000 | 1.000 |
| | | | | | | (0.000) | (0.000) | (0.000) |
| | essential number of harmonics | 20.99±4.46 | 22.26±5.82 | 22.77±4.05 | 21.31±3.97 | 0.912 | 0.725 | 0.109 |
| | | | | | | (0.001) | (0.006) | (0.118) |

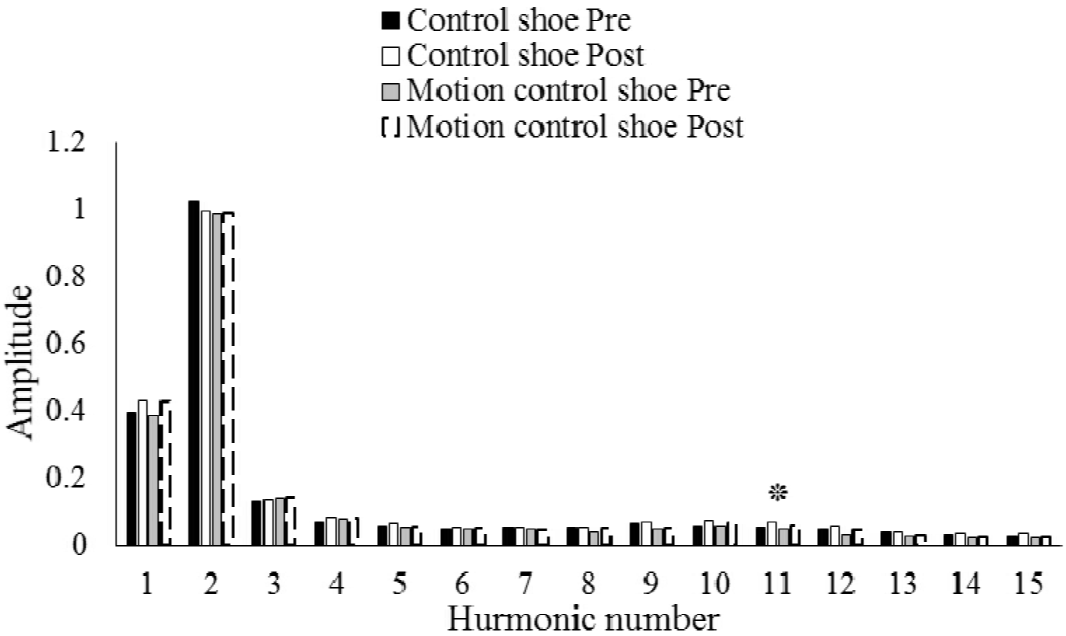


**Figure 7.** Amplitude versus harmonic number plot of the free moments ground reaction force. * Significant difference between the pre and post-fatigue conditions of the motion control shoes testing condition.

## Discussion

Frequency spectrum ground reaction force components are used to evaluate the gait pattern. Differences in neuronal and muscular systems interactions for running the subject can be found using available frequency analysis [34]. Frequency analysis can evaluate the differences in existing frequencies [39]. Studies have shown that the mean value of the frequency of the ground reaction force during the run at the peak is 12.81 ± 1.60 Hz and the active peak is about 04.04 ± 1.39 Hz [51]. However, no study has been done to investigate the effect of motion control shoes on the variables the frequency response of ground reaction force during running. Therefore, it is not possible to compare the results of this study with previous studies.

Our findings showed that the frequency values with the power of 99.5% of the Center of pressure of ground reaction force in the internal-external direction during shoe conditions were lower than the motion control shoe before the fatigue, The variables that affect the impact forces during running are: speed of running, contact surface, and area of the foot contact during the collision with the ground, the properties of the materials used during running such as shoe, supporters Be [52]. Stergiou in his research found that the amount the frequency response of ground reaction force in elderly the subject was significantly reduced compared to young the subject [39]. Researchers have reported that this ground reaction force component in walking and running analysis is an indicator of the balance and stability of the individual in the anterior-posterior plate [45]. Therefore, the use of control shoes in comparison with the motion control shoe may result in greater stability in the anterior-posterior plate.

Frequency values with a power of 99.5% of the Center of pressure of ground reaction force to anterior-posterior direction during the motion control shoes conditions, after fatigue was greater than the same shoes before fatigue. The frequency response of ground reaction force depends on all of the components of the partner in the motion system, including bones, muscles, nerves, and other tissues whose interactive effects make movement moves [34]. Therefore, according to the results of this study, it can be argued that the use of shoe motion control conditions may after fatigue, muscle activity has decreased, as well as the stiffness of the joints of the ankle and knee have decreased and, as a result of this decrease, increased shock absorption. Increasing shock absorption leads to a reduction in impact force and because the frequency response of ground reaction force is formed from the amplitudes of the different frequency components of the force-time signal, and the higher the power component of this signal (force -time), the higher the frequency [51].

Also, the frequency values with the power of 99.5% of the free moment of ground reaction force during the control shoes conditions, after fatigue was lower than the same shoes before fatigue. During the transitional movements, when the body enters the ground, The frequency response of ground reaction force is likely to exhibit some degree of performance of the oscillating component of the nervous-motion system [34, 38]. Muscles are one of the most important components of the nervous-motor system. One of the mechanisms to increase the risk of injury in the lower limb during running, which will increase the loading force increasing the hardness of the lower limbs, which reduces the absorption of shock [53, 54]. According to the research, it was concluded that the level of running and other foot supporters can relate to the factors related to the stiffness of the joints peak vertical ground reaction force and changes in the length of the step in the phase of stance with the ground affects [55, 56].

Frequency values with a power of 99.5% of the vertical ground reaction force direction during the motion control shoe conditions, after the fatigue is lower than the motion control shoes condition before fatigue. Reducing the frequency of the vertical component ground reaction force shows lower swings in the movements [57, 58]. Low oscillations can indicate the posture control sign more in a vertical direction. However, increasing the frequency will increase the instability and slide in the pattern of movements [58]. Wurdeman in his research concluded that the frequency of MS patients was significantly reduced compared with healthy subjects. This decrease could be due to the fact that MS patients to maintain balance, more control and therefore, require a further reduction in frequency [38]. Therefore, a significant decrease in the frequency values with a power of 99.5% in vertical direction may be due to the fact that the motion control shoes, by affecting the muscle strength, have led to a decrease in the performance of oscillating components of the nervous-motion system.

In the medium frequency of frequency content, there was no significant statistical difference in running with any of the shoes and in any of the direction. This finding shows that in the medium frequency domain, there is no difference in running with two types of shoe. In the McGrath study, the median frequency at the frequency response of ground reaction force is lower in subject with peripheral arterial disease than in healthy subject [34]. However, the use of different shoes did not change the frequency response of the ground reaction force components; therefore, the motion control shoes have no particular advantage compared to the control shoes in the frequency median.

Frequency content bandwidth was not found to be statistically significant in all four conditions and in all three dimensions. These findings indicate that the frequency domain of the two types of shoes has not changed, and this variable is independent of the material of the shoes. Frequency content bandwidth seems to

be more affected by living tissue. However, increased fatigued muscle is associated with a decrease in median frequency [59]. If the increase in bandwidth is related to the increase in motor units [59].
There were some limitations in our study that need to be considered when interpreting the findings.
1 Having a small sample size despite the normal distribution that may affect the results.
2. The lack of control of the stress caused by the subjects due to the condition of the test.
3. We only examined female novice runners with pronated feet, and so it is not possible for general male novice runners to have a disability.
4. We used novice runners to measure themselves, and it's unclear whether the professional runners have the same positive impact.

## Conclusion

Motion control shoes can control the frequency with a power of 99.5% of in the vertical ground reaction force in runners with pronated feet regardless of the state of leg muscle fatigue. However, a negative effect of motion control shoes has been an increase in the frequency with a power of 99.5% of in the anterior-posterior Center of pressure ground reaction force during running.

## Acknowledgement

We thank each of the subjects who participated in the study.

## References

1. Chu, M.L., et al., *An in vitro simulation study of impulsive force transmission along the lower skeletal extremity.* Journal of biomechanics, 1986. **19**(12): p. 979-987.
2. Wosk, J. and A. Voloshin, *Wave attenuation in skeletons of young healthy persons.* Journal of Biomechanics, 1981. **14**(4): p. 265-267.
3. Voloshin, A.A. and J. Wosk, *Shock absorbing capacity of the human knee (in vivo properties)*. 1980: Iowa State University College of Engineering.
4. Voloshin, A., J. Wosk, and M. Brull, *Force wave transmission through the human locomotor system.* Journal of Biomechanical Engineering, 1981. **103**(1): p. 48-50.
5. MacLellan, G. and B. Vyvyan, *Management of pain beneath the heel and Achilles tendonitis with visco-elastic heel inserts.* British journal of sports medicine, 1981. **15**(2): p. 117-121.
6. Christina, K.A., S.C. White, and L.A. Gilchrist, *Effect of localized muscle fatigue on vertical ground reaction forces and ankle joint motion during running.* Human movement science, 2001. **20**(3): p. 257-276.
7. Clansey, A.C., et al., *Effects of fatigue on running mechanics associated with tibial stress fracture risk.* Medicine and science in sports and exercise, 2012. **44**(10): p. 1917-1923.
8. Derrick, T.R., D. Dereu, and S.P. Mclean, *Impacts and kinematic adjustments during an exhaustive run.* Medicine & Science in Sports & Exercise, 2002. **34**(6): p. 998-1002.
9. Gerlach, K.E., et al., *Kinetic changes with fatigue and relationship to injury in female runners.* Medicine and science in sports and exercise, 2005. **37**(4): p. 657-663.
10. Willson, J.D. and T.W. Kernozek, *Plantar loading and cadence alterations with fatigue.* Medicine and science in sports and exercise, 1999. **31**(12): p. 1828-1833.
11. Morin, J., et al., *Changes in running mechanics and spring–mass behavior induced by a mountain ultra-marathon race.* Journal of biomechanics, 2011. **44**(6): p. 1104-1107.
12. Nagel, A., et al., *Long distance running increases plantar pressures beneath the metatarsal heads: a barefoot walking investigation of 200 marathon runners.* Gait & posture, 2008. **27**(1): p. 152-155.
13. Rosenbaum, D., T. Engl, and A. Nagel, *Foot loading changes after a fatiguing run.* Journal of Biomechanics, 2008. **41**: p. S109.
14. Weist, R., E. Eils, and D. Rosenbaum, *The influence of muscle fatigue on electromyogram and plantar pressure patterns as an explanation for the incidence of metatarsal stress fractures.* The American journal of sports medicine, 2004. **32**(8): p. 1893-1898.

15. Barton, C.J., et al., *Kinematic gait characteristics associated with patellofemoral pain syndrome: a systematic review.* Gait & posture, 2009. **30**(4): p. 405-416.
16. Boling, M.C., et al., *A prospective investigation of biomechanical risk factors for patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort.* The American journal of sports medicine, 2009. **37**(11): p. 2108-2116.
17. Castro-Méndez, A., P.V. Munuera, and M. Albornoz-Cabello, *The short-term effect of custom-made foot orthoses in subjects with excessive foot pronation and lower back pain: a randomized, double-blinded, clinical trial.* Prosthetics and orthotics international, 2013. **37**(5): p. 384-390.
18. Pinto, R.Z., et al., *Bilateral and unilateral increases in calcaneal eversion affect pelvic alignment in standing position.* Manual therapy, 2008. **13**(6): p. 513-519.
19. Müller, R., T. Ertelt, and R. Blickhan, *Low back pain affects trunk as well as lower limb movements during walking and running.* Journal of biomechanics, 2015. **48**(6): p. 1009-1014.
20. Kendall, J.C., A.R. Bird, and M.F. Azari, *Foot posture, leg length discrepancy and low back pain–Their relationship and clinical management using foot orthoses–An overview.* The Foot, 2014. **24**(2): p. 75-80.
21. Shum, G.L., J. Crosbie, and R.Y. Lee, *Three-dimensional kinetics of the lumbar spine and hips in low back pain patients during sit-to-stand and stand-to-sit.* Spine, 2007. **32**(7): p. E211-E219.
22. Hanada, E.Y., M. Johnson, and C. Hubley-Kozey, *A comparison of trunk muscle activation amplitudes during gait in older adults with and without chronic low back pain.* PM&R, 2011. **3**(10): p. 920-928.
23. van der Hulst, M., et al., *Lumbar and abdominal muscle activity during walking in subjects with chronic low back pain: support of the "guarding" hypothesis?* Journal of Electromyography and Kinesiology, 2010. **20**(1): p. 31-38.
24. Li, Y., et al., *Free vertical moments and transverse forces in human walking and their role in relation to arm-swing.* Journal of Experimental Biology, 2001. **204**(1): p. 47-58.
25. Bolgla, L.A., et al., *Hip strength and hip and knee kinematics during stair descent in females with and without patellofemoral pain syndrome.* journal of orthopaedic & sports physical therapy, 2008. **38**(1): p. 12-18.
26. Cambron, J.A., et al., *Shoe orthotics for the treatment of chronic low back pain: a randomized controlled pilot study.* Journal of manipulative and physiological therapeutics, 2011. **34**(4): p. 254-260.
27. Cheung, R.T. and G.Y. Ng, *Efficacy of motion control shoes for reducing excessive rearfoot motion in fatigued runners.* Physical Therapy in Sport, 2007. **8**(2): p. 75-81.
28. Clarke, T., E. Frederick, and C. Hamill, *The effects of shoe design parameters on rearfoot control in running.* Medicine and Science in Sports and Exercise, 1983. **15**(5): p. 376-381.
29. Nigg, B.M., et al., *The effect of material characteristics of shoe soles on muscle activation and energy aspects during running.* Journal of biomechanics, 2003. **36**(4): p. 569-575.
30. Stacoff, A., et al., *Tibiocalcaneal kinematics of barefoot versus shod running.* Journal of biomechanics, 2000. **33**(11): p. 1387-1395.
31. Stacoff, A., et al., *Effects of shoe sole construction on skeletal motion during running.* Medicine & Science in Sports & Exercise, 2001. **33**(2): p. 311-319.
32. Lilley, K., V. Stiles, and S. Dixon, *The influence of motion control shoes on the running gait of mature and young females.* Gait & posture, 2013. **37**(3): p. 331-335.
33. Malisoux, L., et al., *Injury risk in runners using standard or motion control shoes: a randomised controlled trial with participant and assessor blinding.* Br J Sports Med, 2016: p. bjsports-2015-095031.
34. McGrath, D., et al., *Peripheral arterial disease affects the frequency response of ground reaction forces during walking.* Clinical Biomechanics, 2012. **27**(10): p. 1058-1063.
35. Wu, J., M. Beerse, and T. Ajisafe, *Frequency domain analysis of ground reaction force in preadolescents with and without Down syndrome.* Research in developmental disabilities, 2014. **35**(6): p. 1244-1251.
36. Gruber, A.H., et al., *A comparison of the ground reaction force frequency content during rearfoot and non-rearfoot running patterns.* Gait & Posture, 2017. **56**: p. 54-59.
37. Giakas, G., et al., *Comparison of gait patterns between healthy and scoliotic patients using time and frequency domain analysis of ground reaction forces.* Spine, 1996. **21**(19): p. 2235-2242.
38. Wurdeman, S.R., et al., *Multiple sclerosis affects the frequency content in the vertical ground reaction forces during walking.* Clinical Biomechanics, 2011. **26**(2): p. 207-212.

39. Stergiou, N., et al., *Frequency domain characteristics of ground reaction forces during walking of young and elderly females.* Clinical Biomechanics, 2002. **17**(8): p. 615-617.
40. Lange, B., L. Chipchase, and A. Evans, *The effect of low-Dye taping on plantar pressures, during gait, in subjects with navicular drop exceeding 10 mm.* Journal of Orthopaedic & Sports Physical Therapy, 2004. **34**(4): p. 201-209.
41. Eng, J.J. and M.R. Pierrynowski, *The effect of soft foot orthotics on three-dimensional lower-limb kinematics during walking and running.* Physical Therapy, 1994. **74**(9): p. 836-844.
42. Hintermann, B. and B. Nigg, *Pronation in runners: Implications for injuries.* Occupational Health and Industrial Medicine, 1999. **1**(40): p. 47.
43. Borg, G., *Borg's perceived exertion and pain scales*. 1998: Human kinetics.
44. Koblbauer, I.F., et al., *Kinematic changes during running-induced fatigue and relations with core endurance in novice runners.* Journal of science and medicine in sport, 2014. **17**(4): p. 419-424.
45. Winter, D.A., *Biomechanics and motor control of human movement*. 2009: John Wiley & Sons.
46. Giakas, G. and V. Baltzopoulos, *Time and frequency domain analysis of ground reaction forces during walking: an investigation of variability and symmetry.* Gait & Posture, 1997. **5**(3): p. 189-197.
47. White, R., I. Agouris, and E. Fletcher, *Harmonic analysis of force platform data in normal and cerebral palsy gait.* Clinical Biomechanics, 2005. **20**(5): p. 508-516.
48. Stergiou, N., *Innovative analyses of human movement*. 2004: Human Kinetics Publishers.
49. Crowe, A., et al., *Characterization of gait of young adult females by means of body centre of mass oscillations derived from ground reaction forces.* Gait & Posture, 1993. **1**(1): p. 61-68.
50. Schneider, E. and E. Chao, *Fourier analysis of ground reaction forces in normals and patients with knee joint disease.* Journal of biomechanics, 1983. **16**(8): p. 591-601.
51. Hamill, J., T.R. Derrick, and K.G. Holt, *Shock attenuation and stride frequency during running.* Human movement science, 1995. **14**(1): p. 45-60.
52. Nigg, B.M., *Biomechanics of running shoes*. 1986: Champaign, Ill.: Human Kinetics Publishers.
53. Williams III, D.S., et al., *High-arched runners exhibit increased leg stiffness compared to low-arched runners.* Gait & posture, 2004. **19**(3): p. 263-269.
54. Williams III, D.S., et al., *Lower extremity kinematic and kinetic differences in runners with high and low arches.* Journal of applied biomechanics, 2001. **17**(2): p. 153-163.
55. Ferris, D.P., et al., *Neuromechanical adaptation to hopping with an elastic ankle-foot orthosis.* Journal of Applied Physiology, 2006. **100**(1): p. 163-170.
56. Ferris, D.P., K. Liang, and C.T. Farley, *Runners adjust leg stiffness for their first step on a new running surface.* Journal of biomechanics, 1999. **32**(8): p. 787-794.
57. White, L.J. and R.H. Dressendorfer, *Exercise and multiple sclerosis.* Sports medicine, 2004. **34**(15): p. 1077-1100.
58. Alshami, A.M., T. Souvlis, and M.W. Coppieters, *A review of plantar heel pain of neural origin: differential diagnosis and management.* Manual therapy, 2008. **13**(2): p. 103-111.
59. Judkins, T.N., et al., *Robotic surgery and training: electromyographic correlates of robotic laparoscopic training.* Surgical Endoscopy And Other Interventional Techniques, 2006. **20**(5): p. 824-829.

**Corresponding Author:** Dr. AmirAli Jafarnezhadgero, Department of Physical Education and Sport Sciences, Faculty of Educational Science and Psychology, University of Mohaghegh Ardabili, Ardabil, Iran. Email: Amiralijafarnezhad@gmail.com, Tel: +98 9105146214.