The Effect of the Level of Physical Activity on Electromyography of Core Stability Muscles, Ground Reaction Force, and Changes in Center of Mass to Pressure during Gait

Rasool Ferasat [a], **Heydar Sadeghi** [b,c*], **Hasan Matinhomaee** [d]

a Department of Exercise Biomechanics, Physical Education Faculty, Islamic Azad University, Central Tehran Branch, Tehran, Iran; **b** Department of Sport Biomechanics and Injuries, Faculty of Physical Education and Sport Sciences, Kharazmi University, Tehran, Iran; **c** Department of Sport Biomechanics, Kinesiology Research Center, Kharazmi University, Tehran, Iran; **d** Department of Exercise Physiology, Islamic Azad University, Central Tehran Branch, Tehran, Iran

***Corresponding Author:** Heydar Sadeghi, Faculty member, Kharazmi University, Tehran, Iran; E-mail: sadeghih@yahoo.com/sadeghih@khu.ac.ir

Submitted: 2021-03-12; **Accepted**: 2021-6-09; **Doi:** https://doi.org/10.22037/jcpr.v6i4.35983

Abstract

Introduction: Given that the difference in the neuromuscular functions and the movement patterns in athlete and non-athlete individuals are challenging, this study aimed to investigate a light on the effect of the level of physical activity on the Electromyography of core stability muscles and ground reaction force (GRF) and changes in the center of mass (COM) to center of pressure (COP) during gait. **Material and Methods**: Eighteen young men participated in this quasi-experimental study and were divided into groups including athletes and non-athletes. Mean and standard deviation used to describe data; Shapiro Wilk Test was used to verify the normality of the data and the T-test was used to compare the results of the two groups at $p \leq 0.05$. **Results**: The patterns of muscles functions were found to be similar in the two groups. However, there was a difference in the root mean square (RMS) of the two groups and this difference was significant in the Multifidus muscle. The amount of GRF was found to be higher at the time of heel contact (Fz1) and midstance, (Fz2) in non-athlete subjects compared to athlete subjects. More changes in COM-COP were found in non-athlete subjects than athlete subjects. Moreover, a significant difference was indicated between the two groups in terms of the time to peak force, with athlete subjects experiencing higher loading rate for vertical force. **Conclusion**: Due to findings, the effect of physical activities on gait patterns could lead to changes in the interaction and coordination between core stability, GRF, and changes in COM-COP during walking.

Key Words: Activity level; Center of mass; Core stability; Gait; Ground reaction force; Muscle function

Please cite this paper as: Ferasat R, Sadeghi H, Matinhomaee H. The Effect of the Level of Physical Activity on Electromyography of Core Stability Muscles, Ground Reaction Force, and Changes in Center of Mass to Pressure during Gait. J Clin Physio Res. 2021; 6(3): e43. Doi: https://doi.org/10.22037/jcpr.v6i4.35983

Introduction

A variety of factors, including the level of bodily activities, can impact the walking biomechanics, the pattern of muscle use, and the resulting forces. Given that the recording of electrical activity in muscle occurs through the expansion of electrical stimuli within the muscle, the amount and type of muscle movement units in athletes differ from those in non-athletes (1). Walking indicators are also impacted by the level of physical activity, with inactive people acting differently than active people when it comes to walking at the desired speed (2, 3). This seems inactive people have a weaker musculoskeletal system for controlling their balance and forward movement (2, 3). Also, the level of physical activity adjusts the extent to which fatigue increases (2) so that inactive people experience more fatigue compared to active people.

There has been a focus on the speed and the fitness of neuromuscular system as well as the contribution of muscles synergy to momentary stability, maintenance and control of joints, regarding the role of muscles doing life routines and the complex sports skills, as well as the importance attached to the ability of muscle in the generation of force (4, 5).

Accordingly, the muscles of the core stability area play the most important role in the formation of an integrated and coordinated pattern (6, 7). In addition to stabilizing, they also play a movement role in this area (8). Therefore, it is believed that core stability allows for a thorough transfer of the force produced to the upper and lower extremities (6, 7). In contrast, weakness or decreased coordination of the body's core muscles may lead to abnormal motor patterns, corrective motor patterns, and various types of sports injuries such as strain or overuse injuries (9).

Figure 1. The location of electrodes on the selected muscles

According to the above, performing movements such as walking and complex sports movements requires the interaction of sensory and motor systems by the central nervous system. Therefore, kinetics variables have been used to determine neuromuscular functional levels (10). Kinetics variables including ground reaction forces and changes in center of pressure are the factors influencing gait. Measures of the ground reaction force during walking has been considered as a criterion against which individuals are identified and classified based on their pattern of force use during the gait over time (11).

Moreover, it is assumed that center of pressure, as an indicator of the biomechanical mechanism of leg support in stance phase, is closely related to the balance of the body during walking (12, 13).

Although researchers have addressed the role of the muscles related to the core stability in performance, to our knowledge, no study has investigated the effect of the physical activity level on the core stability function (14-17). Indeed, in present study, it was hypothesized that the physical activity level can affect the muscular function of core stability and consequently kinetics variables during gait of male able-bodied individuals.

Material and Methods

In this quasi-experimental study, eighteen young men who meet the criteria agreed to participate in this study and were divided in two groups (athletes and non-athletes).

The criteria for athlete group were having at least 3 years of regular exercise experience for 3 sessions per week, each session for 1.5 hours at least. Non-athlete group had not any exercises during last three years up to time they participated in this study. Accordingly, 20 subjects who covered 20% of the statistical population were recruited. Eleven subjects in the athlete group and 9 subjects in the non-athlete group were included, and at the time of training, two subjects excluded from the study due to injuries.

The study was approved by Kinesiology Research Center ethic committee of Kharazmi University (IR-KHU.KRC.1000.143). Prior to the test, all subjects were briefed on how the test was administered and how the research project was conducted. The consent form and the questionnaire related to medical-sports information and personal information including age, height and weight were completed by the subjects. Having no previous injury, history of surgery, and specific illness were some of the conditions that subjects had to meet in order to participate in the study.

In the phase of data processing, the appropriateness and relevancy of electromyographic signals and the recorded force plate were examined by reliable sources related to the research and drawing on a previous similar research (17).

EMG data collection

The data on muscle function were collected using a 16-channel wireless EMG-MYON Model- made in Switzerland with a sampling frequency of 1200 Hz. In this study, multifidus muscle (MF), longissimus muscle (LO), rectus abdominis muscle (RA), abdominal internal oblique (IO), and external oblique muscle (EO) were identified as the muscles located in core stability area (18, 19). The electrodes were placed between the nerve center of the muscle and the terminal tendon in parallel to the muscle fibers, with a 2 cm distance from the center to the center of the electrodes (20, 21). The electrodes and cables were fixed on the subjects' body using the adhesive tape in order to prevent movement disturbances (Figure 1).

EMG data processing

The noise from the electromyography signals was removed using the band pass filter between 10 and 500 Hz. As frequency with range of 10-500Hz was more important in electromyography, the frequencies with a range above 500 Hz were eliminated using a low-pass filter. However, given that the artifact movements of the electrodes and cables were a source of noise, typically with low frequency, the high-pass filter with a frequency of 10-20 Hz was deemed suitable.

The electromyography data were normalized using muscle MVIC. To this end, subjects were asked to have the maximum isometric contractions in their muscle through performing movements which require maximum level of muscle movement vs. isotonic movements. Each person performed the maximum isometric contraction for five seconds in each muscle, and then the middle three-second RMS linked to the maximum contraction recorded was used to normalize the signal, that is, the RMS of each muscle during walking was divided by MVIC of the same muscle. Then, the result was multiplied by 100 so that the resulting numbers were presented as normal ones and a percentage of the MVIC. Muscle RMS was obtained from a complete cycle of gait.

Figure 2 shows EMG signal processing in a gait cycle using the MATLAB software and based on the drawn diagrams.

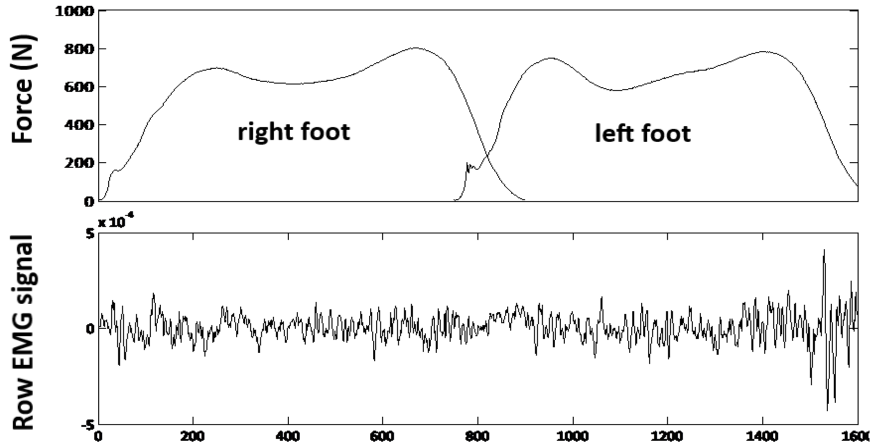

Figure 2. Separating EMG signal through depicting the Fz diagram of left and right feet

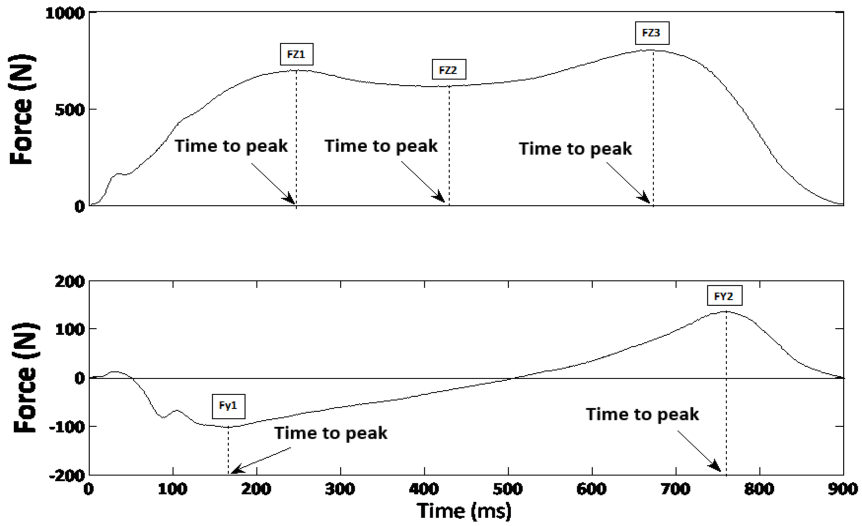

Figure 3. The peak and drop points of Fz and Fy

Kinetics data collection

Two Kistler force plates (Switzerland, 9260AA3, 30*50 and 9260AA6, 50*60) were utilized to determine a gait cycle and gather kinetics information. To do so, the subject's right foot was placed on the first force plate and their left foot was placed on the second force plate.

In the present study, the gait cycle was defined as the time which a subject's right heel strikes the first force plate until the time when the left toe is removed from the second force plate. Given the fact that stance phase of left foot coincides with the right foot swing and left toe off causes another right heel contact, it is fair to claim that a complete cycle is occurred.

In order to have a performance as much similar as to the natural one and to prevent possible changes in the gait pattern due to the focus on gait speed, the subjects were asked to walk at their own selective pace. However, in order to compare the possible effect of the walking speed and to control it in data analysis, the person's walking speed along the route was controlled using a speedometer (Panorama V10c). The average walking speed of the present study was 1.31 m/s which is compatible with a previous study (22).

Kinetic data processing:

Data on the first peak (heel contact, Fz1), the first landing (foot contact during midstance, Fz2), and the second peak (toe off at the end of push off, Fz3) were obtained for the vertical component (Fz) and the time to peak force (Figure 3).

The first landing (heel contact, Fy1) and the first peak (toe off, Fy2) were obtained for the anterior posterior component (Fy), ground reaction force, and the time to peak force (Figure 3).

The following formula was used to obtain the loading rate:

Loading rate=(peak force/body weight)/time to peak

Table 1. demographic characteristics of the subjects

Variables	Athletes	Non-Athletes	T	P-Value
Age (year)	25.44±3.84	24.88±2.66	0.356	0.127
Weight (kg)	80.66±21.86	84.22±15.53	-0.398	0.398
Height (cm)	180.11±9.36	185.44±6.02	-1.437	0.188
BMI (kg/m^2)	24.73±5.49	24.33±4.15	0.172	0.520

Table 2. the mean and standard deviation and results of independent T-test comparisons between vertical (Fz) and posterior-anterior (Fy) components of ground reaction force (N), and time to peak force, and changes in center of pressure (mm)

Variables	Group	Mean (SD)	*P*-value	Variable	Group	Mean (SD)	*P*-value
Fz1	Athlete	100.29 (7.47)	0.124	Time Fz1	Athlete	0.029 (0.003)	*0.007
	Non-athlete	104.76 (10.4)			Non-athlete	0.036 (0.008)	
Fz2	Athlete	80.69 (6.19)	0.295	Time Fz2	Athlete	0.031 (0.003)	*0.017
	Non-athlete	82.44 (10.3)			Non-athlete	0.038 (0.008)	
Fz3	Athlete	106.09 (5.08)	*0.024	Time Fz3	Athlete	0.034 (0.003)	*0.014
	Non-athlete	105.34 (11.07)			Non-athlete	0.041 (0.008)	
Fy1	Athlete	16.65 (3.15)	0.955	Time Fy1	Athlete	0.029 (0.002)	*0.008
	Non-athlete	17.61 (3.13)			Non-athlete	0.036 (0.008)	
Fy2	Athlete	16.39 (1.39)	0.219	Time Cy2	Athlete	0.035 (0.003)	*0.012
	Non-athlete	15.78 (1.9)			Non-athlete	0.042 (0.008)	
COPx	Athlete	10.68 (5.16)	0.606	COPy	Athlete	47.20 (14.06)	0.285
	Non-athlete	16.75 (6.33)			Non-athlete	50.63 (19.53)	
Loading Rate	Athlete	33.33 (6.72)	0.786				
	Non-athlete	28.77 (6.15)					

**significant difference at p≤0/05*

Using the inverted pendulum model, which is related to the path center of mass and the center of pressure, is useful in the dynamic analysis of balance. The changes in pressure center relative to the center of mass of the body in the anterior-posterior (x) and medial-lateral (y) directions were calculated using Winter's inverted pendulum formula (20) as follows:

COP-COM=-Kx

Given that this study utilized two force plates to record information and each of them recorded the coordinates of a separate pressure center (right foot and left foot), changes in center of pressure in the anterior-posterior and medial- lateral directions were calculated using the formula 1 and 2, respectively (22).

Formula 1: $\mathrm{COPx}=\frac{x1fz1+x2fz2}{fz1+fz2}$

Formula 2: $\mathrm{COPy}=\frac{y1fz1+y2fz2}{fz1+fz2}$

x1 and x2 represent the changes in the center of pressure in the anterior-posterior direction for the right foot and left foot, respectively; y1 and y2 denote the changes in the center of pressure in the medial-lateral direction for the right and left foot, respectively; and Fz1 and Fz2 represent the vertical force of the force plate for the right and left foot, respectively.

Force plate information was filtered using the fourth order of lower-pass Butterworth filter and then normalized using the individual's weight. The data were analyzed by MATLAB software.

Statistical analysis

Mean and standard deviation was used to describe data, Shapiro-Wilk Test was applied to verify the normality of the data, and independent t-test was utilized to compare the variables of the two groups at a significance level of $P\leq0.05$.

Results

General information on the subjects

The mean and standard deviations of the demographic characteristics of the subjects are presents in Table 1. There was no significant difference between subjects of two groups with regard to their body mass, age, height and BMI index.

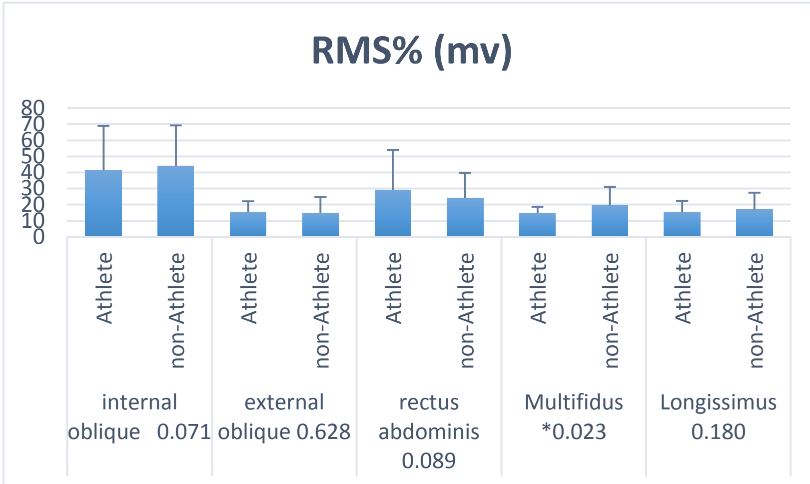

Figure 4. The results of independent T-test for the RMS of selected core stability muscles

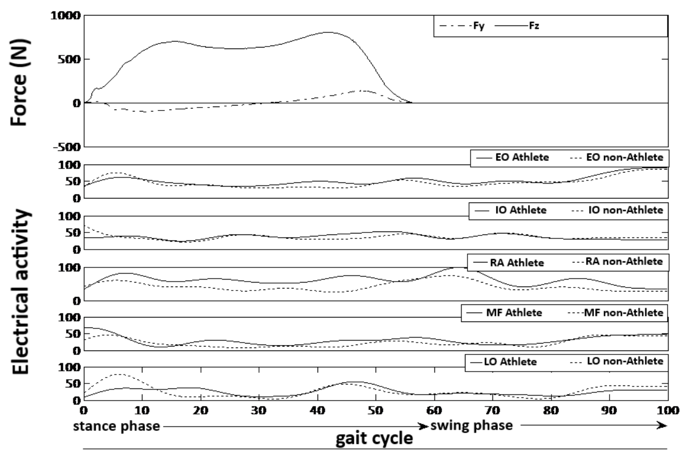

Figure 5. comparing the pattern of muscle function in a gait cycle

RMS and muscle function pattern

The results of the independent t-test showed a signifcant difference between the two groups with regard to MF muscle. Also, the findings indicated higher level of RMS in the IO, MF, and LO muscles of the non-athlete group than in those of athelet group. A higher level of RMS was found in the EO and RA muscles for the athlete group (Figure 4).

The patterns of muscle function in the two phases of gait, i.e. stance and swing were similar, but the electrical oscillations of the muscles in the core stability area were variable in athletes and non-athletes groups (Figure 5).

Components of ground reaction force and changes in center of pressure:

Given that the amount of force at the peak and drop points of the ground reaction force vertical component was higher in the non-athlete group than athlete group, the results demonstrated a significant difference in the second peak (Fz3), while no significant difference was found in the anterior-posterior component.

Moreover, the results showed a higher peak in the vertical and anterior-posterior components of the ground reaction force in non-athlete group. In both components, there was a significant difference between all peaks and drops. Although the changes in the center of pressure showed no significant difference between the two groups, more changes in this parameter were found in the non-athlete group than the athlete group (Table 2). As can be seen, although the loading rate was shown to be higher in the athlete group, there was no significant difference on the loading rates between two groups.

Fz1: the first peak during heel contact, Fz2: the first landing during mid-stance, Fz3: the second peak during toe-off for the vertical component of ground reaction force, Fy1: the first landing during heel contact, Fy2: the first peak at toe-off for the anterior posterior component of ground reaction force.

Time: time to reach peak and landing force, SD: standard deviation

Discussion

This study was driven by a presumption that simultaneous examination of selected muscles function in the target area (to determine the level of neuromuscular function), the variables of the ground reaction forces, and changes in the center of mass to pressure can provide useful information about the patterns of the movement during gait.

The results indicated no significant difference between the force generated at the time of heel contact in the vertical (Fz1) and anterior-posterior (Fy1) components, though the amount of force generataed was higher for the non-athlete group. The stronger reaction force at this stage of non-athlete subjects' gait is likely due to the lack of adequate control and stability in the core muscles. Moreover, the findings of this study corroborate the results of previous studies indicating the muscles of core stability play an important role in the function and coordination of the lower limbs, thereby leading to their stability and power generation (23-25). In other words, the stabilization and functional efficiency require the coordinated activation of the core muscles. This, in turn, requires power control, balance, and activation of core muscles (26). At the same time, results showed that non-athlete subjects experienced more changes in center of pressure compared to athlete subjects. Higher RMS in the IO, MF, and longisimus muscles, which have a more stablizing effect in the body (8) in a complete cycle, indicates more changes in the center of pressure in non-athlete people.

Therefore, low electrical activity oscillations in the IO and longisimus muscles in the heel contact bear witness to the greater stability in the behavioral pattern of the lower limbs during walking given the changes in the center of pressure.

Although higher level of force was generated during the mid-stance phase in the vertical component (Fz2) for non-athlete subjects than athlete subjects, there was no significant difference. This phase occurs immediately after heel contact, and the intensity of force can be attributed to the contractions of tibialis posterior as well as the calf muscles. It seems that the muscles of the core stability have no direct effect on the magnitude of ground reaction force at this phase. The pattern of muscle function showed that the electrical activity of the muscles at the foot contact phase, in most of the muscles of the core stability, seemed to be insignificant for both athlete and non-athlete groups.

However, the level of activity, *i.e.* athletic vs. non-athletic, which strengthens and weakens the muscles in this area, respectively, leads to differential control of the ankle plantar flexion against the gravity and body weight done by eccentric contraction of the tibialis anterior leg muscle. This phenomenon may be one of the reasons for the increased force of the reaction force in non-athletes' mid stance given the weaker control of the ankle plantar flexion by the tibialis anterior muscle.

Based on the findings, the amount of force at the time of toe off was higher in the anterior-posterior component (Fy2) and in the vertical component (Fz3) for the athlete group than in non-athlete group. Thigh movement, which is directly affected by the core stability, is very important at toe off, resulting directly in the creation of ground force reaction. Indeed, it seems that the muscles of the core stability play an crucial role in creating a stable support surface for the movements of the limbs (27) including the thighs.

It is claimed that even if the upper and lower limbs are strong, but the core muscles are weak, a decrease in the overall forces of core muscle of the trunk can lead to a reduction in the force generated in upper and lower limbs (28). It seems that the use of stronger core muscles in athletes compared to non-athletes is one of the factors contributing to the creation of stronger reaction force at toe off.

The pattern of muscle function in the core stability area also shows that athletes experience more electrical activity than non-athletes at toe off. Therefore, an increase in ground reaction force at toe off can be attributed to the greater electrical activity of the athlete's muscles, which have more movement's role at this moment. This conclusion is also corroborated by finding the significant difference in the vertical component (Fz3) at toe off.

There is debate over whether or not the ground reaction force and higher loading rates during walking in athletes make them more vulnarable to injury. Weakness or decreased coordination of the core muscles can lead to abnormal motor patterns, corrective motor patterns, and various types of sports injuries such as strain or overuse injuries (9).

The results of the present study regarding the loading rates appear to be inconsistent with the previous studies (29) since the non-athletes who appear to have weaker core stability muscles, experience lower loading rates and hence lower risk of injury. However, it may be fair to say that the acceleration and higher speed of the athletes' lower limbs due to the efficient and stronger muscles during walking shortens the time required to reach peak force.

Therefore, the lower the time to peak force, the higher the loading rate. On the other hand, it is claimed that higher loading rate is also associated with the material stiffness (29). As athletes have less fat tissue in their subcutaneous layers especially in the heel and have more muscle stiffness, it seems that musculoskeletal stiffness due to sport activites is one of the reasons for the higher loading rate in the athletes.

Conclusion

The findings of this study showed the interactive effect of physical activities on the neuromuscular function of core stability as well as the behavioral pattern of changes in Center of Mass to Pressure and ground reaction force during walking in athletes and non-athletes.

Acknowledgment:
None

Conflict of interest:
None

Funding support:
None

Authors' contributions:
All authors made substantial contributions to the conception, design, analysis, and interpretation of data.

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