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Changes of trunk muscle synergies due to different insole wedges during squatting in athletes

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ABSTRACT

Background: Squatting is an important and common movement repeatedly performed during sports and daily activities. Insoles have been used as an intervention for several purposes, although their effect on trunk muscle synergy among athletes is not clear. The present study aimed to investigate the influence of different insole wedges on trunk muscle synergies during squatting in athletes.

Methods: Twenty-eight (n=28) healthy athlete males participated in this study voluntarily. They performed a squatting task with 20% of their body mass. Pearson correlation coefficient and repeated measures analysis of variance tests were used to assess the similarity and compare trunk muscle synergies among five wedge conditions: non-wedge, anterior-medial, posterior-medial, anterior-lateral, and posterior-lateral wedges ($P < 0.005$ for Bonferroni correction).

Results: The result of the analysis identified three synergies for trunk muscles in all wedge conditions during squatting. In the first and third synergy, the activation pattern of trunk muscle synergy was highly similar among all wedge conditions ($r > 0.9$). In the second synergy, a moderate similarity was observed between the non-wedge with the anterior-medial and anterior-lateral wedge conditions ($r = 0.584$ and 0.654, respectively). In all three synergies, no significant difference was observed in the relative weight of muscles among wedge conditions ($P > 0.005$).

Conclusions: It seems that muscle activity levels have only minor effects on the structures of muscle synergies during squatting. These changes may be a strategy by the central nervous system (CNS) to optimally control squat movement. Future studies are necessary to find other outcomes of these interventions on injury prevention or performance improvement.

K E Y W O R D S: Squat, Electromyography, Muscle Synergies, Footwear, Athletes.
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Introduction

Many sports and daily activities of individuals are assigned to complete various work-related tasks. Among these tasks, squatting is an important and common movement repeatedly performed during sports and daily activities [1]. The lumbar region undergoes significant mechanical load during squats, which may cause various musculoskeletal disorders, such as low back pain (LBP) [2, 3].

Generally, risk factors for sport-related LBP can be divided into two categories: external or environmental factors and internal factors [4]. Several attempts have been made to identify and improve external factors in order to affect the LBP prevalence by manipulating the load weight or location [3] and teaching or training [5, 6]. These efforts have not been entirely successful [3, 5], and refocus research on modifiable internal factors that might mediate loading on the lumbar spine is needed (e.g. biomechanical aspects of movement). In this regard, recent studies have investigated kinematics [7, 8], kinetics [8], electromyography (EMG) [9], body balance [10], and spinal load and stability [11, 12] during squatting. Many biomechanical aspects of this task related to movement patterns have not yet been identified, which highlights the need for further research to fully understand effective movement strategies. This enables researchers to effectively identify risk factors and reduce the risk of injury via a multifactorial approach, including the use of effective interventions [13].

During squatting, it is necessary for both global (large superficial muscles more adapted to spinal movement) and local (small deep muscles more suited as spine stabilizers) trunk muscles to contract in a coordinated manner [14], modulated by the central nervous system (CNS). A new model, assuming human-body movements are controlled by a unit of muscle groups (vs. a single muscle), was recently presented to determine the control of motor performance [15, 16]. These units of muscle groups, called modules or synergies, are activated by the CNS [17]. Mirakhorlo and Azghani (2015) showed that despite kinematic differences, trunk muscle synergies were similar among various squat movements [18]. However, Tan et al. (2019) observed that the use of a lumbar support exoskeleton changes some trunk muscle synergy indices during repetitive stoop lifting [16]. It seems that there is no well-studied and also common agreement on the factors affecting muscle synergy during squatting.

Insoles have been shown to be useful interventions to reduce the risk of work-related LBP [19–21], because of their effect on the kinematics and kinetics of the lower extremities [20, 21]. Furthermore, studies have shown that squat tasks can be performed more efficiently while using anterior and medial insole wedges probably because of the higher body balance [10], lower activation levels of lower limb muscles [22], and higher trunk muscle co-contraction [14].

The assessment of muscle synergies can provide insight into how the musculoskeletal system is loaded and thereby lead to effective strategies that might decrease or alter those loads [16]. To the author's knowledge, this is the first study conducted to identify the effect of different insole wedges on trunk muscle synergies in healthy athletes during squatting. Therefore, the aim of the study was to investigate how insole wedges influence trunk muscle synergies during squatting.

Material and Methods

Participants

The minimum number for sample size was estimated using G*Power software (n = 26, number of measurements = 5, effect size = 0.25, statistical power = 0.9, and alpha level = 0.05). In the present study, twenty-eight (n = 28) male athletes participated (age = 23.04 ± 3.23 years, mass = 67.65 ± 6.24 kilograms, height = 1.77 ± 0.05 meters). The criteria for inclusion were: being healthy, doing regular exercise (twice a week in local gyms in the last two years), having no malalignments or musculoskeletal injuries in lower limbs, providing written consent before participation in the research, and having the ability to perform dumbbell (20% body mass) goblet squat. The choice of 20% was made to equalize and control individual differences among subjects. This percentage of weight was the minimum amount that all subjects could perform the dumbbell goblet squat without expressing discomfort and noticeable errors, including increasing lumbar lordosis, excessively moving the trunk forward, having valgus of the knees, lifting the heel, and shifting the trunk to one side. The study was confirmed by the University Research Ethics Committee (code number: IR.SSRI.REC.1401.1611).

Task description

One day before testing, subjects were trained to perform the dumbbell goblet squatting with video instruction and then practiced until successful performance. They held a dumbbell (20% body mass) at the sternum height with both hands and then descended with the knee and hip flexion at normal self-speed for about 1.5 seconds until the knee joint angle reached 90 ± 10 degrees using verbal instructions. They maintained this position for about 1.5 seconds and then ascended to the standing position in the same way. They wore the identical standard shoe (Gel Nimbus 11, Asics, Vietnam) during all squat movements (Figure 1).

Figure 1. The shoe used in this research.

Data acquisition

The activity of trunk muscles was recorded using a Biomonitor EMG system (ME6000, Mega Electronics, Finland, amplifier system gain = 1000, common mode rejection ratio = 90 dB, accuracy = ±2.5%, input signal range = ±5 Volts, maximum noise = 1.6 μV, sampling frequency = 1000 Hertz). The knee joint angle was measured by an electrogoniometer (Biometrics Ltd., England) during squatting to identify the beginning and ending time [14]. After skin preparation, including hair abrasion and cleaning with alcohol, surface Ag/AgCl electrodes were attached in a bipolar arrangement, parallel to muscle fibers, with inter-electrode distance of 25 mm over the right selected trunk muscles, including rectus abdominis (RA: 2 cm lateral and 3 cm superior to umbilicus) [14, 23], abdominal external oblique (EO: 15 cm lateral to umbilicus) [14, 23], abdominal internal oblique (IO: approximately 2 cm medial and inferior to the Anterior Superior Iliac Spine and at an angle facing the umbilicus) [14, 23], latissimus dorsi (LD: most lateral portion of muscle at T9 level) [14], thoracic erector spinae (TES: 5 cm lateral to the ninth thoracic spinous process) [3, 14], lumbar erector spinae (LES: 5 cm lateral to the third lumbar spinous process) [3, 14], multifidus (MU: 2 cm lateral to midline and centered at the level of L5 spinal process) [14, 23], and quadratus lumborum (QL: 4 cm lateral from the vertebral ridge of the belly of the erector spinae muscle and at a slightly oblique angle at half the distance between the 12th rib and the iliac crest) [14]. The placement of electrodes for neutral zone (7th cervical vertebrae) was considered according to a recent study [14]. Finally, the subjects accomplished maximum voluntary isometric contractions (MVICs) to normalize the amplitude of EMG signals using verbal encouragement (repeat number for each trial = 3 times, test duration = 3 seconds, resting time between tests = 3 minutes) [14]. For each muscle, the highest peak value of MVIC trials was considered for amplitude normalization.

Using Solidwork software version 2011, five insole types were designed with various wedges, including non-wedge (NW), anterior-medial (AM), posterior-medial (PL), anterior-lateral (AL), and posterior-lateral wedges (PL). The insoles had similar material, and wedge angles were chosen based on previous studies [10, 14]: 7.1º and 2.2º respectively for the medial-lateral and anterior-posterior directions (Figure 2).

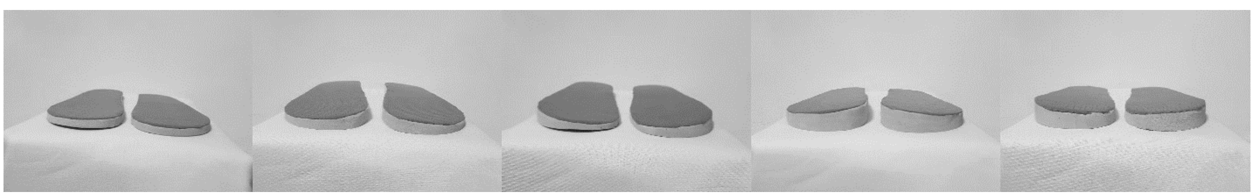

Figure 2. Printed insoles in five conditions (from left to right: NW, PM, PL, AM, and AL).

To calculate the average value for each dependent variable, six successive squatting tests were performed for each condition. The random order of conditions and 30 minutes of resting time among conditions were given to avoid learning and fatigue effects, respectively [14].

Data Processing

The processing steps of the EMG signal were as follows: band-pass filtration (10-400 Hertz), full-wave rectification, filtration with a zero-lag Butterworth second-order filter with low-pass cut-off frequency 2.5 Hertz, amplitude normalization with MVIC peak value, interpretation of data as a percent (multiplying by 100), and time normalization to 300 datapoints (100 datapoints for each descending, holding, and ascending phases of squat lifting task).

To obtain muscle synergy components, including neural commands (NCs) and synergy vectors (SVs), we used a non-negative matrix factorization (NNMF) algorithm, which converts a matrix of all EMG signals (k frames × n muscle) into NCs and SVs matrices [15, 24]. The NCs matrix is the temporal activation pattern of each module (k frames × c modules), and the SVs matrix contains the weights of each muscle in each module (c modules × n muscle). The number of required synergies for the reconstruction of the original matrix was determined using the variance accounted for (VAF) method, where the first assumption is that one synergy is enough for the reconstruction of the original signal matrix: if the VAF of all muscles in all participants reaches or exceeds 0.9, then the addition of another synergy is not needed; if not, another synergy is added until the minimum number of synergies in the VAF of all muscles in all individuals reaches 0.9 [15, 24].

Statistical Analysis

SPSS version 19 software was used for all statistical tests, with a significance level of 0.05. The two-way random model of the intraclass correlation coefficient (ICC) test showed a value of more than 0.7 for all dependent variables, which indicates that all dependent variables have high relative reliability based on Munro's classification for reliability degree [25]. Moreover, the normality of data distribution and homogeneity of variances were confirmed by Kolmogorov-Smirnov and Levene's tests ($P>0.05$). To compare dependent variables among different insole wedge conditions, we used one-way repeated measure analysis of variance tests, with Mauchly's test for sphericity assumption, Greenhouse-Geisser correction for violated sphericity situation, and Bonferroni correction for post-hoc comparisons and avoidance of type I error violation ($P<0.005$). Also, Pearson correlation coefficient analysis was used to assess the similarity of synergy patterns (NCs) between the NW with other wedge conditions, where the correlation value for small similarity is between 0.0-0.29, for moderate similarity between 0.3-0.69, and for high similarity between 0.7-1 [26].

Results

The number of synergies

VAF results showed that three synergies were sufficient for reconstruction of the original signal in all five insole wedge conditions (Figure 3).

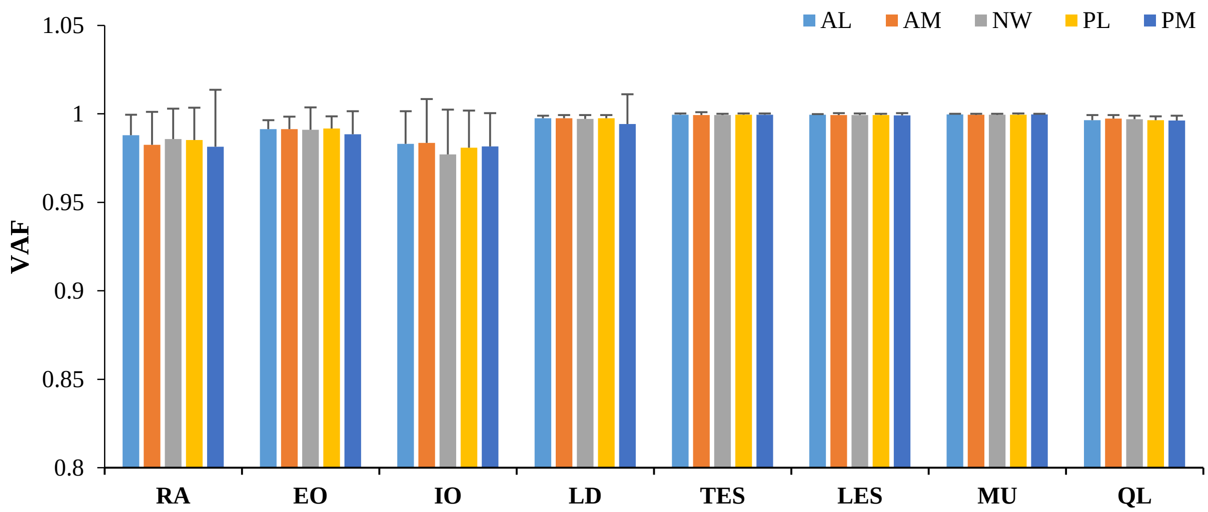

Figure 3. VAF values (mean and standard deviation) of all muscles using 3 synergies in five insole wedges.

Effect of different insole wedge conditions on the synergy activation pattern of trunk muscles in synergies

Results related to the effect of different insole wedge conditions are seen in Table 1. In the first synergy, the activation pattern of this synergy showed high similarity among different insole wedge conditions ($r > 0.97$). In the second synergy, a moderate similarity was observed between the NW condition with the AM and AL conditions ($r = 0.584$ and 0.654, respectively), but the PM and PL conditions showed a high similarity to the NW condition ($r = 0.868$ and 0.707, respectively). In the third synergy, using different insole wedge conditions showed a high similarity to the NW condition ($r > 0.9$).

Table 1. Correlation of mean synergies between non-wedge with other wedge conditions.

	AL		AM		NW		PL	
NC 1								
	r	p-value	r	p-value	r	p-value	r	p-value
AM	.981	.001						
NW	**.976**	**.001**	**.987**	**.001**				
PL	.993	.001	.973	.001	**.973**	**.001**		
PM	.985	.001	.990	.001	**.989**	**.001**	.981	.001
NC 2								
	r	p-value	r	p-value	r	p-value	r	p-value
AM	.737	.001						
NW	**.584**	**.001**	**.654**	**.001**				
PL	.869	.001	.825	.001	**.868**	**.001**		
PM	.813	.001	.906	.001	**.707**	**.001**	.879	.001
NC 3								
	r	p-value	r	p-value	r	p-value	r	p-value
AM	.940	.001						
NW	**.963**	**.001**	**.930**	**.001**				
PL	.966	.001	.921	.001	**.984**	**.001**		
PM	.975	.001	.968	.001	**.951**	**.001**	.937	.001

Effect of wedge conditions on synergy vectors

Results related to the effect of different wedge conditions on synergy vectors are seen in Figure 4. In the first and second synergy, there was no significant difference in the relative weight of trunk muscles in different insole wedge conditions (P > 0.05). In the third synergy, there was also no significant difference in the relative weight of the trunk muscles in different insole wedge conditions (P > 0.05), although the relative weight of MU increased in the AL condition compared to the NW condition (mean difference = 0.104; P = 0.049).

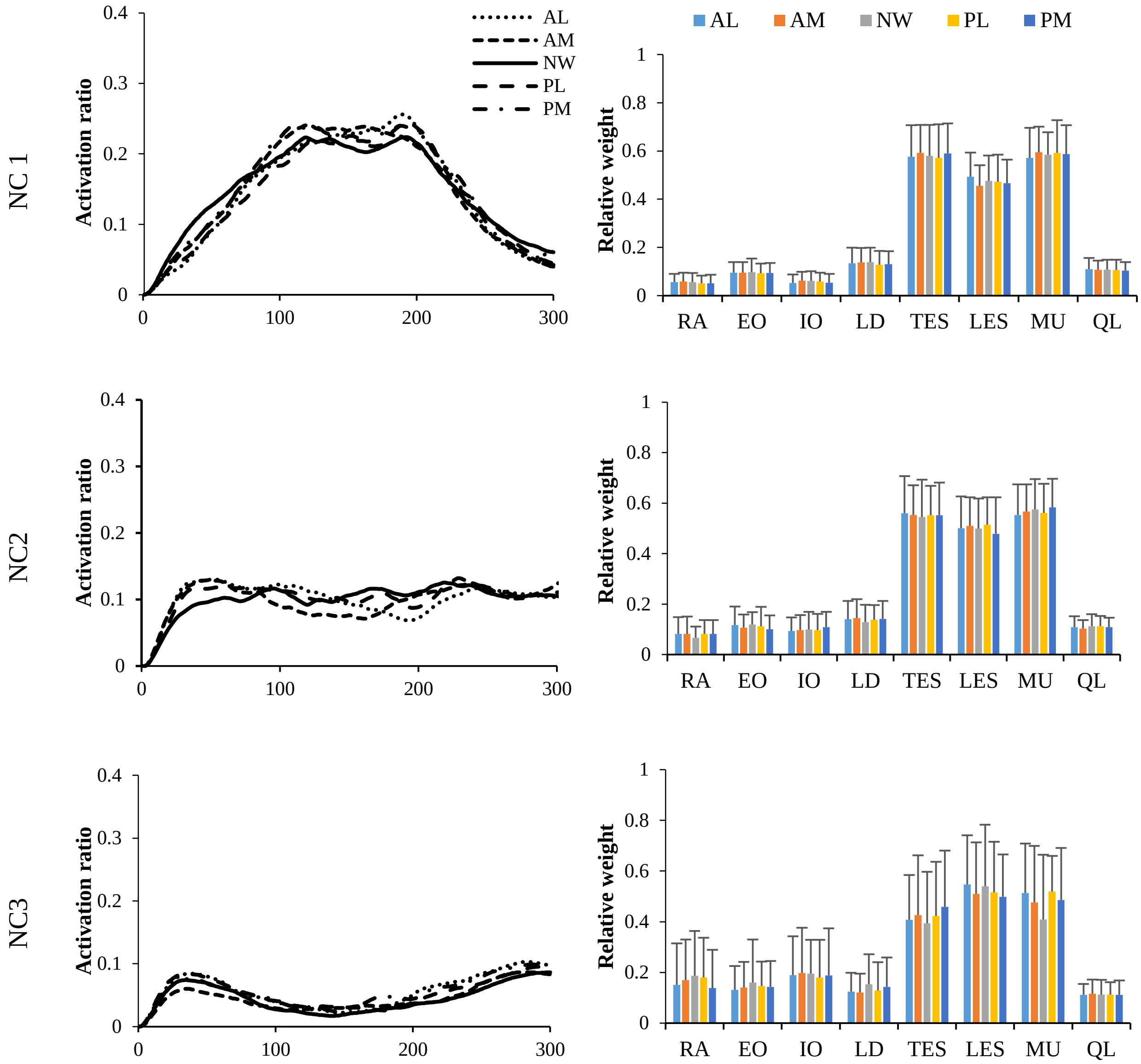

Figure 4. Mean NCs (neural commands) (left) and SVs (synergy vectors) (right) for three synergies. Right plots show the mean plus standard deviation values of muscle weights in five wedge conditions.

Discussion

The present study aimed to investigate the influence of different insole wedges on trunk muscle synergies during squatting. We found three synergies for the reconstruction of the original signal in all five insole wedge conditions during squatting, which is consistent with the result of some previous studies [16, 18, 27].

We also observed that although the activation pattern of trunk muscle synergy was highly similar among conditions in the first and third synergies ($r > 0.9$), a moderate similarity was observed between the NW with the AM and AL conditions in the second synergy ($r < 0.7$). These findings are congruent with the study of Smale et al. (2016) who showed that a small change occurs in the synergy of lower extremity muscles due to a fatigue protocol in single and double-leg squatting movements [27]. Tan et al. (2019) also observed that two factors of fatigue and intervention (lumbar support device) had a subtle effect on muscle synergy vectors during stoop lifting [16]. In contrast, our findings are incongruent with the study of Mirakhorlo and Azghani (2015), who reported that the VAF synergy and related time-varying coefficients for trunk muscles were similar during different squat techniques [18], but their methods and results were based on modeling data which can have different results from in-vivo data. Turpin et al. (2020) also showed similar synergy structures of upper extremity muscles during load lowering and lifting [28]. However, they assessed synergies of 13 muscles in the shoulder area, which is different from the method of the present study.

No significant difference was found in the relative weight of trunk muscles among insole wedge conditions in all three synergies, although the relative weight of multifidus muscle tended to increase in the PL and AL conditions compared to the NW condition. Several studies have tried to increase body stability using various modalities during squatting [10, 29]. Different insole wedges have been shown to influence body stability [10] and change the activity level of specific trunk stability muscles during lifting [14], although insole material can also be import [33]. Wojtara et al. (2014) observed a positive correlation between balance situation and synergy stability index during lateral perturbation while standing [30]. Without balance data, we can just speculate that different insole wedges would alter the contribution of local muscles (e.g., multifidus) in synergies by influencing balance.

The present study showed a subtle change in the relative weight of trunk muscles. This may indicate that the CNS has a flexible manner to change the pattern of synergy from very similar to moderate in different insole wedges. Hence, it may be plausible to use interventions such as insole to change trunk muscle synergies in athletes. The origin of muscle synergy structures is related to the CNS [31]. Hajiloo et al. (2020) showed that some factors like fatigue change only relative muscle weights rather than muscle synergy structure, which this strategy is probably created by the CNS to optimally maintain motor system function [15]. However, as muscle synergy is a new approach in biomechanical studies, combining methods are better way to identify other influential factors [27].

While insoles have immediate effects [10, 14, 32], they are commonly considered as a long-term intervention method [19]. Therefore, future studies are needed to assess the long-term effects of using different insole wedges on muscle synergies during sport-related techinques (e.g., squat). Also, we only had healthy athletes (rather than injured or LBP subjects). Since to the author's knowledge, this is the first study conducted to identify the effect of different insole wedges on trunk muscle synergies, future studies are recommended aiming to use these interventions on the improvement of athletes' performance.

Conclusion

Based on our findings, it can be inferred that various insole wedges have a minor effect on muscle synergies and relative weights. These changes may be a strategy by the CNS to optimally control squat movement.

Ethical Considerations

The study was confirmed by the University Research Ethics Committee (code number: IR.SSRI.REC.1401.1611)

Compliance with ethical guidelines

All procedures performed in this study were conducted in accordance with institutional and national ethical standards and the principles outlined in the Declaration of Helsinki.

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Conflict of Interest

No potential conflict of interest was reported by the authors.

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«مقاله پژوهشی»

تغییرات سینرژی عضلات تنه بر اثر وِج های مختلف توکفشی حین حرکت اسکات در ورزشکاران

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چکیده

هدف: اسکوات یک حرکت مهم و رایج است که به طور مکرر حین ورزش و فعالیت‌های روزانه انجام می‌شود. توکفشی‌ها به عنوان مداخله برای اهداف مختلفی مورد استفاده قرار گرفته‌اند، اگرچه تأثیر آن‌ها بر سینرژی عضلات تنه در ورزشکاران مشخص نیست. مطالعه حاضر با هدف بررسی تأثیر انواع مختلف وِج (گُوِه) توکفشی بر سینرژی عضلات تنه حین حرکت اسکوات در ورزشکاران انجام شد.

روش‌شناسی: تعداد ۲۸ مرد ورزشکار سالم به‌طور داوطلبانه در این پژوهش شرکت کردند. شرکت‌کنندگان حرکت اسکوات را با ۲۰ درصد وزن بدن خود انجام دادند. ضریب همبستگی پیرسون و آزمون‌های تجزیه‌وتحلیل واریانس اندازه‌گیری‌های مکرر برای ارزیابی شباهت و مقایسه سینرژی‌های عضلات تنه بین پنج وضعیت بدون وِج (گُوِه) و وِج‌های قدامی–میانی، خلفی–میانی، قدامی–جانبی و خلفی–جانبی استفاده شد (با تصحیح بونفرونی).

نتایج: نتایج تجزیه‌وتحلیل سه سینرژی را برای عضلات تنه در تمام شرایط وِج حین حرکت اسکوات شناسایی کرد. الگوی فعال‌سازی سینرژی عضلات تنه در سینرژی اول و سوم بین تمام شرایط وِج بسیار مشابه بود (r=۰/۹). در سینرژی دوم، شباهت متوسطی بین شرایط بدون وِج با شرایط وِج قدامی–میانی و قدامی–جانبی مشاهده شد (به ترتیب ۰/۵۸۴ و ۰/۶۵۴ = r). در هر سه سینرژی، تفاوت معناداری در وزن نسبی عضلات بین شرایط مختلف وِج مشاهده نشد (P > ۰/۰۰۵).

نتیجه‌گیری: به نظر می‌رسد که سطح فعالیت عضلات تنها تأثیرات جزئی بر ساختارهای سینرژی عضلات حین حرکت اسکوات دارد. این تغییرات ممکن است یک استراتژی توسط سیستم عصبی مرکزی برای کنترل بهینه حرکت اسکوات باشد. مطالعات آینده برای بررسی سایر نتایج این مداخلات در پیشگیری از آسیب یا بهبود عملکرد ضروری است.

واژه‌های کلیدی: اسکات، الکترومیوگرافی، سینرژی عضلات، پاپوش، ورزشکاران.

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