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**Original Research**

# The Effect of Inertia and Initial Body Posture on Lower Extremity Kinetics and Trunk Muscle Activity in Lifting

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## ABSTRACT

Objectives: Because the function of the lower limb joints and the role of the trunk muscles in bearing and lifting loads are important, this study aimed to investigate the intervention of the initial body position when controlling inertia and its effect on the strategy of kinetic patterns in the lower limb joints as well as trunk muscle activity during symmetric lifting. Methods: This experimental study included 10 healthy young man. They were asked to lift an 8.4 Kg box at 3 different knee angles (-10, 35, and 70 degree angle of the thigh to the horizon line) symmetrically and at a constant velocity. The extensor moment and power of lower limb joints were calculated using the 3D linked-segment model (LSM). To measure the muscular activity of abdominal muscles (rectus abdominus (RA) and external oblique (EO)) and lumbar muscles (iliocostalis lumborum (IL) and multifidus (MU)), an 8 channel electromyogram (EMG) were used. Results: The results showed the similarity of extensor moment pattern of lower limb joints and lumbar joint (L5/S1) in accelerated lifting and constant velocity lifting with similar techniques. The results revealed that there was no significant effect of squat lifting while controlling inertia on lumbar extensor moment and back muscles activity. In all knee postures. There was also a significant effect of different knee postures on the extension power of lower limb joints. Discussion: According to the findings, in the lifting without the effect of the inertial force of the lifted box, lumbar extension moment and back muscle activity was independent of different knee postures. However, the extensor power of the lower limb joints was affected by the knee postures. Moreover, the lumbar extensor moment and lumbar muscles activity in different knee postures had a similar behavior.

**Keywords**: Knee posture, Trunk muscle activity, Inertia controlling, Squat lifting

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## INTRODUCTION

Load lifting and transferring are very applicable and frequently used in different industries and dangerous jobs. Frequent lifting is the main cause of pain in the lumbar region (1). Mechanical loads also are one of the most important risk factors resulting in back pain (2). Load lifting from ground level causes compressing forces in the lumbar spine (3). These forces as a result of the trunk muscle activity and extensor moment that applied to neutralize and reduce shear stresses exerted to the lumbar spine (4). Moreover, the lumbosacral joint (L5/S1) is one of the most sensitive critical points in the lumbar region which endures a maximum extension moment and receives high risks. For this reason, understanding the kinetic patterns of lower limb joints and the strategy of muscular activity of trunk muscle to determine and measure critical values for lumbar spine injuries in different lifting positions are necessary and helpful. So, many ergonomic studies have been focused on effective parameters of loads exerted on the lumbar region. They usually have investigated lumbar loads using extensor moment of joints and muscular activity of trunk muscles.

During load lifting, extensor moments, and muscular activity of trunk muscle is affected by many factors. Understanding the effective factors exerted on the lumbar region during lifting is of great importance, which can be very useful in measuring risk factors and determining preventative effective values. In this regard, several studies investigated the involved parameters to reduce exerted force in the lumbar region, such as load weight (5), load-lifting velocity (6, 7), horizontal distance (8), vertical distance (9) and load lifting techniques (10). Initial body posture is one of the most effective parameters as a risk factor for back pains which is specifically focused on by researchers (11). Despite the different opinion of researchers about different lifting methods, which are often due to differences in the effect of interventional variables and how lifting is performed, most researchers have suggested Squat lifting (12, 13). On one hand, Squat lifting is a multi-joint movement in contrast to the Stoop method which the former includes more trunk involvement and other lower limb joints. On the other hand, complaints and injuries relative to the Squat technique are less than the Stoop technique (14). For this reason, the Squat technique is considered a better technique rather than common ones and researchers are more encouraged to investigate this technique. Moreover, in many studies, the Squat technique was performed in different forms of knee flexion from low to full flexion which causes different results.

Inertia is one of the other effective parameters affecting exerted load to lumbar joint (L5/S1) which received more attention from many researchers. Marras & Mirka showed that the increased of trunk extension velocity that increases the applied load to the lumbar joint (15), the muscular activity of trunk muscles (16) and lumbar extensor moment (17). These findings strengthen kinematic and kinetic changes of the trunk due to changes in lifting velocity. Besides, some of these contradictions lower peak moment of Stoop method respect to Squat technique in static linked-segment model (LSM), Van Dieen et al., 1999 to prefer Squat technique respect to Stoop method between static and dynamic linked-segment model is also due to ignoring the inertia factor (acceleration) in static models (18). Therefore, inertia inhibition (the velocity of load-lifting) is of great importance for the reliability and accuracy of obtained results and needs more investigation.

So, inertia and initial body posture during load lifting are the most effective factors resulting in kinetic changes and muscular activity of trunk muscle and there might be a different result in load lifting in different body postures (knee flexion angle) in Squat technique while inertia controlling. Therefore, in this study, the lower limb kinetic (extensor moment, power of lower limb joints and lumbosacral joint L5/S1) were estimated using the Finite elements dynamic model by taking in to account a wide range of knee flexion (low flexion (LF): 70 degrees of tight angle respect to the horizon, semi flexion (SF): 35 degrees of tight angle respect to the horizon and full flexion (FF): -10 degree of tight angle respect to the horizon) and inertia controlling during movement. To matching and verifying the results, the muscular activity of trunk muscle was evaluated based on maximum voluntary contraction (MVC).

Since many studies have investigated the effect of work performing variables on extensor moment and trunk muscle activity level qualitatively and quantitatively, but none of them have not evaluated extensor moment of the lower limbs, lumbosacral joints, and muscle activity of trunk muscles while controlling the effect of inertia (constant velocity) on different knee flexion in Squat technique. The findings of this study can be used to measure and determine the risk factors in spatial-temporal coordination as well as to detect the effect of long-term variables during load lifting in high-risk jobs and other industries.

## MATERIAL AND METHODS

### Participants

In this study, 10 healthy young men were voluntary selected (age: 23.54±1.90 years old, weight: 77.65±12.34 kg, Height: 177.75±6.44 cm, BMI: 24.69±4.51 Kg/m2) (19). They had no back pain history, musculoskeletal disorder, or professional experience of load lifting. The load was set at 8.4 Kg (Equivalent to 11% of the average weight of the subjects) for all subjects. All procedures and ethical principles were verified by the ethical committee of the biomechanics department of medical technology and science faculty of Islamic Azad University (science and research branch) according to the World Medical Association Declaration of Helsinki.

### Test protocol (Lifting Task)

The subjects were asked to lift a box (32 [w] cm× 40 [L] cm × 25 [H] cm) while holding the handles (20). The box was placed on the mid-sagittal plane in front of their feet, and the distribution of the weight in the box is also symmetrically considered in the center of the box and is located 40 Cm from the lateral malleolus on the ground (21). The symmetric load distribution was intended to prevent lateral movements and instability during lifting.

Each subject lifted the box based on the Squat technique in different knee flexion with constant velocity (2 seconds for each lifting) (22). The lifting velocity was measured by the metronome. While the subject holds the handles (lordotic position), the knee flexion angle in each lifting was determined by the examiner. According to Fig. 1, three flexion angles were set: FF full flexion position (-10 degrees of tight angle respect to the horizon), SF semi flexion (35 degrees of tight angle respect to the horizon), and LF low flexion (70 degrees of tight angle respect to the horizon) (22, 23,24). The initial body posture was differently set during lifting in each task to be able to evaluate its effect on trunk muscle activity, the kinetics of the lower extremity, and lumbar joints. Each lifting was performed three times with 2-minute relaxing intervals. Lifts were randomly performed to prevent the effect of exercise or fatigue. The subjects were asked to not exercise about a week before the first test to ensure the validity of results. A training session was conducted three days before the test for subjects to familiarize themselves with the procedure. Moreover, all test steps were instructed to subjects by examiners before the test (20).

A: B: C:

Fig. 1: **Initial body postures at three knee flexion angles (FF (A), SF (B) and LF (C)).**

## Electromyography

An eight-channel EMG (DataLOG MWX8, Biometrics LTD., lon, UK) of the lower back muscles and abdominal muscles were used to analyze muscle activity. Before each experiment, surface EMG electrodes were attached to the participant's carefully cleansed skin as follows. Bipolar silver-silver surface electrodes (with a diameter of 10 mm and an inter-electrode distance of 20 mm) were placed over two pairs of lower back muscles as extensors (multifidus and iliocostalis lumborum) and two pairs of abdominal muscles as flexors (external oblique and rectus abdominis). Because multifidus is a deep muscle that is covered by superficial muscles, crosstalk could occur between muscles during signal recording. To minimize such crosstalk, the following steps were taken. Electrodes were placed carefully within the borders of the muscles and parallel to the muscle fibers. The skin impedance was measured and was accepted if it was less than 5000 U. Multifidus electrodes were placed at the L5 level, parallel to the line between PSIS and the L1-L2 interspinous space, at which the multifidus fibers are shallower, as many researchers believe multifidus activation can be detected by surface EMG (25,40). The iliocostalis lumborum electrodes were placed at the L2 level, parallel to the line connecting the posterior superior iliac spine (PSIS) to the lateral border of the muscle at the 12th rib. External oblique muscle electrodes were placed just below the rib cage, along the line connecting the most inferior point of the costal margin to the contralateral pubic tubercle. Rectus abdominus electrodes were placed 1 cm above the umbilicus and 2 cm lateral to the midline (26,27). The procedure used for EMG testing of back and abdominal muscles was consistent with that of Ng (28). EMG signals were band-pass filtered at 10-250 Hz and sampled at 1000 Hz and with 24-bit resolution, and A–D converted and stored synchronized to VICON Motion Analysis (Bonita-10 VICON camera , UK) and two force plate (AMTI, AccuGait Optimized Multi-axis, US). EMG signals were then rectified, and the root mean square value was calculated using a time constant of 4 ms. To normalize the EMG data, the maximum voluntary contractions (MVCs) in the extension (for back muscles) and the flexion and lateral flexion on both sides (for abdominal muscles) were measured twice for 5 s in each subject with a 2-min rest interval between trials. For the MVCs of the extensors, subjects were asked to lie in a prone position with their hands beside their bodies and their legs extended (20). Resistance was applied to the upper thoracic area in the direction of the trunk flexion. For the external obliques on the right and left sides, subjects were asked to assume a supine position with knees flexed and hands behind their heads. Subjects flexed their trunk and rotated to the right and left. Resistance was applied at the shoulders in the trunk extension and right and left rotation directions. For the MVC of the rectus abdominis, subjects were asked to assume a partial sit-up position with their hands behind their heads and knees flexed. Subjects flexed their abdominal muscles, and resistance was applied to their shoulders in the trunk extension direction. Subjects were asked to gradually increase the force to reach an absolute maximum force. Contractions were executed in a randomized order, and one warm-up trial was performed before each test. The subjects were asked to exert their maximum effort and avoid jerky contractions during the MVC tests. To obtain high reliability in the EMG measurements, the same procedures were reproduced exactly in the second testing session.

## Motion analysis & kinematic and kinetic data collection

Markers and forces data were filtered by using a 4th order Butterworth low-pass filter at a cut-off frequency of 6 Hz. A dynamic bottom-up 3D linked segment Model (LSM) comprising 19 segments (Head, Neck, Upper Arms, Lower Arms, Hands, Thorax, Lumbar (L1-S1), Pelvis, Thighs, Shanks, Foots) and 54 passive reflective markers. Marking based on the standard template full-body VICON Plug-in Gait Model with the 15 additional markers were added to the model to increase the accuracy of the kinetic calculation, including internal markers of the joints, L5/S1, anterior lower limb, 5th Metatarsal, and Hips markers (29, 41). Kinematics data were collected at 100 Hz using a VICON system and ground reaction forces were measured with a sampling rate of 300 Hz using two AMTI force plates. We were used to calculating 3D moments and reaction forces at the

L5/S1 joint using inverse dynamics. The angle between the C7 and T12 markers defined the trunk posture and the angle between the T12 and L5 marker defined the lumbar spine posture (29). As well as to reduce the calculation error and More similarities kinematic patterns with reference was used the SARA compensatory algorithms to obtain the axis of rotation of the hinge-joints and SCORE algorithm to obtain the center of rotation of the Ball & Socket joints (30).

## Data analysis

The pattern of angles variation in each lower extremity and lumbosacral joint (L5/S1) were normalized with respect to time (from 0 to 100%) in each position during lifting. The angle between crossing line from L5/S1 joint to T1 and crossing line from hip joint to L5/S1 joint was measured. The angle of each lower limb joint with crossing a line through two connected limbs in each joint was determined. Anthropometric data, ground reaction forces, and kinematic data were included in the three-dimensional linked-segment model. In this study, to calculate extensor moment and joint power in each lifting, inverse dynamics were used. The starting point of each lift "zero points" was determined by a micro-switch located between the box and the ground. By starting lifting, all synchronized equipment started to record. Each lifting started from zero points and ended to trunk full extension. The trends of the extensor moment of the lower extremity and lumbosacral joint (L5/S1), the power of ankle, knee, and hip joint in LF, FF, and SF positions were obtained. Moreover, the mean, maximum value, and time to peak (TTP) for each dependent variable in each position were calculated. All calculations and data analysis were implemented in three-dimensional coordination using VICON Nexus 2.5 and MATLAB-R2010 software.

To obtain mean and maximum muscular activity in each trunk muscle, the root mean square of each muscle was normalized concerning MVC. This can be used for comparing and evaluating the kinetic calculation in each position. Finally, to investigate the time to reach to maximum muscular activity of each muscle during lifting, EMG normalized data with respect to MCV were normalized for time (from 0 to 100%). All EMG data processing was implemented by MATLAB-R2010 software.

## Statistical analysis

One-way repeated measure ANOVA was used to investigate the effect of initial body position in different knee flexion angles during lifting with constant velocity on mean, maximum, and time to peak of extensor moment, power of lower limb, and lumbosacral joint extensor moment (L5/S1). To determine the significant changes of dependent variables (mean, peak, and TTP) related to trunk muscle activity during lifting in different knee positions, one-way repeated measured ANOVA was used. Subsequently, once this analysis resulted in a significant effect of particular dependent variables, the Bonferroni post-hoc test was used to examine the differences between the three levels of knee postures. To investigate the significant relationship between trunk muscle activity and lumbar extensor moment and the correlation of knee extensor moment and lumbar extensor moment, Pearson correlation Test was used. For all tests, a significance level of $p<0.05$ was used. The means obtained were significantly different, with p-values less than 0.05. The reliability was examined using intra-class correlation coefficients (ICCs). Statistical analyses were conducted using the SPSS statistical package (version 16), and ICCs were interpreted according to Domholdt (2005) (31).

# RESULTS

The mean, peak, and time to peak (TTP) of extensor moment of lower limb joints, lumbar extensor moment (L5/S1), and muscular activity of trunk muscle for each lifting were determined. The validity of all dependent kinetic variables ($0.80<ICC<0.88$) was higher than the dependent variables of muscular activity in each position ($0.72<ICC<0.76$). Fig.4 (a) shows the trend of changing of the knee, hip, and lumbar extensor moment curve during lifting with constant velocity in LF, SF, and FF positions. Except for the knee joint, all other joints had the same pattern in FF, SF, and LF positions. The ankle extensor moment had a similar curve in SF and FF position so that they hit the peak at approximately 29% of total time. Then, they showed a decreasing trend from peak to end of lifting time. This decrease was approximately continued to half of the moment peak and in LF posture it continued to zero point (Fig. 4). The results of one-way repeated measured ANOVA showed no significant change at the mean, peak, and TTP in different knee postures. But, there was a significant difference at peak of ankle extensor moment in all postures (P=0.001) whereas by increasing knee flexion the peak value of the power of ankle joint would be increased at 60%, 83%, and 45%, respectively.

The maximum variation of extensor moment was obtained in the knee joint whereas, by increasing the knee flexion angle the mean and peak value of knee extensor moment increased significantly. The results only showed a significantly increasing peak extensor moment of the knee. The peak value of knee extensor moment in FF position showed significant increasing respect to SF and LF, 40% and 45%, respectively. The test results showed that the peak value of the knee extensor moment had significant differences in all three postures (P<0.009). Despite SF and FF position, lift starting in LF posture accompanied by an increasing flexor resistive moment and continued to 42% of the period of lifting. But, it started with decreasing extensor moment in SF and FF postures. In SF and FF positions, peak extensor moment of knee joint occurred at 17% and 8% of lifting period whereas it occurred about 32% of lifting period in LF posture. In other words, by increasing knee flexion angle, the time to peak of knee extensor moment was significantly reduced (Fig. 3). The results of post-hoc test showed only the significance of TTP decreasing in the FF position with respect to the LF position (P=0.035).

Fig. 4 shows the extensor moment curve of the knee joint in LF, SF, and FF postures with constant velocity during lifting. In SF and FF positions, there is a similar pattern with different amplitudes. The maximum extensor power of the knee joint, like knee extensor moment, occurred at beginning of lifting in the FF position. By increasing the knee flexion angle, the peak power of the knee would significantly increase (Fig. 1) whereas the peak extensor power of the knee was increased by 235% in LF position with respect to the FF position. The most changing of the peak of knee extensor power occurred at 47%, 58%, and 15% of lifting time in LF, SF, and FF positions, respectively. The results of post-hoc test of TTP were significant in all three postures (p=0.003).

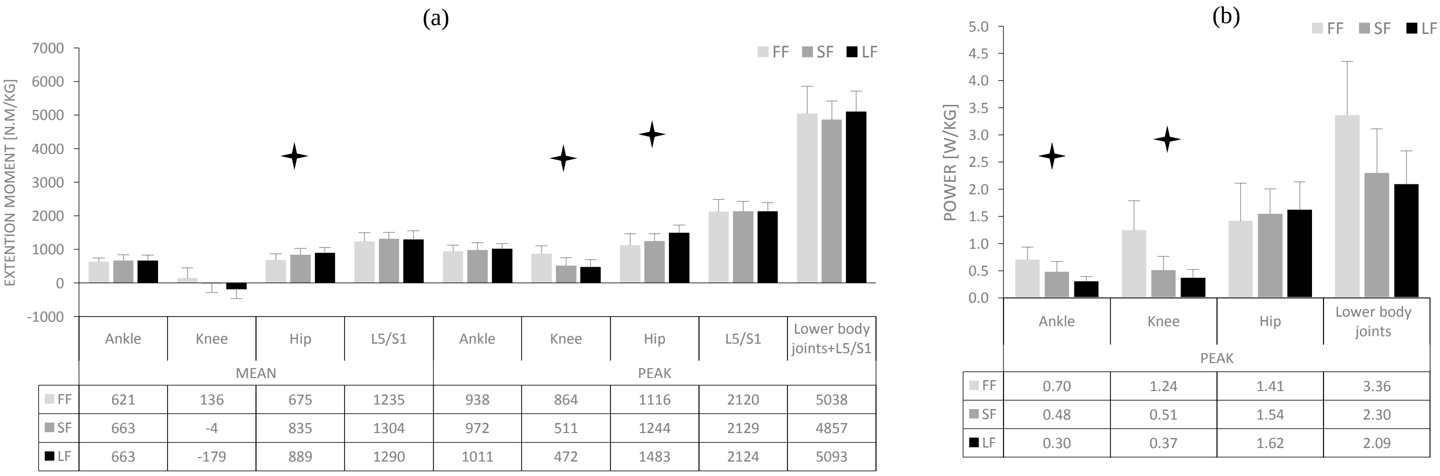
The pattern of hip extensor moment was the same in three postures (Fig. 4) and the peak of hip extensor moment occurred at 27% of lifting time in all knee positions (Fig. 3). But, the results showed a significant decrease of mean and peak of hip extensor moment by increasing knee flexion angle. The results of post-hoc revealed that there is a significant change at mean and the peak of hip extensor moment in LF and FF postures (p=0.012, p=0.049) whereas by increasing knee flexion they decreased. Moreover, there was no significant variation in TTP of extensor moment and peak of hip extensor moment during lifting with constant velocity in three positions (Table 1).

The pattern of lumbar extensor moment (L5/S1) was almost identical in three positions. The peak of the lumbar extensor moment occurred at 14% of lifting time and ultimately decreased to zero (Fig. 4). Different knee postures had no significant effect on mean, peak, and TTP of lumbar extensor moment (Table 1). Despite the peak of the sum of the lower limb and lumbar extensor moment in SF posture were less than LF and FF postures, 4.63% and 3.59%, respectively, but it was not significant. However, the peak of the sum of lower limb extensor power showed a significant increase by increasing the knee flexion angle. The results of post-hoc revealed that there is a significant difference between SF-FF and LF-FF positions by increasing knee flexion angle, 0.8 (46%), and 76 (60%), respectively (Fig. 1(b)).

To match and investigate the muscular activity of trunk muscle and kinetic values of the lumbar joint (moment and power) during lifting with constant velocity in different knee flexion angles, the peak and mean value of muscular activity were normalized with respect to MCV. Also, the time to peak was measured. The results of one-way repeated measured ANOVA showed that there was no significant effect of increasing knee flexion angle on the mean and peak of lumbar muscles (multifidus and iliocostalis) and external oblique, except for rectus abdominis (Table 1). In contrast, there was a significant effect of knee flexion angle on mean and peak of rectus abdominus during lifting (p=0.001, p=0.003). The results of post-hoc showed that there was only a significant difference at the mean value of muscular activity in SF and FF positions (p=0.002) whereas increasing knee flexion angle caused 48.81% growth in rectus abdominis muscular activity (Fig. 2). Moreover, the peak value of rectus abdominis muscular activity showed a significant difference in FF-SF and FF-LF postures (p=0.001, p=0.001), whereas increasing knee flexion angle from SF to FF and LF to FF caused 105% and 82% growth in rectus abdominis muscular activity (Fig. 2). Besides, the results reported the TTP of rectus abdominis (p=0.034) and iliocostalis lumborum muscular activity (p=0.001) in different knee flexions. The results of post-hoc test showed that the TTP value of rectus abdominis only received significant changes in FF and SF position (p=0.035) and SF to FF position, the TTP value increased by 50%. The TTP of rectus abdominis in SF and FF postures occurred at 6% and 12% of lifting time, respectively (Fig. 3). Similarly, the TTP of rectus abdominis in FF-LF and SF-LF postures showed significant difference whereas the TTP value increased by 152% and 194% in FF to LF and SF to LF positions, respectively. By increasing the knee flexion angle, the trend of TTP of multifidus and iliocostalis muscles with lumbar extensor moment was almost similar whereas by increasing knee flexion angle, the time required to reach peak decreases (Fig. 3).

To investigate the relationship between dependent variables in each posture, the Pearson correlation test was used. The results showed there was no significant correlation between the muscular activity of multifidus and iliocostalic lumborum and the mean and peak of lumbar extensor moment during lifting with constant velocity in knee postures. While there was a significant relationship between the mean of muscular activity of external oblique and TTP of lumbar extensor moment in SF (p=0.048, p-c=0.670) and LF (p=0.033, p-c=0.673) postures and mean of muscular activity of iliocostalic lumboum and TTP of lumbar extensor moment, there was a significant relationship between the mean of rectus abdominis muscular activity and mean (p=0.005, p-c=-0.871) and peak (p=0.006, p-c=0.864) of lumbar extensor moment in SF posture.

According to results, it was observed a meaningful relationship between peak of ankle and lumbar extensor moment in FF (p=0.020, p-c=0.717) and SF (p=0.038, p-c=-0.6940 postures as well as the mean of knee extensor moment and lumbar extensor moment in SF (p=0.009, p-c=-0.807) and LF (p=0.041, p-c=0.653) postures. Moreover, it was reported a significant relationship between the peak of hip and lumbar extensor moment in FF (p=0.008, p-c=0.780) position. In FF (p=0.001, p-c=0.938) and LF (p=0.021, p-c=0.938) positions, there was a meaningful correlation between the peak of lumbar and total lumbar extensor moment.




| | Ankle (MEAN) | Knee (MEAN) | Hip (MEAN) | L5/S1 (MEAN) | Ankle (PEAK) | Knee (PEAK) | Hip (PEAK) | L5/S1 (PEAK) | Lower body joints+L5/S1 (PEAK) |
|---|---|---|---|---|---|---|---|---|---|
| FF | 621 | 136 | 675 | 1235 | 938 | 864 | 1116 | 2120 | 5038 |
| SF | 663 | -4 | 835 | 1304 | 972 | 511 | 1244 | 2129 | 4857 |
| LF | 663 | -179 | 889 | 1290 | 1011 | 472 | 1483 | 2124 | 5093 |

| | Ankle (PEAK) | Knee (PEAK) | Hip (PEAK) | Lower body joints (PEAK) |
|---|---|---|---|---|
| FF | 0.70 | 1.24 | 1.41 | 3.36 |
| SF | 0.48 | 0.51 | 1.54 | 2.30 |
| LF | 0.30 | 0.37 | 1.62 | 2.09 |

**Fig. 1:** (a) The mean (MEAN) and peak value of lower extremity extensor moments, lumbar extensor moment (L5/S1) and total moment (sum of lumbar and other lower extremity extensor moment), (b) The peak of extensor power of lower limb joints and total power (sum of all lower extremity extensor powers) during lifting with constant velocity in FF, LF and SF positions.

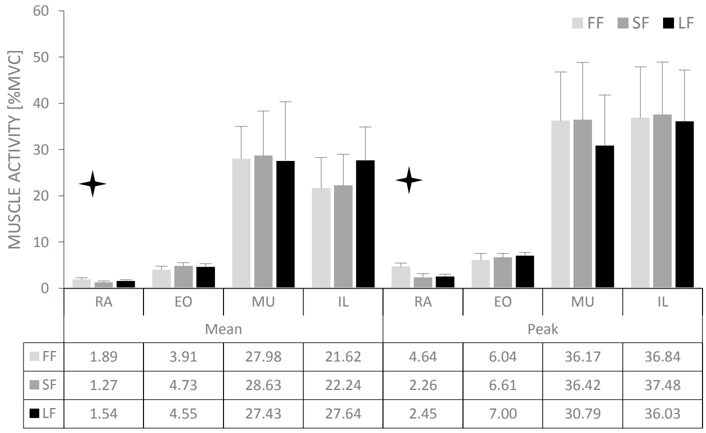



**Fig. 2:** The normalized mean and peak value of muscular activity of trunk muscles respect to MCV during lifting with constant velocity in different knee positions (FF, LF and SF); RA: rectus abdominis, EO: external oblique, IL: iliocostalis lumborum, MU: multifidus.

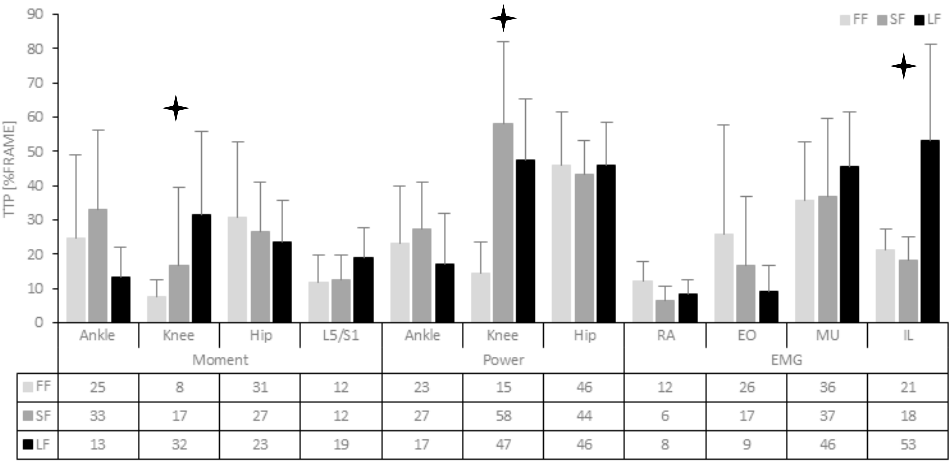
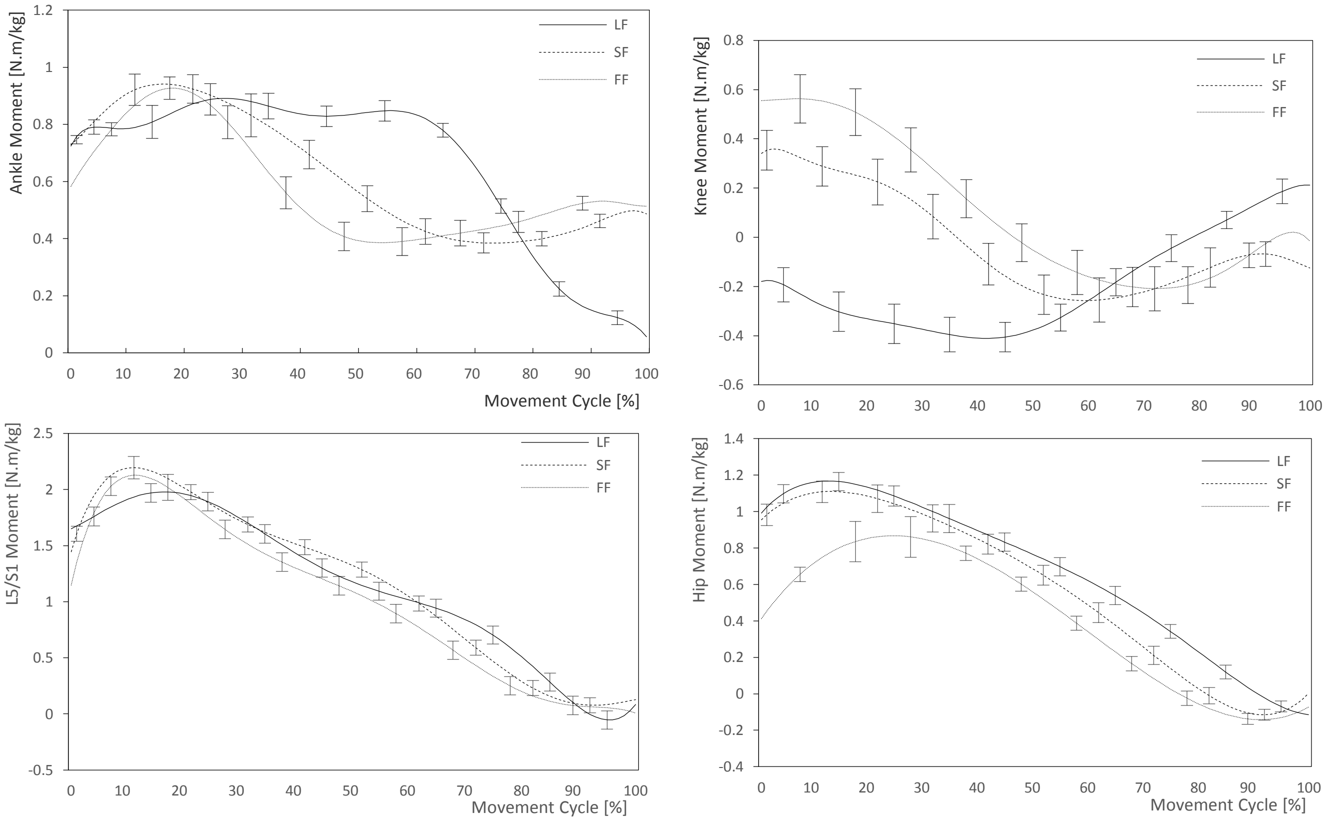


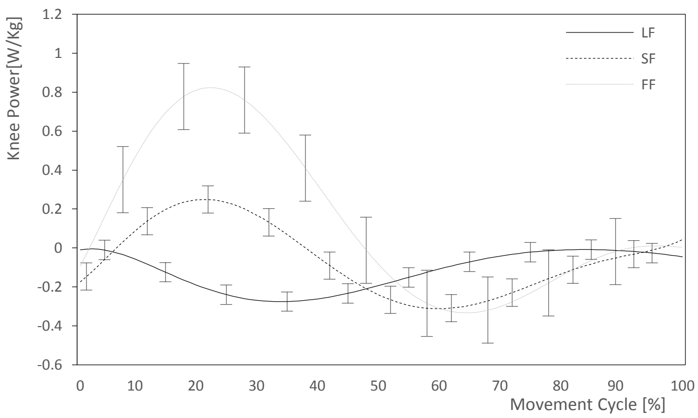

**Fig. 3:** The time required to reach peak (TTP) of lower limb extensor power and moment, lumbar moment (L5/S1) and peak of muscular activity of trunk muscle during lifting with constant velocity in different knee positions (LF, SF and FF).

Table 1. MEAN: Mean normalized, PEAK: maximum normalized, TTP: Time to peak, Lower body joints: lower limb extensor power and moment, lumbar moment (L5/S1), MU: multifidus, IL: iliocostalis lumborium, RA: rectus abdominis, EO: external oblique. Significant differences are shown in bold. The significance level ¼ 0.05 are shown in italics.

| | **Extension Moment** | | | | | **Extension Power** | | | | **Muscle Activity** | | | | **Net L5/S1 moment** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Ankle | Knee | Hip | $L_5/S_1$ | Lower body joints+$L_5/S_1$ | Ankle | Knee | Hip | Lower body joints | RA | EO | MU | IL | |
| **MEAN** | P=0.793 | P=0.071 | **P=0.044** | P=0.809 | - | - | - | - | - | **P=0.003** | P=0.065 | P=0.978 | P=0.105 | - |
| | F=0.233 | F=2.941 | F=3.533 | F=0.213 | - | - | - | - | - | F=7.593 | F=3.029 | F=0.022 | F=1.989 | - |
| **PEAK** | P=0.698 | **P=0.002** | **P=0.023** | P=0.998 | P=0.737 | **P=0.000** | **P=0.000** | P=0.746 | **P=0.004** | **P=0.000** | P=0.194 | P=0.605 | P=0.959 | P=0.121 |
| | F=0.365 | F=8.228 | F=4.355 | F=0.002 | F=0.308 | F=12.212 | F=15.078 | F=0.297 | F=6.817 | F=26.047 | F=1.751 | F=0.516 | F=0.042 | F=0.590 |
| **TTP** | P=0.143 | **P=0.042** | P=0.615 | P=0.117 | - | P=0.399 | **P=0.000** | P=0.909 | - | **P=0.034** | P=0.262 | P=0.588 | **P=0.001** | - |
| | F=2.121 | F=3.621 | F=0.495 | F=2.335 | - | F=0.955 | F=12.700 | F=0.096 | - | F=3.831 | F=1.408 | F=0.546 | F=9.736 | - |




(a)




(b)

**Fig. 4:** (a) curve patterns of ankle, hip and lumbar extensor moment, (b) curve patterns of ankle extensor power during lifting with constant velocity in different knee position (LF, SF and FF).

## DISCUSSION

### Extensor Moment and power of lower extremity joints and lumbosacral joint

Many studies have evaluated the effect of different dynamic factors on different postures of the body during loads lifting on the lumbar and other lower limb joints. They showed that the inertia factor could play an important role in the process of changing the pattern of the curve and the range of kinetic parameters (force, moment, and mechanical work) of joints involved in a lifting motion. The results showed a significant and high correlation between the maximum value of the box linear acceleration and the maximum value of the

extensor moment of the lumbar spine in the dynamic lifts, as well as they, found that the peak of the time in both parameters was approximately the same (33, 15, 6). Consequently, most researchers have often suggested lifting with no acceleration or controlled lifting based on NIOSH guidelines, whereas biomechanical evaluations of the joints involved in the lifting motion while controlling the inertia factor (lifting box by constant velocity) in different knee positions have received less attention.
It has been hypothesized by the researcher that despite the inertia effect inhibition and the equal horizontal and vertical load-to-body distances, different knee flexion angle could have effects on the pattern of moment curve of lumbar extensor and other lower limb joints as well as trunk muscle activity level during lifting, and whether the body's initial posture during lifting could be an effective and significant factor on lumbar extensor moment if inertia restrained and equal horizontal and vertical load distances from the body were considered, and whether different extensor moment values (10, 34, 35) according to different lifting techniques could be reasonable despite the researcher's assumptions including inertia effect inhibition, equal horizontal and vertical load distances from the body and different angles of knee flexion. For these reasons, the current study was of particular importance. The results showed that curve pattern and peak values of extensor moment of lumbar, ankle, knee, and hip joints in all three SF, FF, and LF postures, by considering assumptions in lifting, were totally consistent with previous studies which conducted in the same conditions so that curve pattern of extensor moments of ankle and knee were almost identical with Squat technique in SF and FF postures and with Stoop technique in LF posture (32, 35, 39).
Like previous studies, knee joint positioning has been considered different in the common squat technique in different articles and there are many differences which may have a significant effect on the extensor moments of the involved joints, especially the lumbar spine, and on the other hand, different acceleration and lifting velocities can lead to different results. For this reason, in this study, more precise primary positional effects of the body during lifting through controlling the inertia effect on extensor moments of involved joints were investigated by removing the inertia effect and maintaining a constant horizontal and vertical distance in all lifts and by taking into account the specific range of motion for knee joint. Under the assumed conditions (horizontal distance: the nearest distance to legs and load weight: 15 kg with no inertia control), Kingma (35) reported a significantly increased extensor moment in Stoop versus Squat, while Faber et al., (2008) (34) showed a significant increased lumbar extensor moment in Squat versus Stoop in almost the same condition (horizontal distance: not specified, load weight: 20 Kg, not controlling of inertia). Despite the effect of inertia and horizontal distance between load and leg, knee positioning might also have a significant effect on differences among Faber, Kingma, and others (10), whereas in Kingma's study the Squat technique was considered as FF posture and in Faber's study the Squat technique was considered as SF posture of the current study. The comparison between the Squat and Stoop technique could be slightly considered as the comparison between LF and FF postures in Kingma study as well as SF and LF postures in Faber's study. However, the results of the current study are totally consistent with Faber's and Kingma's ones. In contrast, in this study, by removing the effect of inertia and horizontal and vertical distance between load and legs and considering different flexion of the knee, the mean and peak of lumbar extensor moment were not significantly affected in all knee postures. Therefore, the differences between lumbar extensor moments in various lifting techniques could be interpreted as a different range of knee flexion but logically this difference could not be caused by knee positioning during load lifting. Finally, it could be noted that there is no significant effect on lumbar extensor moment in different knee positions during non-accelerated lifting (constant velocity) as well as accelerated lifting. But, these different values of lumbar extensor moment could be interpreted by the muscular power of involved joints and energy transferring between lower limb joints in different knee positions.
Given that the box is lifted at a constant velocity and the impact of external inertia forces on the function of the joints are neglected, we know that in different knee positions, the position of trunk mass center varies with respect to the mass center of the knee joint that could have an effect on loading of the lumbar joint. Moreover, in different postures of the knee, the power of knee extensor muscles and other involving joints varies caused

by resistive forces against the movement as a result of changing of cross-section and length of muscle as well as moment arms. In the other words, based on the initial posture of the body in different positions of lifting, various groups of muscles play a role to produce resistive moment. For this purpose, the simultaneous calculation of lower extremity joint strength can help in better elucidating the process of changes and factors affecting extensor moment during load lifting with constant velocity. In all knee postures, the distance from the beginning to the end of the lift was almost the same (the box is carried from ground level to chest level in a constant path). But in each posture the range of motion of the lower limb joints is different and this affects the angular momentum of each joint. This difference in the range of motion of lower limb joints in different postures showed a significant changing of kinetic of lower limb joints whereas by increasing the knee flexion angle the extensor power of ankle and knee joints was decreased but it had no significant effect on dependent values of research including mean, peak and TTP values of extensor moment (Table 1) which is consistent with results of M. Gagnon and G. Gagnon (1992) (21) about produced work of knee and ankle joints. They reported that work produced by ankle and knee joints are totally small and they provide 15% and 20% of the required energy for lifting, respectively. The total energy transition from these joints to other upper joints is ignorable, about 2%, and hip and lumbar joints have more portion of external loading in dynamic lifting. According to significant changing of mean and peak extensor moment of hip between LF and FF postures during load lifting with constant velocity, there is no significant effect on hip joint power. This procedure could be interpreted by the results of the study of M. P. De Looze (12) about the similarity of gluteus maximus muscle activity in both methods of leg and back lifting. In addition, except for FF posture, there is no correlation between the peak of lumbar extensor moment and hip joint in knee various positions for load lifting by constant velocity.
The results showed that the mean and peak value of the lumbar extensor moment was seen in the SF position and the least ones were seen in the FF position, but the difference was not meaningful. The most difference of the mean and peak of lumbar extensor moment was observed between SF and FF positions, 4.49% and 4.60%, respectively. These results are consistent with the Dolan and De Looze ones which were reported 5% based on Norman and McGill models (1994) (38). According to ignoring the effect of inertia forces and other assumptions (horizontal and vertical distance from legs), this could be caused by increasing the distance of involved limb mass center especially upper limbs from body mass center in different knee postures. That is because, in the SF position, the distance of the upper limb mass center from the body mass center is more than two other positions. But, the peak value of the extensor moment of total joints had the least value in SF posture with respect to two other positions. The reason could be interpreted based on muscular power of total lower limb joints and the consumed energy in different body positions, whereas the results showed significant decreasing of the total power of lower limb joint (ankle, knee, and hip) by increasing knee extension flexion angle in different lifts with constant velocity. On the other hand, Kumar (1984) (37) reported that the knee flex technique in accelerated lifting was principally more frustrating than the straight leg technique and freestyle technique had accompanied the least fatigue. In the other words, the amount of metabolic energy for knee flex and freestyle showed the most energy consumption and the least for the straight leg (37). So, because the less total joint power (sum of the power of ankle, knee, and hip joints) produced by the muscular activity of lower limb muscles in LF posture, it is expected to obtain less total extensor moment than two other postures.

**Muscular activity of trunk muscles**

As the results showed, there was no meaningful relationship between lumbar extensor moment and different knee postures during lifting with constant velocity. Also, different knee flexion angles had no significant effect on lumbar muscle activity (MU-IL). Moreover, there was no meaningful relationship between the mean and peak of IL and MU muscular activity level. Although, in the literature, significant effectiveness of lumbar muscles (MU-IL) with respect to box weight and lifting velocity has been severally notified in an accelerated movement (20) and many studies have focused on the effectiveness of different lifting techniques on trunk muscle activity level in accelerated movement (13), but they reported different results due to differences in the lower limb and lumbar joint angles in each lifting common techniques, lifting procedure and other controlling variables. The behavior of trunk muscles at different knee flexion angles in the non-accelerated movement has been paid little attention to the Squat technique. Despite differences in curve pattern of IL and MU muscle activity in each knee posture, there was no meaningful effect of the mean and peak of IL and MU muscle activity with respect to different knee flexion postures during lifting with constant velocity is assumed conditions (fixed box weight, fixed horizontal and vertical distance and ignoring inertia effect during motion). Vahdat et al. (2016) showed that the curve pattern of both IL and MU muscles followed an accelerated curve in accelerated movement with a fixed position (Squat) and the time to peak of acceleration was almost identical with muscular peak. But, in contrast to Vahdat's results, the mean value of MU muscular activity was greater than IL muscle with constant velocity (zero acceleration) in SF and FF postures, showing the uniform activity of MU muscle during lifting. However, this trend was changed by increasing the knee extension angle and the mean value of MU muscle activity decreased with respect to IL muscle activity. The peak value of IL muscle activity was greater than MU ones in all three knee postures which was consistent with previous studies in accelerated lifting. Because of the greater distance of different lower limb segments' mass center especially trunk from body mass center in SF posture with respect to FF and LF postures, both MU and IL muscle had the greatest muscular activity. Given that the greatest lumbar extension moment has also occurred in this posture, the greater activity of lower limb extensor muscles is to be expected in SF posture. The mean of TTP for IL muscle occurred at 21% and 53% of lifting period in FF and LF postures, respectively, whereas a significant delay to TTP was seen in IL muscle activity which was consistent with Hwang's study showed quantitatively the muscular activity curve of erector spinae of Stoop and Squat techniques. In LF posture, the upper limb is positioned in high flexion and lumbar muscles are in the extension position. This caused to decrease in the IL muscular activity level which resulted in more active force by increasing trunk extension during mid-lifting.

The EO muscle showed irregular behavior in each knee postures. This could be due to the effect of intra-abdominal pressure (IAP) on abdominal muscle activity (38, 40). Despite EO muscle, there was a significant effect of different knee positions on RA muscle activity level during lifting with constant velocity (Table 1). As shown in Fig. 2. The most value of mean and peak of RA muscle activity was similar to the full-squat technique in FF posture and the least ones occurred in SF posture in the semi-squat technique. According to increasing the knee extension angle from FF to SF position, it is observed a significant decrease of mean and peak value of RA muscular activity. In addition, the peak value of RA muscular activity from FF to LF postures showed a meaningful decrease. Since horizontal load distance and upper limb mass center from the rotation center of the body (L5/S1) during lifting in FF position is positioned in unstable posture rather than LF and FF positions (363636), abdominal muscles (especially RA) play an important role to maintain stability. Moreover, in FF position in which the heels inevitably rise from the ground (Burgess, Limerick, 2003), the body is positioned in unstable condition and as a result, abdominal muscles are involved to maintain stability which is consistent with the research literature. As we closed to the LF position, these distances decrease and the stability increases in the sagittal plane, and subsequently RA muscular activity decreases significantly from FF to LF postures. The TTP value of RA muscle occurred at beginning of lifting in all three postures but, the difference was only significant between FF and SF postures. Moreover, there was only a significant reverse correlation between the mean ($p=0.005$, $p\text{-}c=-0.871$) and peak ($p=0.006$, $p\text{-}c=-0.864$) of lumbar extensor moment in SF posture.

## CONCLUSION

According to results regarding the role of abdominal muscles especially RA muscle during lifting with a constant velocity while controlling inertia in different positions and its correlation with lumbar extensor moment, it could be noted that RA muscle plays a role in both the stability of the body and in controlling the amount of required lumbar extensor moment, except that this proportion varies with the position. Based on the current study and full-Squat technique in several studies, the RA muscle has the greatest muscular activity level to maintain body stability while instability in FF posture respect to SF and LF postures meanwhile there was no significant relationship with lumbar extensor moment. But, this correlation is clearly seen in postures with high body stability while a full extension of the elbow joint (the controlling role of muscle to produce lumbar extensor moment is seen). Finally, the RA muscle activity could be considered related to knee positioning during lifting with constant velocity.

## PRACTICAL APPLICATION

The findings of the present study can be effectively applied in ergonomics relating to symmetric squat lifting. Based on these findings, the activation of lower trunk muscles during squat lifting is affected by initial body postures on different knee flexion while controlling the effect of inertia force (constant velocity) of the system. Clearly, increases in knee flexion while controlling the effect of inertia was no significant effect on mean and peak values of lumbar extensor moment. Moreover, No significant changes in dorsal trunk muscle activity (MU, IL) were observed. These findings also demonstrate the compensatory actions of joints and muscles for the purpose of controlling movement. The results reported herein may inform and help subjects to achieve proper motion in order to prevent jerky movements and excessive forces on the lower joints during risky squat lifting tasks in industrial and labor applications. Based on the mean and TTP values of the lower trunk musculature during squat lifting on different knee flexion, clinicians may be able to evaluate subjects' muscular functions. Understanding the natural EMG Quantity values, lower extremity and lumbar extensor moment, and power of joints and innervation behavior of the lower trunk musculature during squat lifting performance, clinicians may be able to diagnose abnormal and jerky muscular contractions that may have to be present during risky industrial or labor lifting tasks. Also, with access to curve pattern of the lower extremity and lumbar extensor moment and TTP data on different knee flexion angles, while controlling the effect of inertia (constant velocity) in Squat technique, undesirable peaks can be detected within extensor moment curve patterns and specified as indications of jerky movements.

## LIMITATIONS

The limited range of BMI indices, sexuality, age and box masses and lift speeds were limitations of this study. All participants in this study were young males and their BMI indices were limited to a specific range. In addition, the box masses and lift speeds employed fell within limited ranges. The results might have differed if the participants were not young males or if their BMI indices were limited to a different range. Furthermore, the findings of the study may be valid only for the determined box masses and lift velocity. The experiments in this study were conducted utilizing only symmetric lifting conditions. Other lifting conditions, such as asymmetric lifting, may cause different innervation behavior in the muscles investigated.

## ETHICAL CONSIDERATIONS

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## CONFLICT OF INTEREST

We confirm that no potential conflict of interest has been reported with regard to this article.

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## REFERENCES


1. Kuiper JI, Burdorf A, Verbeek JH, Frings-Dresen MH, van der Beek AJ, Viikari-Juntura ER. Epidemiologic evidence on manual materials handling as a risk factor for back disorders: a systematic review. International Journal of Industrial Ergonomics. 1999; 24(4):389-404. [DOI:10.1080/00140138408963506]
2. Potvin JR, McGill SM, Norman RW. Trunk muscle and lumbar ligament contributions to dynamic lifts with varying degrees of trunk flexion. Spine. 1991; 16(9):1099-107. [DOI:10.1097/00007632-199109000-00015]
3. Waters TR, Putz-Anderson V, Garg A, Fine LJ. Revised NIOSH equation for the design and evaluation of manual lifting tasks. Ergonomics. 1993; 36(7):749-76. [DOI:10.1080/00140139308967940]
4. McGill SM, Norman RW. Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine. 1986 ;11(7):666-78. [DOI:10.1016/0268-0033(87)90144-6]
5. Davis KG, Marras WS. Assessment of the relationship between box weight and trunk kinematics: does a reduction in box weight necessarily correspond to a decrease in spinal loading?. Human factors. 2000; 42(2):195-208. [DOI:10.1518/001872000779656499]
6. Kingma I, Baten CT, Dolan P, Toussaint HM, van Dieën JH, de Looze MP, Adams MA. Lumbar loading during lifting: a comparative study of three measurement techniques. Journal of Electromyography and Kinesiology. 2001; 11(5):337-45. [DOI:10.1016/S1050-6411(01)00011-6]
7. De Looze MD, Kingma I, Thunnissen W, Van Wijk MJ, Toussaint HM. The evaluation of a practical biomechanical model estimating lumbar moments in occupational activities. Ergonomics. 1994; 37(9):1495-502. [DOI:10.1080/00140139408964929]
8. Dolan P, Adams MA. The relationship between EMG activity and extensor moment generation in the erector spinae muscles during bending and lifting activities. Journal of biomechanics. 1993; 26(4-5):513-22. [DOI:10.1016/0021-9290(93)90013-5]
9. Dolan P, Adams MA. The relationship between EMG activity and extensor moment generation in the erector spinae muscles during bending and lifting activities. Journal of biomechanics. 1993; 26(4-5):513-22. [DOI:10.1080/00140130210123507]
10. van Dieën JH, Hoozemans MJ, Toussaint HM. Stoop or squat: a review of biomechanical studies on lifting technique. Clinical Biomechanics. 1999; 14(10):685-96. [DOI:10.1016/S0268-0033(99)00031-5]
11. Kingma I, Bosch T, Bruins L, Van Dieën JH. Foot positioning instruction, initial vertical load position and lifting technique: effects on low back loading. Ergonomics. 2004; 47(13):1365-85. [DOI:10.1080/00140130410001714742]
12. De Looze MP, Toussaint HM, Van Dieen JH, Kemper HC. Joint moments and muscle activity in the lower extremities and lower back in lifting and lowering tasks. Journal of biomechanics. 1993; 26(9):1067-76. [DOI:10.1016/S0021-9290(05)80006-5]
13. Hwang S, Kim Y, Kim Y. Lower extremity joint kinetics and lumbar curvature during squat and stoop lifting. BMC musculoskeletal disorders. 2009; 10(1):1-0. [DOI:10.1186/1471-2474-10-15]
14. Dolan P, Earley M, Adams MA. Bending and compressive stresses acting on the lumbar spine during lifting activities. Journal of biomechanics. 1994; 27(10):1237-48. [DOI:10.1016/0021-9290(94)90277-1]
15. Marras WS, Mirka GA. A comprehensive evaluation of trunk response to asymmetric trunk motion. Spine. 1992; 17(3):318-26. [DOI:10.1097/00007632-199203000-00013]
16. Marras WS, Granata KP. Spine loading during trunk lateral bending motions. Journal of biomechanics. 1997; 30(7):697-703. [DOI:10.1016/S0021-9290(97)00010-9]

17. Bush-Joseph C, Schipplein O, Andersson GB, Andriacchi TP. Influence of dynamic factors on the lumbar spine moment in lifting. Ergonomics. 1988; 31(2):211-6. [DOI: 10.1080/00140138808966662]
18. Kingma I, Bosch T, Bruins L, Van Dieën JH. Foot positioning instruction, initial vertical load position and lifting technique: effects on low back loading. Ergonomics. 2004; 47(13):1365-85. [DOI:10.1080/00140130410001714742]
19. Seroussi RE, Pope MH. The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. Journal of biomechanics. 1987; 20(2):135-46.. [DOI: 10.1016/0021-9290(87)90305-8]
20. Vahdat I, Rostami M, Ghomsheh FT, Khorramymehr S, Tanbakoosaz A. The effects of task execution variables on the musculature activation strategy of the lower trunk during squat lifting. International Journal of Industrial Ergonomics. 2016; 55:77-85. [DOI: 10.1016/j.ergon.2016.07.007]
21. Gagnon M, Smyth G. Biomechanical exploration on dynamic modes of lifting. Ergonomics. 1992; 35(3):329-45. [DOI: 10.1080/00140139208967817]
22. Straker L. Evidence to support using squat, semi-squat and stoop techniques to lift low-lying objects. International Journal of Industrial Ergonomics. 2003; 31(3):149-60. [DOI: 10.1016/S0169-8141(02)00191-9]
23. Kingma I, Bosch T, Bruins L, Van Dieën JH. Foot positioning instruction, initial vertical load position and lifting technique: effects on low back loading. Ergonomics. 2004; 47(13):1365-85. [DOI: 10.1080/00140130410001714742]
24. Burgess-Limerick R. Squat, stoop, or something in between?. International Journal of Industrial Ergonomics. 2003; 31(3):143-8. [DOI: 10.1016/S0169-8141(02)00190-7]
25. Ekstorm M. Smärtbedömning på akutmottagningen [Internet] [Dissertation]. 2008. Available from: http://urn.kb.se/resolve?urn=urn:nbn:se:kau:diva-58179
26. Ng JK, Parnianpour M, Richardson CA, Kippers V. Functional roles of abdominal and back muscles during isometric axial rotation of the trunk. Journal of Orthopaedic Research. 2001; 19(3):463-71. [DOI: 10.1016/S0736-0266(00)90027-5]
27. Ng JK, Kippers V, Parnianpour M, Richardson CA. EMG activity normalization for trunk muscles in subjects with and without back pain. Medicine and science in sports and exercise. 2002; 34(7):1082-6. [DOI: 10.1097/00005768-200207000-00005]
28. Ng JK, Richardson CA, Parnianpour M, Kippers V. EMG activity of trunk muscles and torque output during isometric axial rotation exertion: a comparison between back pain patients and matched controls. Journal of Orthopaedic Research. 2002; 20(1):112-21. [DOI:10.1016/S0736-0266(01)00067-5]
29. Tabakin D. A comparison of 3D gait models based on the Helen Hayes Hospital marker set (Master's thesis, University of Cape Town). [http://hdl.handle.net/11427/3206]
30. Taylor WR, Kornaropoulos EI, Duda GN, Kratzenstein S, Ehrig RM, Arampatzis A, Heller MO. Repeatability and reproducibility of OSSCA, a functional approach for assessing the kinematics of the lower limb. Gait & posture. 2010; 32(2):231-6. [DOI:10.1016/j.gaitpost.2010.05.005]
31. Domholdt E. Rehabilitation research: principles and applications. 3rd ed. St. Louis, MO: Elsevier Saunders; 2005. p. 17–28.
32. Faber GS, Kingma I, Bakker AJ, Van Dieën JH. Low-back loading in lifting two loads beside the body compared to lifting one load in front of the body. Journal of biomechanics. 2009; 42(1):35-41. [DOI:10.1016/j.jbiomech.2008.10.013]
33. Vahdat, I., Tabatabai Ghomsheh, F., Khorramymehr, S. and Tanbakoosaz, A., 2017. Effects of external loading on lumbar extension moment during squat lifting. [DOI:10.13075/ijomeh.1896.00896]

34. Faber GS, Kingma I, van Dieën JH. Effect of initial horizontal object position on peak L5/S1 moments in manual lifting is dependent on task type and familiarity with alternative lifting strategies. Ergonomics. 2011; 54(1):72-81. [DOI:10.1080/00140139.2010.535019]
35. Coenen P, Kingma I, Boot CR, Bongers PM, van Dieën JH. Detailed assessment of low-back loads may not be worth the effort: A comparison of two methods for exposure-outcome assessment of low-back pain. Applied ergonomics. 2015; 51:322-30. [DOI:10.1016/j.apergo.2015.06.005]
36. Solomonow M. Neuromuscular manifestations of viscoelastic tissue degradation following high and low risk repetitive lumbar flexion. Journal of Electromyography and Kinesiology. 2012; 22(2):155-75. [DOI:10.1016/j.jelekin.2011.11.008]
37. Kumar S. Energy-cost of lifting in sagittal and lateral planes by different techniques. Inergonomics 1982 (Vol. 25, No. 6, pp. 453-453). One gunpowder square, london, england ec4a 3de: taylor & francis ltd. [DOI:10.1080/00140138408963506]
38. McGill SM, Sharratt MT. Relationship between intra-abdominal pressure and trunk EMG. Clinical Biomechanics. 1990 ;5(2):59-67. [DOI:10.1016/0268-0033(90)90039-9]
39. Golparian M, Anbarian M, Golparian A. Effects of Trunk and Foot Positions on Electromyographic Activity and Co-contraction of Selected Lower Extremity Muscles During Leg-Press Resistance Training. Journal of Advanced Sport Technology. 2021 ;5(1):17-26.
40. Jafarnezhadgero, A., Sadri, A., Bahrami Sharif, M., Alipour Sarinasirloo, M. Comparison of the EMG Frequency Spectrum of Lower Limb Muscles during Weight Training with Traditional and Novel Equipment. Journal of Advanced Sport Technology, 2020; 3(2): 19-31.
41. Farjad Pezeshk, A., Sadeghi, H., Safaeepour, Z., Shariat Zadeh, M. The effect of a custom Area Elastic Surface with different stiffness on hopping performance and safety with an emphasis on familiarity to the surface. Journal of Advanced Sport Technology, 2017; 1(1): 5-14.

## چکیده فارسی

# تاثیر وضعیت اولیه بدن بر کینتیک اندام تحتانی و فعالیت عضلانی عضلات تنه با مهار عامل اینرسی


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**هدف:**

از آنجا که عملکرد مفاصل اندام تحتانی و نقش عضلات تنه در تحمل و برداشتن بار مهم است، این مطالعه با هدف بررسی مداخله موقعیت اولیه بدن هنگام کنترل اینرسی و تأثیر آن بر استراتژی الگوهای حرکتی در پایین مفاصل اندام و همچنین فعالیت عضله تنه در هنگام لیفت متقارن می باشد.

**روش بررسی:**

این مطالعه تجربی شامل ۱۰ مرد جوان سالم بود. از آنها خواسته شد که یک جعبه ۸,۴ کیلوگرمی را در ۳ زاویه مختلف زانو (۱۰- ، ۳۵ و ۷۰ درجه زاویه ران تا خط افق) به صورت متقارن و با سرعت ثابت بردارند. گشتاور اکستانسوری و توان مفاصل اندام تحتانی با استفاده از مدل سه بعدی (LSM) محاسبه شد. برای اندازه گیری فعالیت عضلانی عضلات شکم (rectus abdominus (RA و (external oblique (EO و عضلات کمر ( iliocostalis lumborum (IL و (multifidus (MU، از الکترومیوگرام ۸ کانال استفاده شد.

**یافته‌ها:**

نتایج شباهت الگوی گشتاور اکستنسوری مفاصل اندام تحتانی و کمر ($L_5/S_1$) در بلند کردن سریع و سرعت ثابت را در تکنیک های مشابه نشان دادند. علاوه بر آن هیچگونه اثر معناداری در گشتاور اکستنسوری و فعالیت عضلات کمر هنگام کنترل اینرسی در حرکت اسکات مشاهده نشد و وضعیت های مختلف زانو بر توان اکستنسوری مفاصل اندام تحتانی تاثیر معناداری داشت.

**نتیجه‌گیری:**

مطابق با یافته ها، در بلند کردن بار بدون اثر نیروی اینرسی جعبه، گشتاور اکستنسوری کمر و فعالیت عضلات کمری مستقل از وضعیت قرارگیری زانو بودند. هرچند، توان اکستنسوری مفاصل اندام تحتانی متاثر از وضعیت قرارگیری زانو بودند. علاوه بر این، گشتاور اکستانسوری و فعالیت عضلات کمر در حالت های مختلف زانو نیز رفتار مشابه ای داشتند.



**واژه‌های کلیدی**: پاسچر زانو، گشتاور اکستانسوری لومبار، فعالیت عضلانی عضلات تنه، کنترل اینرسی، اسکات لیفتینگ