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Force and EMG Comparison Between a Weight-bearing Clinical Assessment of Hip Strength Assessment and Non-weight-bearing Tasks

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Purpose: Altered hip strength is a risk factor for lower extremity injury but its relationship to biomechanical dysfunction is debated. Hip strength assessment methods are criticized for using unidirectional, non-weight-bearing positions which may not be representative of athletic activity and may affect comparison to biomechanical analysis of athletic tasks. A weight-bearing task may better represent hip muscle function during these movements. The aim of this study was to identify EMG and force differences for a clinical weight-bearing method of hip strength (the squat-hold) to traditional non-weight-bearing maximal voluntary isometric contractions (MVICs) for hip abduction, extension, and external rotation. **Methods:** Twenty-nine healthy volunteers (23 female, 6 male; 23.3±5.8 years) performed the squat-hold, sidelying abduction, prone extension, and seated hip external rotation MVICs. The squat-hold was performed by exerting a bilateral, maximal force against a rigid strap encircling both knees in a semi-squatted position. Surface electromyography (EMG) recorded peak activation of the gluteus medius (GMed), gluteus maximus (Gmax), and tensor fascia lata (TFL) and a handheld dynamometer simultaneously measured force during all tasks. Peak activation was compared between the squat-hold and each MVIC using paired t-tests. Force was compared across tasks using a one-way ANOVA. **Results:** Greater force was observed during the squat-hold than the external rotation MVIC, but abduction and extension MVICs yielded greater force than the squat-hold. GMax activation was higher during the squat-hold than the external rotation task. TFL activation was higher during the abduction MVIC than the squat-hold but GMed activation was similar across tasks. Peak GMax activation was similar between the extension MVIC and squat-hold. **Conclusions:** Squat-hold force may have been reduced due to altered gluteal moment arms, which affected the length-tension relationship. Clinicians should consider the squat-hold as an alternative assessment of external rotation force, but should continue to assess abduction and extension force with MVICs. Researchers should examine positions optimizing length-tension relationships to better relate motor function and movement patterns. ***Keywords:*** *Hip Strength, EMG, Handheld Dynamometry, Weight-bearing*

INTRODUCTION

Increased peak knee abduction angles (also known as knee valgus position) has been identified as a common predisposition to non-contact anterior cruciate ligament injury, patellofemoral pain, medial tibial stress syndrome, chronic ankle instability, and noncontact ankle sprains.[1-11] Weak hip external rotator and abductors are hypothesized to increase hip internal rotation and adduction, which increases peak knee abduction angles during running and landing.[1,7-10,12-17] For this reason, recent research has focused on identifying relationships between hip strength (i.e. maximal force production), biomechanical dysfunction, and lower extremity injury.[6,8,9,11] Reduced hip strength has been observed in participants with lower extremity injuries compared to healthy individuals, but prospective studies have not identified weakness of the same muscles prior to injury.[1,6,10-11,14,18-19-29] This discrepancy has led to speculation whether hip muscle weakness is risk factor for or a consequence of injury.[19,24,26] Moreover, observed relationships between hip weakness and increased peak knee abduction during athletic activities are inconsistent.[30-33] Knee abduction angles can be influenced by lower leg or foot posture or biomechanics (i.e. pes planus or cavus, subtalar pronation or supination, and/or tibial rotation), external forces that may contribute to an external knee abduction moment (i.e. location and direction of ground reaction forces through the lower limb), or to changes in the role and demand on the hip muscles between non-weight-bearing and weight-bearing positions.[12,15,34] In fact, criticisms of traditional hip strength assessment measures include the unidirectional, non-weight-bearing positions that may not be reflective of athletic movements.[24,30,35,36] This debate can leave clinicians uncertain of how best to assess hip strength and whether traditional measures even relate to athletic movement.

Isokinetic dynamometry is the current gold standard for research methods assessing hip strength.[37] This method requires large, costly equipment that is not always clinically accessible. Handheld dynamometry (HHD) is another reliable and valid measure that allows quantification of manual muscle tests. [37-41] The relatively low cost and ease of use offers a more efficient manner to collect force data in a clinical setting. Both of the aforementioned methods assess force in non-weight-bearing positions in a single plane of motion to despite the fact that during weight-bearing, hip muscles are responsible for postural and pelvic stability in addition to simply moving the legs.[34] It is unknown whether hip muscle force production differs between non-weight-bearing and weight-bearing positions.

Measuring force in a weight-bearing position is difficult due to limitations of isokinetic equipment, but recently, a weight-bearing task was demonstrated to be reliable (ICC=0.99) with force production measured by a lower cost strain gauge, and was comparable to isokinetic dynamometry for the same muscle groups.[39] The evaluated task was a semi-squatted position [30° of hip flexion and 50° of knee flexion, known hence forth as the "squat-hold" (Figure 1)], during which time participants were asked to exert a maximal bilateral force against a band.[39] The resulting motion assessed was concurrent hip abduction and femoral external rotation. Weakness of these muscle groups are commonly associated with injury, and faulty movement patterns associated with injury commonly include hip adduction and femoral internal rotation.[6,11,17,21,24,27] The previous study still used a custom made strain gauge device, as opposed to the more commercially and clinically available HHD, and only assessed hip abduction force, neglecting the external rotation included in the task. In that study, hip abductor force only accounted for about 59% of the variance of the squat-hold results, so inclusion of hip external rotation may be paramount to better assessment of this task. A weightbearing measure of hip strength may be possible using HHD.

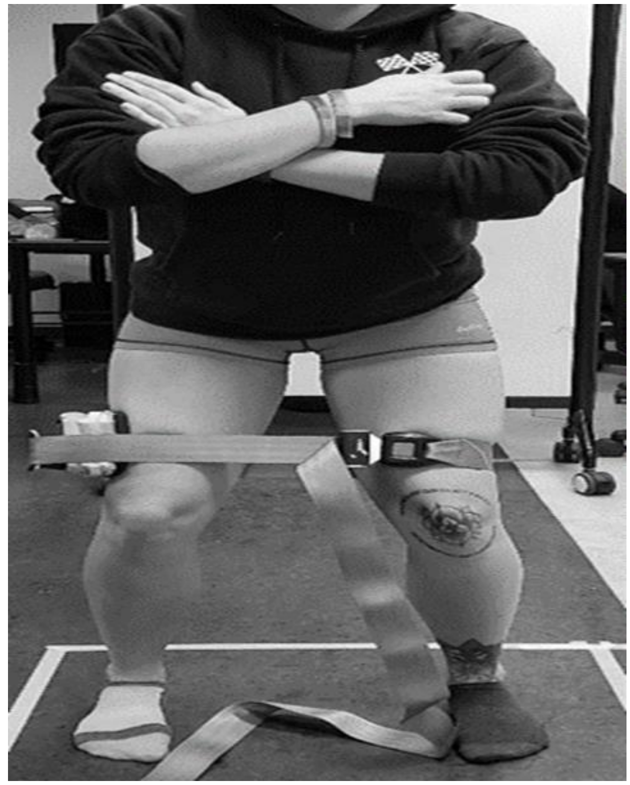

Figure 1. Squat-Hold Task

Gluteal muscle activation has also been assessed using surface electromyography (EMG) in a number of studies examining weight-bearing tasks, however the role of the tensor fascia latae (TFL) during weight-bearing exercises has largely been neglected. Greater gluteus maximus (GMax) muscle activation has consistently been observed during weight-bearing tasks compared to non-weight-bearing tasks[39,42,43] Higher gluteus medius (GMed) activation has been observed during in the weight-bearing leg compared to the non-weight-bearing leg during standing straight leg abduction, in the stance leg during stair descent, and during the squat-hold task[39,42-45] Due to its weight-bearing position, higher peak hip muscle activation may be observed during the squat-hold task than during traditional non-weight-bearing measures.

The purpose of this study was to identify differences in force and peak muscle activation of the GMed, GMax, and TFL muscles between a weight-bearing isometric measure of hip strength (the squat-hold task) and traditional non-weight-bearing maximal voluntary isometric contraction (MVIC) tasks. We hypothesized that greater peak muscle activation would be observed during the weight-bearing squat-hold task but similar force output would be observed between the squat-hold and non-weight-bearing abduction (TFL and GMed) and extension (GMax) MVIC.

MATERIALS & METHODS

Participants

This cross-sectional study was approved by the campus Institutional Review Board. All participants provided informed consent prior to taking part in the study. Healthy adults between the 18-50 years were recruited on campus. Eligible participants were free of trunk, back, and lower extremity injury for the past 6 months, had not undergone surgery to those areas in the past year, were not experiencing pain to those areas at the time of testing, were not pregnant, and were free of neurological conditions. Additionally, due to the negative impact of high body fat content on EMG signals, any participant with a body mass index equal to, or greater than 30 were excluded from participation.[45] To calculate body mass index, the participant's weight in kilograms was divided by their height in meters, squared [(weight (kg) / height squared (m^2)].

An a priori power analysis was conducted using G*Power 3.1, a statistical software package to estimate sample size based on study design.[46] Calculated based on data from a previous study, in order to achieve statistical power of 0.80 with an α=0.017, assuming a moderate effect, a total sample of 26 participants was needed.[39] In order to account for potential attrition, a recruitment effort of 20% was added, for a total recruitment effort of 32 participants.

Thirty participants volunteered to take part in the study. One participant's data was lost due to technical error, leaving the data of 29 participants for analysis. Twenty-three participants were female and six were male (24.1 ± 5.84 years, 170.8 ± 7.37cm, 66.2 ± 10.2kg, mean BMI 22.4 ± 2.5). While males typically exhibit higher hip muscle force production than females, the within-subjects nature of the analysis does not consider between-individual or between-sex differences. Participants reported to the laboratory for one test session where EMG and force were recorded simultaneously for each of three tasks (abduction MVIC, extension MVIC, and Squat-Hold task).

Force & Tasks

Hip abduction, extension, and external rotation MVICs were completed using the "make" technique with a HHD (Lafayette Instruments, Lafayette, IN, USA) measuring force (kg).[47-48] The abductors (GMed and TFL) were assessed in a side-lying position on a table with the torso and pelvis strapped to the table (Figure 2). A third strap secured the dynamometer just superior to the lateral femoral condyle. The participant exerted a maximal isometric force against the HHD.

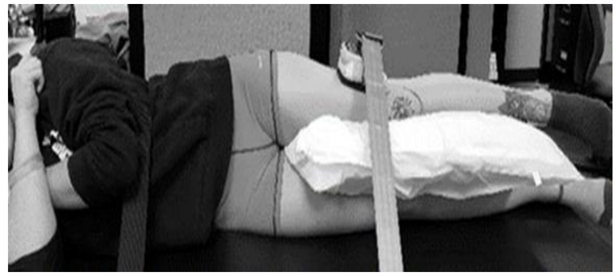

Figure 2. Hip Abduction MVIC

Hip extension force was assessed, as that is the primary function of the GMax. The extensors were assessed in a prone position with the knee flexed to 90° (Figure 3) with similar strapping of the pelvis, torso, and legs. The HHD was secured just superior to the posterior knee joint line while the participant extended the hip while maintaining knee position. Hip internal rotation force was not measured, as the internal rotators were not hypothesized to contribute to the force in the squat-hold task.

The GMax is an active external rotator during the squat-hold, therefore external rotation MVICs were also assessed. External rotation was assessed with the participant seated at the edge of a table. A rigid strap held the dynamometer in place just superior to the medial malleolus and around an immovable post. The participant was instructed to maximally externally rotate the femur against the HHD (Figure 4).

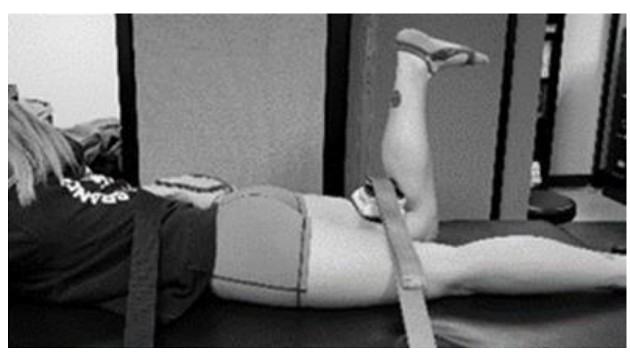

Figure 3. Hip Extension MVIC

Figure 4. External Rotation MVIC

The squat-hold consisted of a bipedal semi-squatted position with a HHD secured just superior to the right lateral femoral condyle with a rigid strap that encircled both legs (Figure 1). The semi-squatted position was measured using two digital inclinometers (Lafayette Instruments, Lafayette, IN; sensitive to 0.01°) such that the hip flexion was 30° and knee flexion was 50°.[39] While maintaining this position, the participant exerted a bilateral outward force against the strap and HHD. Proper position was re-confirmed before subsequent trials.

During all tasks, participants were instructed to maintain maximal force for five seconds. Two practice trials occurred at increasing levels of effort, followed by three recorded trials. The average of these trials was used for analysis. One minute of rest occurred between each trial and five minutes of rest occurred between each task. Participants were verbally encouraged to facilitate a maximal response. Force and EMG data were recorded simultaneously during this task, and the peak average RMS of three trials was used for analysis. Following completion of MVICs, participants were provided five minutes of rest before beginning the squat-hold task.

Electromyography

Surface EMG data were collected for the GMed and TFL, representing the hip abductors, and from the GMax, as it is the largest, most superficial hip external rotator.[34] GMed and TFL EMG were both assessed during two tasks: the hip abduction MVIC and the squat-hold. GMax EMG was assessed during three tasks: the extension MVIC, external rotation MVIC, and the squat-hold. Procedures for skin preparation, electrode placement, and data recording were performed according to Surface EMG for the Non-Invasive Assessment of Muscles (SENIAM) guidelines.[49] Surface EMG were collected using the Noraxon MyoMuscle (Noraxon, Scottsdale, AZ, USA) system, recording at 1500 KHz. Prior to data collection the skin was lightly abraded with coarse gauze and then cleaned with an alcohol wipe. Square Ag/AgCl electrodes with a 20 mm interelectrode distance were placed on the GMax, GMed, and TFL and proper location was confirmed by performing maximal voluntary isometric contractions (MVICs). Analog signals were converted to digital signals and differential amplifiers were used to reject common noise (90dB) and to amplify signal gain (2000).

EMG data were processed using Visual 3D (C-motion, Germantown, MD). EMG data were rectified, providing root mean square (RMS) values, which were smoothed over a 150ms moving window. The average peak amplitude for each trial, muscle, and task was used for analysis. Because within-subjects comparisons were made using the MVIC as the comparator, we did not normalize EMG to a percentage of that MVIC. All EMG recordings occurred within a single session, limiting the likelihood of electrode movement during testing.

Statistical Analysis

Paired-samples t-tests (α=0.05) analyzed differences in peak muscle activity between the following comparisons: 1) abduction MVIC and squat-hold, 2) extension MVIC and squat-hold, and 3) external rotation MVIC and squat-hold.

A one-way ANOVA analyzed differences between force (kg) across four tasks (squat-hold, abduction, extension, and external rotation MVICs). Planned comparisons were compared to a Bonferroni-adjusted α (0.05/3= 0.017) to compare force between each MVIC to the squat-hold.

RESULTS

Muscle Activity

Greater peak GMax activation was observed during the squat-hold task than the external rotation MVIC [mean difference= -20.1 μV, t_{28}=-5.6, p<0.001; 95% Confidence Interval (CI)= -28.1, -12.2].

		Peak Activation (μV)		
Task	**Force (kg)**	**GMed**	**TFL**	**GMax**
Abduction MVIC	24.9 ± 6.1[a]	13.1 ± 12.0	36.6 ± 29.2[b]	--
Extension MVIC	23.7 ± 8.7[a]	--	--	20.9 ± 27.7
External Rotation MVIC	6.7 ± 3.0[a]	--	--	5.5 ± 3.9[b]
Squat-Hold	16.8 ± 6.5[a]	19.2 ± 26.0	23.0 ± 16.5[b]	26.9 ± 20.7[b]

Table 1. Force & Peak Activation by Task presented as Means ± SD
[a]=Significant force difference to the p≤0.001 level [b]=Significantly peak activation difference to the p<0.001 level.

There was a large effect for this difference (d= 0.96). Greater peak TFL activation was observed during the abduction MVIC task (m=36.6μV) than the squat-hold task (mean difference= 13.7μV; t_{28}=3.4, p=0.002; 95%CI= 5.5, 21.8) with a moderate effect (d=0.63). No significant differences in peak GMed activation between the abduction MVIC (m=13.1μV) and the squat-hold

task (mean difference= -6.13μV; t_{28}=-1.8 p=0.081; 95% CI= -13.07, 0.81), or for peak GMax activation between the extension MVIC (mean difference= -5.88μV) and squat-hold [m=26.9μV; t_{28}=-2.0, p=0.55; 95%CI= -11.9, 0.15; (see Table 1)]. Small effect sizes were noted for both the GMed (d=0.33) and GMax in extension (d=0.37).

Force Output

Significant force differences were observed across tasks ($F_{3,116}$= 50.9, p<0.001). The external rotation MVIC yielded less force than the squat-hold (mean difference= 10.1kg; p<0.001, 95% CI= 4.88, 15.3). However, the abduction and extension MVICs yielded greater force than the squat-hold (abduction MVIC vs. squat-hold: mean difference= 8.1kg, p<0.0001 95%CI= -13.4, -2.8; Extension MVIC vs squat-hold: mean difference= -1.17, p=0.001; 95%CI= -12.2, -1.7; (see Table 1)].

DISCUSSION

The purpose of this study was to determine whether force or peak activation differences occurred between a weight-bearing squat-hold task and non-weight-bearing hip MVIC tasks. The squat-hold was originally conceived as an alternative hip abduction task, however, the original authors concluded that hip external rotation also occurred during the task.[34] We hypothesized that we would observe greater peak muscle activation, but similar force production during the squat-hold compared to the non-weight-bearing MVICs. This hypothesis was based on previous findings suggesting that peak hip muscle activation is greater during weight-bearing tasks but assumed that increased activity may be due to pelvic and postura stabilization, which may not equate to increased force production during the task.[34,39,42,43] The key finding from this study is that greater external rotation force was observed with the squat-hold compared to the external rotation MVIC. We also observed similar peak GMed and GMax muscle activation across tasks but greater force production was observed during abduction and extension MVICs compared to the squat-hold. Additionally, higher peak TFL activation was observed during the abduction MVIC compared to the squat-hold. This suggests that the squat-hold task may be a better measure of hip external rotation force than a seated external rotation MVIC but standard abduction and extension force may be better assessed with traditional MVICs.

Lee & Powers were the first to describe the squat-hold position.[39] In that study, the muscle demonstrating the greatest activation was the GMax (93% MVIC). The authors hypothesized that this was due to the external rotation torque required during the task however, external rotation MVICs were not assessed or compared to the squat-hold.[39] In our study, we found greater force production during the squat-hold compared to the external rotation MVIC, but no significant differences in peak activation across tasks. While the GMax is a large, superficial external rotator, external rotation is not the primary action of the GMax. Superficial EMG cannot be assessed for the primary external rotators, which lie deep to the gluteal muscles. We chose to assess EMG of the GMax during extension and external rotation as well as force for both of these motions. The greater force generated during the squat-hold indicates that it may be a viable alternative to clinically assess hip external rotation force, however, less peak GMax activation during the squat-hold indicates that the GMax may not be the biggest contributor to this external rotation.

Greater peak GMed activation has also been reported during the squat-hold compared to an abduction MVIC but no significant force differences were observed in that study.[39] This was contrary to our study, in which no significant differences were observed for peak GMed or GMax activation, and the MVICs yielded greater force production. For EMG, we observed slightly lower activity of the GMed (67% MVIC) than the previous study (77% MVIC) despite similar participant samples.[39] The difference in data collection procedures may partially explain the difference in MVIC. Lee and Powers [2013] used a force transducer for the squat-hold and isokinetic dynamometry to measure MVICs, and visual feedback was provided to participants, which could have facilitated maximal responses. In our study, a HHD was used to collect all force assessments.[39] This should not be a reason for differences in results, as similar force outputs have been reported between devices.[50] Also, we verbally encouraged participants to facilitate maximal response but no visual feedback was provided. The use of visual feedback increases force production and reduces force variability.[51-52]

Another rationale for the reduced force observed during the squat-hold compared to the MVICs may be related to altered moment arms in the hip flexed position of the squat-hold compared to the neutral (sagittal plane) hip position during abduction and extension MVICs. As the hip flexes, the internal rotation moment arm of the GMed and GMax increases, which may result in lower force production despite similar activation levels.[34,53] So, while participants were being asked to generate force in external rotation and abduction during the squat-hold, the GMed and GMax had reduced mechanical advantage to complete the external rotation of the task. Additionally, the TFL is commonly referred to as a hip internal rotator, however, Dostal et al demonstrated that the TFL has no transverse plane leverage in a neutral sagittal plane hip position (0°) and limited evidence has been provided of its transverse plane leverage at other sagittal plane angles.[54] This change in leverage may explain the higher TFL activation and force observed during the abduction MVIC in our study.

Peak GMed peak muscle activation was also assessed by Bolgla and Uhl during a variety of weight-bearing and non-weight-bearing positions commonly used in rehabilitation programs.[42] One key difference in the Bolga and Uhl study is that weight-bearing positions were initiated with the hips in neutral sagittal plane position.[42] This neutral starting position theoretically optimizes the length-tension relationship for the GMed, allowing a similar mechanical advantage as the non-weight-bearing position for both muscles.[34,50] However, in that study, lower overall GMed activation was observed (range= 28-57% MVIC for all exercises) than with the squat-hold in our study (67% MVIC).[42] It should be noted that the authors assessed EMG during repetitions of isotonic exercises rather than during maximal contractions, which may account for the lower range of muscle activation observed.[42] While this study did not utilize the squat-hold position, the results support muscle activation differences during weight-bearing and non-weight-bearing positions and supports continued exploration of weight-bearing positions that optimize hip muscle activation and force production.

The results of this study should be considered within its limitations. An isometric task was selected in order to avoid movement artifact that could interfere with the EMG signals and while maximizing potential force production. This task was also selected because of its ability to activate the hip abductors and external rotators, however, the task was still isometric so the question of whether the weight-bearing task is more comparable to hip muscle function during athletic movements (i.e. running, landing) has yet to be answered. We observed relatively low GMax force and EMG during the seated external rotation task, which may be due to the position of the task itself. One author suggested that in order to achieve maximal force against the band in the seated position, participants may adduct against the bad rather than externally rotate.[50] Careful instruction was provided and participants were passively taken through the external rotation movement prior to MVIC performance, but we did not assess adductor EMG, so there is no way to determine whether this compensation occurred during this study.[50] Additionally, force couples at the hip may rely on a balance between internal rotation and external rotation to produce rotation of the femur.[34] Hip internal rotation force and EMG were not measured in this study, as the muscle activity was not hypothesized to contribute to the force of the task. Future research would need to address the contribution of the hip internal rotators to the squat-hold task. While rotational force couples at the hip can also be affected by the available range of motion and the degree of femoral torsion (i.e. retro- or antero-version), these measures were not measured during this preliminary investigation as their influence on the squat-hold task was thought to be minimal.[34]

We were unable to assess muscle activity of the deep hip external rotators (obturator, gemellus), which may have contributed more to external rotation than the GMax during the squat-hold. Fine-wire EMG can be used to assess activity of the deep muscles, however, it is a more invasive technique that may not be clinically applicable. The hip muscles are deep to adipose tissue as well as adjacent to surrounding musculature, which could cause signal interference. We attempted to control for the depth of muscle under adipose tissue by excluding individuals with BMI over 30.[45] Lastly, the statistical analysis which included multiple t-tests increases the risk of Type I error, or falsely rejecting the null hypothesis. Despite these limitations, it appears that the squat-hold may be a viable alternative clinical assessment of hip external rotation force.

CONCLUSIONS

Clinicians should consider using the squat-hold task to assess hip external rotation force, but should continue to assess hip abduction and extension using traditional MVIC positions. It is important to note that the squat-hold task requires both abduction and external rotation in order to accomplish the task, however, using this position to assess hip abduction force as previously suggested, is debatable.[39] Due to the established relationship between hip strength and injury occurrence, researchers should continue to investigate optimal positions for weight-bearing hip abduction force assessment that also optimizes muscle activation. This may be the next step to identify more effective, efficient, and consistent measures that relate well to athletic tasks such as running and landing.

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