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THE INFLUENCE OF CORE MUSCULATURE ENGAGEMENT ON HIP AND KNEE KINEMATICS IN WOMEN DURING A SINGLE LEG SQUAT

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ABSTRACT

Purpose/Background: Excessive frontal plane motion and valgus torques have been linked to knee injuries, particularly in women. Studies have investigated the role of lower extremity musculature, yet few have studied the activation of trunk or "core" musculature on hip and knee kinematics. Therefore, this study evaluated the influence of intentional core engagement on hip and knee kinematics during a single leg squat.

Methods: Participants (n = 14) performed a single leg squat from a 6 inch step under 2 conditions: core intentionally engaged (CORE) and no intentional core engagement (NOCORE). Participants were also evaluated for core activation ability using Sahrmann's model, and the resulting scores were used to divide participants into low (LOWCORE) and high scoring (HIGHCORE) groups. All trials were captured using 3-D motion analysis, and data were normalized for height and time. Paired t-tests and repeated measures, mixed model MANOVAs were used to assess condition and group differences.

Results: The CORE condition, compared to NOCORE, was characterized by smaller right [t(13) = 3.03, p = .01] and left [t(13) = 3.04, p = .01] hip frontal plane displacement and larger knee flexion range of motion [t(13) = 3.08, p = .009]. Subsequent MANOVAs and follow-up analyses revealed that: (1) the CORE condition demonstrated smaller right and left hip medial-lateral displacement in the LOWCORE group (p = .001), but not in the HIGHCORE group; (2) the CORE condition showed larger overall knee flexion range of motion across LOWCORE and HIGHCORE groups (p = .021); and (3) the HIGHCORE group exhibited less knee varus range of motion across CORE and NOCORE conditions (p = .028).

Conclusions: Intentional core activation influenced hip and knee kinematics during single leg squats, with greater positive effect noted in the LOWCORE group. These findings may have implications for preventing and rehabilitating knee injuries among women.

Level of Evidence: 2B, Cohort laboratory study, mixed model design

Key Words: Biomechanics, Core Musculature, Kinematics, Knee

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INTRODUCTION

It is widely accepted that excessive frontal plane motion and valgus torque in the knee contribute to an increased risk of anterior cruciate ligament (ACL) injuries.[1] However, the reasons for the greater frontal plane motion are not well understood, as a lack of consensus exists in the explanations posited in the literature.[2,3] The lack of consensus presents limitations to effective training of healthy individuals and to the rehabilitation of individuals with knee injuries.[4,5]

The majority of ACL injuries are non-contact in nature, often occurring when the foot is planted during a cutting motion or landing from a jump.[6] During such activities, the knee is often subjected to excessive frontal plane motion, as a result of internal rotation of the hip, adduction of the femur, and external rotation of the tibia.[7,8] The comparative risk of injury per exposure is 2 to 8 times greater in women than in men,[9] and researchers have shown that women often exhibit greater knee frontal plane motion and valgus torques during dynamic activities than men, which has been shown to contribute to the greater incidence of ACL injury in this subgroup.[7]

Dynamic knee stability is achieved through the neuromuscular control of a multifactoral kinetic chain, as the knee is directly supported by the immediately surrounding muscles, yet is also dependent on more proximal muscles of the hip and trunk.[2,10,11,12] Differences in neuromuscular control of the hip have been cited as an important source of sex differences in lower extremity movement patterns.[12] Bohannon et al.[13] identified 19% greater body weight-normalized hip abduction isometric strength in males than in females. Similarly, Cahalan et al.[14] reported 39% greater hip external rotation torque in males versus females without normalization to body weight. Other researchers have investigated the role of the hip abductors in eccentrically controlling hip adduction during weight bearing activities.[8] Claiborne et al.[12] found that the hip abductors play a significant role in stabilizing the femur during unilateral weight bearing activities. Wilson et al.[15] also found that weakness in the hip external rotators was associated with abnormal knee motion in both men and women. In contrast, in a study of women performing a step down task, Hollman et al. found that hip stability as quantified by knee valgus patterns was more related to gluteus maximus recruitment capacity (as measured with EMG) than gluteus maximus strength (as measured with dynamometry). This finding led Hollman et al[11] to theorize that while gluteus medius strength may provide benefits in resisting adduction of the femur, these benefits are somewhat mitigated by the role the gluteus medius plays in the internal rotation of the femur at some positions of available hip range of motion. In other words, effective neuromuscular control of the lower extremity kinetic chain is subtly influenced by issues such as hip joint position, strength, and proprioception during functional activities. The variety of findings has led researchers to investigate additional anatomical regions in an effort to further describe the variability seen at the knee during unilateral functional tasks and during landing.

Knee motion during functional activities cannot be solely attributed to hip or knee musculature.[4,10] While their study did not assess muscular contributions to lumbopelvic or core control, Claiborne et al.[12] noted that only 22% of knee variability could be attributed solely to hip muscles surrounding the hip, pointing out the limitations of including only the lower extremity in analysis of functional activities such as cutting and landing. Consequently, researchers are increasingly looking at the role that core musculature may play in lower extremity function.[16,17,15] Although not a universally-accepted definition, "core musculature" or "core" in this paper will be defined as "muscles of the trunk and pelvis that are responsible for the maintenance of stability of the spine and pelvis and help in the generation and transfer of energy from large to small body parts", or from the trunk to the extremities.[16,p.189] As this definition suggests, the core musculature, due to proximal position in the axial skeleton, can influence motion of distal segments in the appendicular skeleton. Existing work suggests that proximal core muscles, such as the transversus abdominis, obliques, and multifidi, play an important role in spinal stabilization, and that poor function of these muscles is associated with low back pain and dysfunctional movement patterns.[18,19,20] Furthermore, a growing number of authors advocate that "core stability" contributes to improved function of the lower extremities during gross motor activities.[4,15,21,22,16]

No studies have specifically investigated the role of core engagement on lower extremity kinematics during a functional task. Therefore, the purpose of this study was to determine how intentional activation of the core musculature affected hip and knee kinematics during a single leg squat. The authors hypothesized that purposeful core engagement – compared to *in vivo* core activity – would result in significant differences in hip and knee linear displacements and angular measures. A secondary purpose of this study was to compare the performance during the single leg squat between two groups, based on their scores on a common clinical measure for recruitment of core muscles. Based upon the authors' observations working with clinical populations, we hypothesized that individuals with lower core recruitment scores would differ in their lower extremity kinematics, compared to those who scored higher on this clinical measure for core recruitment.

METHODS

Participants

A quasi-experimental design was used to assess the combined effects of core engagement and recruitment scores on kinematic measures in the lower extremities. A sample of convenience consisting of fourteen women age 20 to 24 years (mean (SD): age of 22.0 (1.2) yrs, height of 170.7 (6.2) cm, weight of 62.8 (8.5) kg) participated in this study. Only healthy, college-age women were included in order to minimize confounding effects related to sex or health status. Participants were excluded if they reported current pain or injury to the lower extremities or torso (including the spine and abdominal cavity), or if they had a history of any lower extremity injuries or surgical procedures in the twelve months prior to testing. The study protocol was approved by Rockhurst University's Institutional Review Board, and all testing was conducted in the Human Performance Laboratory in the Department of Physical Therapy Education. Before testing, written informed consent was acquired from all participants following the federal guidelines related to research involving human participants. Demographic data were collected from each participant and are compiled in Table 1.

Procedures

All participants were assessed for their capacity to effectively recruit the "core stabilizers" using the lower abdominal strength test as described by Sahrmann as commonly used in the clinical evaluation of patients with low back pain.[23] This evaluation model is based, in part, on the notion that the abdominal muscles provide important support for the spine during functional activities and low level muscle activation is needed for many tasks. For example, a study completed at Yale University suggests that only 2-3% of maximum voluntary activity of the abdominal muscles is needed to stabilize the spine during upright unloaded tasks.[24] Thus, the Sahrmann protocol aims to assess this level of abdominal level muscle activation and contains 5 testing levels, each designed to make it increasingly difficult to maintain a neutral spinal position using the involved core stabilizers. During testing, the physical therapist places the participant in the supine hook lying position, and then palpates the participant's engagement of the lower abdominals with one hand and the spinous processes of the participant's lumbar spine with the other hand. The physical therapist then monitors the participant's spine as one hip is flexed passively to 90 degrees and the other lower extremity is then lifted from the supporting surface. In application, a participant would be scored as a "0" if she could elicit a palpable engagement of the lower abdominal muscles but could not maintain a spinal neutral posture when instructed to lift one foot from the supporting surface, and scored as a "1" if she could, and so on through the five increasingly challenging levels of the test (Table 2). Previous authors have found this a valid and reliable clinical measure of the capacity to isometrically recruit lower abdominal muscles involved in core stabilization.[25,26] As a means of reducing the potential for measurement error related to inter-rater variability, one mem-

Table 1. *Characteristics of the participants (n = 14).*

	Minimum	Maximum	Median	Mode	Mean	SD
Age	20.00	24.00	22.00	23.00	22.00	1.24
Height (cm)	161.29	179.07	171.45	162.56	170.70	6.18
Weight (kg)	48.64	77.73	63.86	53.18	62.78	8.52
Core Rating	0.00	2.00	0.00	0.00	0.36	0.63

Table 2. *Sahrmann's testing progression/scoring criteria for lower abdominal strength/core activation.*

Manual Muscle Test Grade	Criteria
1/5	The subject lifts one leg at a time to 90° of flexion with the knees positioned in flexion. From this position the subject lowers one leg at a time to the client position. Back remains flat.
2/5	The subject successfully performs Level 1, but upon lowering one leg to the table, s/he slides the leg into extension. The heel of the active leg may slide on or touch the surface of the treatment table during execution. The opposite leg must maintain a position of hip flexion of 90°, but no more, and its heel cannot touch the treatment table. Once the active leg has completed the slide into extension, the subject will rest the leg on a table, lift it back off the table, and return to the position of 90° of hip flexion before repeating with the other leg.
3/5	For Level 3, the subject performs Level 2, but instead of sliding the leg, s/he extends the leg while maintaining it off the treatment table through the entire range of motion. Once the subject completes extension, she rests the leg on the table, lifts the leg from the table, and returns it to the 90° hip flexed position before repeating the motion within the other leg.
4/5	The subject repeats level 1, but instead of lifting one leg at a time off the table, both legs and lifted simultaneously to the 90° hip flexed position, returned to the hook lying position, and fully extended. The return movement is completed by simultaneously sliding both legs back to the hook lying position followed by a bilateral leg lift into 90° of hip flexion.
5/5	For Level 5, the subject repeats the task for Level 4, but rather than sliding both legs along the surface of the treatment table, s/he extends both legs simultaneously, rests the legs of the completion of extension, lifts both legs from the table, and finally returns lands to the 90° hip flexed position.

ber of the research team completed all assessments of core stabilization on the participants in this study. Each participant was tested using this model three times as a means of repeated measures assessment. Five participants scored 1 or 2 on the Sahrmann test, with the remaining participants scoring 0. Due to the small range, resulting scores were used to divide participants into a low-scoring group (LOWCORE: test score of 0) and a high-scoring group (HIGHCORE: test score of 1 or 2).

After assessment of core stability, a single leg squat (SLS) task using a six inch step was explained and demonstrated to each participant. Again, one member of the research team provided all demonstration and instruction to the participants as a means of minimizing confounding variables. The SLS was chosen because while it is a controlled movement, it is a dynamic maneuver that can be extrapolated to many functional activities such as landing, running and cutting,[12] and it is widely used in contemporary physical therapy practice for evaluating lower extremity motion.[20] Experimental conditions were standardized by having each participant perform the SLS using her dominant leg. Leg dominance was identified from an item on the intake questionnaire in which the participant identified which leg she used to kick a ball, as used in previous work.[27] All participants were found to be right-leg dominant. Each participant performed two SLS. Starting with the toes pointing forward, the participant maintained the weight bearing, dominant (right) foot on the step and lightly touched the heel of non-dominant (left) foot on the ground during the squat (Figure 1). Once the participant's heel tapped the ground the participant returned to a standing position on top of the step. As a means of constraining the motor task in a way that minimized confounding variables, participants were allowed to practice the task until the speed and consistency of heel contact were satisfactorily completed within a three second timeframe as measured by a member of the research team using a timer. Participants performed the SLS under two conditions: performing the squat with or without core musculature engagement (CORE and NOCORE,

Figure 1. *Participant performing single leg squat (SLS) in laboratory environment.*

respectively). For the CORE condition, participants were instructed to engage their abdominal muscles as they had done during the Sahrmann test of core activation described above, and as is commonly done when teaching therapeutic exercise to patients undergoing physical therapy. Verbal cues only were used during this portion of the protocol. As no tactile cues were provided, no assessment was made on the quality of the core musculature engagement during this portion of the testing. The order in which the participants completed the CORE and NOCORE conditions was randomized to prevent order bias.

All participants wore their own athletic shoes and shorts during testing. The shirts and shorts were adjusted to reveal the anterior superior iliac spines (ASIS) of each participant. Retroreflective markers (12.7 mm) were placed on bony prominences of each subject's left and right ASIS (LASI and RASI), lateral femoral condyle of the dominant leg (RKNE) and lateral malleolus of the dominant leg (RANK).

Measurements

Marker positions were tracked in three dimensions during each SLS using an Ariel Performance Analysis System (APAS) 2007 (Ariel Dynamics, Inc., San Diego, CA, USA) with two Panasonic PV-GS320 digital video cameras (Panasonic Corporation of North America, Secaucus, NJ, USA), each positioned at 45 degrees to the participant and operating at a sampling rate of 60 Hz. The APAS motion analysis system has been found valid and reliable and is widely used in biomechanics research.[28,29,30] All corresponding three-dimensional data underwent a three Hz low pass digital filter to smooth marker trajectories.

Data Analysis

All data was imported and processed using MATLAB R2009b (The MathWorks, Inc., Natick, MA, USA). Each participant's data sequence was time normalized to 0-100% of the SLS duration to reduce variation among participants and allow direct comparison between trials. Motion data was used to create joint angle data sequences: knee flexion and varus angles were defined as sagittal and frontal-plane projection angles subtended by one line connecting the RASI and RKNE markers and a second line connecting the RKNE and RANK markers; hip abduction angle was defined as the frontal-plane projection angle subtended by one line connecting the RASI and LASI markers and a second line connecting the RASI and RKNE markers.[11] Joint ranges of motion for knee flexion ($\theta_{knee,flx}$), knee varus ($\theta_{knee,var}$), and hip abduction ($\theta_{hip,abd}$), defined as the difference between maximum and minimum values, were extracted from joint angle data sequences. Since excessive frontal plane knee motion is often associated with ACL injury as discussed above, motion data sequences were also used to define maximum joint marker displacements, defined as the difference between maximum and minimum marker positions in the medial-lateral direction for RASI ($d_{RHIP,lateral}$), LASI ($d_{LHIP,lateral}$), and RKNE ($d_{RKNE,lateral}$) markers. Similarly, since knee flexion range of motion is often associated with higher function when performing a squat during sporting and other functional activities,[16,31,32] motion data were also used to define maximum joint marker displacements in the vertical direction for RASI ($d_{RHIP,vertical}$), LASI ($d_{LHIP,vertical}$), and RKNE ($d_{RKNE,vertical}$) markers. All displacement variables were normalized to participant height (cm).

STATISTICAL METHODS

Statistical analysis was performed with SPSS 17.0 (SPSS, Inc., Chicago, IL, USA). To address the pri-

Table 3. *Mean and standard deviation values for hip and knee kinematics during the single leg squat.*

Variable		Group	Mean (SD)	Comparison
Hip Displacement	$d_{RHIP,vertical}$ (cm/cm)	CORE NOCORE	0.09 (0.02) 0.09 (0.02)	t=2.02 p=.064
	$d_{RHIP,lateral}$ (cm/cm)	CORE NOCORE	0.05 (0.02) 0.06 (0.02)	t=-3.03 p=.010†
	$d_{LHIP,vertical}$ (cm/cm)	CORE NOCORE	0.12 (0.02) 0.11 (0.01)	t=2.31 p=.038
	$d_{LHIP,lateral}$ (cm/cm)	CORE NOCORE	0.05 (0.02) 0.06 (0.02)	t=-3.03 p=.010†
Hip Range of Motion	$\theta_{hip,abd}$ (degrees)	CORE NOCORE	15.26 (3.44) 15.11 (3.76)	t=0.15 p=.883
Knee Displacement	$d_{RKNE,vertical}$ (cm/cm)	CORE NOCORE	0.06 (0.01) 0.06 (0.01)	t=0.91 p=.378
	$d_{RKNE,lateral}$ (cm/cm)	CORE NOCORE	0.03 (0.01) 0.03 (0.01)	t=-0.35 p=.732
Knee Range of Motion	$\theta_{knee,flx}$ (degrees)	CORE NOCORE	55.78 (6.55) 54.47 (6.17)	t=3.08 p=.009‡
	$\theta_{knee,var}$ (degrees)	CORE NOCORE	7.71 (2.82) 6.86 (2.14)	t=1.16 p=.266

*Comparisons are based on normalized data, as displacement is expressed in centimeters (cm) relative to participant height in centimeters (cm).
†Significant core effect at the α = .0125 level.
‡Significant core effect at the α = .025 level.

mary purpose of the study, paired t-tests were utilized to compare the CORE and NOCORE conditions in regards to joint marker displacement and ranges of motion. To correct for the effect of multiple comparisons, the Bonferroni method was applied to groups of variables (hip displacements, hip range of motion, knee displacements, and knee range of motion) to obtain corrected alpha values of .0125, .05, .025, and .025, respectively. To address the secondary purpose, 2 X 2 mixed model repeated measures multivariate analyses of variance (MANOVA) were conducted to investigate the interaction between core stabilization test scores (HIGHCORE and LOWCORE) and core activation condition (CORE and NOCORE), with core condition used as a within-participant factor and stabilization score as a between-participant factor. Separate MANOVAs were performed on the same groups of variables described above (hip displacements, hip range of motion, knee displacements, and knee range of motion). Bonferroni corrections were applied within each analysis to correct for the effects of multiple comparisons.

RESULTS

Effects of Core Engagement on Hip Kinematics

Mean (SD) normalized displacements (normalized displacements are expressed in centimeters relative to participant height in centimeters) for both right and left hips were 0.05 (0.02) and 0.06 (0.02) for the CORE and NOCORE conditions, respectively. Means and standard deviations for all hip displacement and angular range of motion are presented in Table 3. Paired t-tests revealed significant core engagement effects on hip frontal plane displacement, but not angular range of motion. The CORE condition, compared to NOCORE, was characterized by smaller hip displacements, as measured by $d_{RHIP,\ lateral}$ [$t(13) = -3.03$, $p = 0.01$] and $d_{LHIP,\ lateral}$ [$t(13) = -3.04$, $p = 0.01$], respectively.

Effects of Core Engagement on Knee Kinematics

Mean (SD) knee angular range of motion was 55.78 (6.55) degrees and 54.47 (6.17) degrees for CORE and NOCORE conditions, respectively. Means and standard deviations for all knee displacement and angular range of motion are presented in Table 3. Paired t-

Table 4. *Results of 2x2 mixed model MANOVAs comparing hip and knee displacement and range of motion values within conditions (CORE and NOCORE) and between groups (LOWCORE and HIGHCORE).*

Variable		Group	CORE Mean (SD)	NOCORE Mean (SD)	Condition Main Effect	Group Main Effect	Interaction
Hip Displacement†	$d_{RHIP,vertical}$ (cm/cm)	LOWCORE HIGHCORE	0.09 (0.02) 0.09 (0.02)	0.09 (0.02) 0.09 (0.02)	F(1,12)=2.55 p=.137	F(1,12)=0.28 p=.606	F(1,12)=2.58 p=.135
	$d_{RHIP,lateral}$ (cm/cm)	LOWCORE HIGHCORE	0.04 (0.02) 0.05 (0.02)	0.06 (0.01) 0.05 (0.02)	F(1,12)=7.38 p=.019	F(1,12)=0.37 p=.555	F(1,12)=5.95 p=.031
	$d_{LHIP,vertical}$ (cm/cm)	LOWCORE HIGHCORE	0.12 (0.01) 0.11 (0.02)	0.11 (0.01) 0.11 (0.01)	F(1,12)=3.55 p=.084	F(1,12)=0.85 p=.374	F(1,12)=2.85 p=.117
	$d_{LHIP,lateral}$ (cm/cm)	LOWCORE HIGHCORE	0.04 (0.02) 0.06 (0.02)	0.06 (0.01) 0.06 (0.02)	F(1,12)=7.68 p=.017	F(1,12)=0.18 p=.675	F(1,12)=7.31 p=.019
Hip Range of Motion	$\theta_{hip,abd}$ (degrees)	LOWCORE HIGHCORE	15.77 (3.82) 14.33 (2.74)	14.42 (4.46) 16.35 (1.79)	F(1,12)=0.13 p=.727	F(1,12)=0.02 p=.896	F(1,12)=3.23 p=.097
Knee Displacement	$d_{RKNE,vertical}$ (cm/cm)	LOWCORE HIGHCORE	0.06 (0.01) 0.06 (0.01)	0.06 (0.01) 0.05 (0.01)	F(1,12)=1.42 p=.257	F(1,12)=0.82 p=.384	F(1,12)=1.17 p=.300
	$d_{RKNE,lateral}$ (cm/cm)	LOWCORE HIGHCORE	0.03 (0.01) 0.03 (0.00)	0.03 (0.01) 0.03 (0.01)	F(1,12)=0.02 p=.888	F(1,12)=0.29 p=.601	F(1,12)=3.13 p=.102
Knee Range of Motion‡	$\theta_{knee,flx}$ (degrees)	LOWCORE HIGHCORE	57.01 (6.06) 53.56 (7.51)	55.29 (5.17) 53.01 (8.13)	F(1,12)=7.06 p=.021	F(1,12)=0.64 p=.439	F(1,12)=1.87 p=.197
	$\theta_{knee,var}$ (degrees)	LOWCORE HIGHCORE	8.69 (2.94) 5.96 (1.62)	7.65 (1.77) 5.44 (2.15)	F(1,12)=0.96 p=.346	F(1,12)=6.26 p=.028	F(1,12)=0.10 p=.752

*Displacement measures are normalized to participant height (cm).
†Significant multivariate main effect for condition [F(4,9) = 7.88, p = .005] and condition x group interaction [F(4,9) = 8.11, p = .005] for hip displacement variables.
‡Significant multivariate main effect for condition [F(2,11) = 5.34, p = .024] and group [F(2,11) = 7.85, p = .008] for knee range of motion variables.

tests revealed significant core engagement effects on knee angular range of motion, but not displacement. The CORE condition, compared to NOCORE, was characterized by larger knee flexion range of motion, as measured by $\theta_{knee,flx}$ [t(13) = 3.08, p = 0.009].

Effects of Core Activation Score and Core Engagement on Hip and Knee Kinematics

The MANOVA (Table 4) performed on hip displacements revealed a significant condition x core score interaction [Wilks' Lambda = 0.22, F(4,9) = 8.11, p = .005]. Follow-up univariate analyses revealed that significant condition x core score interactions in $d_{RHIP,lateral}$ [F(1,12) = 5.95, p = .031] and $d_{LHIP,lateral}$ [F(1,12) = 7.31, p = .019] contributed to the multivariate interaction. Subsequent pairwise comparisons revealed no significant group differences for core score in $d_{RHIP,lateral}$ within CORE (p = .167) and NOCORE (p = .772) conditions, or in $d_{LHIP,lateral}$ within CORE (p = .211) and NOCORE (p = .643) conditions. Pairwise comparisons revealed significant CORE-related decreases in $d_{RHIP,lateral}$ and $d_{LHIP,lateral}$ within the LOWCORE group (p = .001 and p = .001, respectively), but not within the HIGHCORE group (p = .865 and p = .966, respectively). Although not contributing to the multivariate interaction, $d_{RHIP,vertical}$ and $d_{LHIP,vertical}$ were observed to have significant CORE-related increases within the LOWCORE group (p = .020 and p = .011, respectively), but not within the HIGHCORE group (p = .995 and p = .904, respectively).

The MANOVA performed on knee ranges of motion did not reveal a significant condition x core score interaction [Wilks' Lambda = 0.82, F(2,11) = 1.23, p = .329]. However the MANOVA did reveal significant main effects for condition [Wilks' Lambda = 0.51, F(2,11) = 5.34, p = .024] and core score group [Wilks' Lambda = 0.41, F(2,11) = 7.85, p = .008]. Follow-up univariate analyses revealed a significant overall core-related increase in $\theta_{knee,flx}$ (p = .021), and a significant decrease in $\theta_{knee,var}$ for the HIGHCORE group, compared to the LOWCORE group (p = .028). Significant interactions and main effects for condition and core score group are depicted in Figure 2.

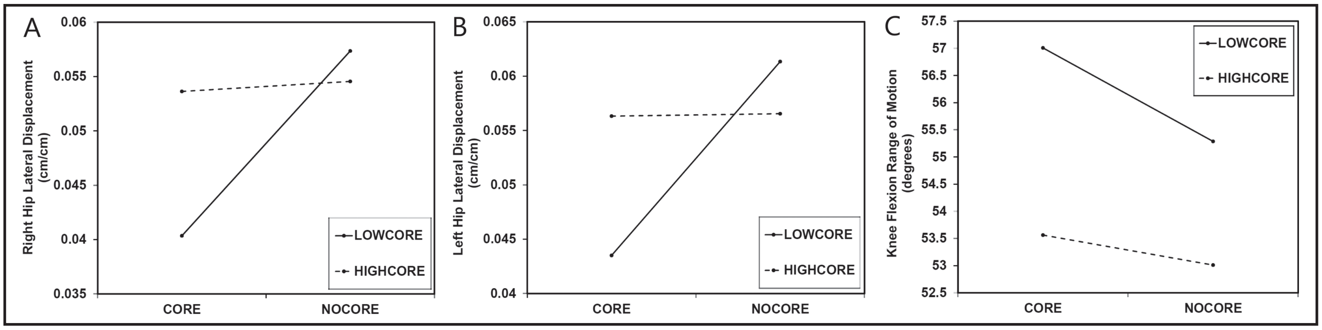

Figure 2. *Plots from 2x2 mixed model MANOVAs, comparing right (A) and left (B) hip medial-lateral displacement; and knee range of motion (C) within conditions (CORE and NOCORE) and between groups (LOWCORE and HIGHCORE). Significant differences were found in each of the comparisons depicted (p < .05).*

The MANOVAs performed on knee displacements and hip range of motion did not reveal any significant condition x core score interactions, condition main effects, or core score main effects.

DISCUSSION

The purpose of this study was to investigate the influence of intentionally activating the core musculature on lower extremity kinematics during a SLS. The findings suggest that activating the core in this way does affect hip and knee kinematics in women during a SLS. A secondary purpose was to examine the combined effects of intentional core engagement and core recruitment score on the performance measures. To achieve these goals, female subjects completed a SLS task with and without intentional core musculature engagement, during which performance measures were extracted from kinematic data captured during each trial. To study core effects alone, statistical comparisons were made between CORE and NOCORE conditions. To study the combined effect of core condition and core recruitment score, a 2 x 2 MANOVA was used.

The primary finding of the current investigation was that purposefully activating the core musculature did affect hip and knee kinematics in women during a SLS. Specifically, paired t-tests revealed core-related decreases in hip medial-lateral movement, suggesting that core activation contributes to greater frontal plane hip stability. This finding is consistent with Kaji and coauthors, who reported less medial-lateral motion in the frontal plane and smaller total excursions of center of pressure (COP) during quiet standing following core training.[33] The results of the paired t-tests from the current study also revealed core-related increases in knee flexion range of motion, suggesting that core activation positively affected lower extremity function during the SLS task; based on the premise that greater knee flexion is a measure of higher function when performing a squat task during sporting and other functional activities.[16,31,32]

These results are consistent with the idea of a feed-forward recruitment pattern of proximal to distal musculature. For example, Hodges and coworkers[18] have done a number of studies showing that trunk muscle activity, specifically the transversus abdominis and multifidus muscles, in healthy individuals precedes the muscular activity in the extremities, whereas this characteristic is absent or reversed in individuals with movement dysfunction related to low back pain.[19,34] It is believed that such a feed-forward recruitment pattern of core musculature provides a more stable neuromuscular foundation for movement to occur. Similarly, other authors have reported increased neuromuscular control at the hip as a significant contributing factor to knee kinematics.[11,35,36,37,31] Interpreting the greater knee flexion during the CORE condition as greater distal mobility, the results of the current study support this theory of proximal stability promoting greater distal mobility in the lower extremities. However, the magnitude to which engaged core musculature prepares or influences lower extremity motion is not completely understood. Although the results demonstrated evidence of such a preparatory effect, finding a causal relationship of this effect is beyond the scope of the current study. Future studies should employ EMG measurements to further explore this phenomenon. Regardless of preparatory effect however, the results of this investigation may have implications for ACL injury, as incidence of injury has been shown to decrease in athletes participating in training programs targeting core musculature.[10,36]

A secondary finding was that individuals with lower scores on the core recruitment test demonstrated the greatest differences in performance between the CORE and NOCORE conditions. Specifically, our 2x2 MANOVA revealed that the CORE condition, compared to NOCORE, was associated with larger hip vertical displacement, smaller hip medial-lateral displacement, and greater knee flexion range of motion in the LOWCORE group, but not the HIGHCORE group. This finding illustrates that those with the poorest ability to activate core musculature exhibited the greatest change in lower extremity kinematics when the core was engaged, suggesting that LOWCORE participants had the most to gain from core activation. Given evidence that ACL injury risk is associated with low core performance ability,[38] our results suggest that core training may have a greater impact among athletes with high ACL injury risk, compared to low-risk individuals. Similarly, core stability training has been shown to result in smaller movements of the center of mass trajectory during quiet standing.[33]

The result that HIGHCORE participants did not exhibit significant differences between CORE and NOCORE conditions may suggest that HIGHCORE were not benefitting as much from a voluntary contraction as did the LOWCORE group. To the authors' knowledge, this issue has not been addressed to date in the scientific literature; however this finding may possess great practical relevance given its clinical implications. One such clinical implication is that individuals with lower core recruitment scores have greater lower extremity mobility to gain from improved proximal stability. This is intriguing given that participants in this study scored between 0/5 and 2/5 on Sahrmann's test (Table 1). While these scores are at the low end of the spectrum of this clinical measure, these scores are consistent with those seen in patients entering physical therapy for treatment of musculoskeletal pain syndromes. Consequently it is tempting to speculate that participants scoring 5/5 on the test would exhibit little or no differences between CORE and NOCORE conditions, based on the premise that they would already be activating the core to a very high level during functional activities. However, further study is needed to test this notion.

Secondary analysis also revealed overall larger knee varus range of motion in the LOWCORE group, compared to HIGHCORE. This result indicates that LOWCORE participants had reduced ML knee stability compared to HIGHCORE participants *regardless of voluntary core activation*. This suggests that, although LOWCORE participants benefited from voluntary core activation, they may still have an elevated risk of injury in comparison to the HIGHCORE group. Such a finding provides additional evidence that training emphasizing core activation during functional tasks such as the SLS may help to reduce injury risk, as noted above, however, longitudinal studies would be helpful in further substantiating the impact of core training on lower extremity function in young women.

To the best of the authors' knowledge, this study is the first to compare the effects of voluntary core engagement on lower extremity kinematics at the hip and knee during a functional task. A clinical measure of core stability was obtained for each participant using the Sahrmann test for key core muscles (namely the transversus abdominis and internal and external obliques), which has been found reliable in previous studies.[25,26,39] Other authors have investigated the influence of the hip musculature on hip and knee kinematics,[11,12,40,31] and have demonstrated that activation of hip muscles does affect lower extremity motion and is therefore a predictor of knee injury.[22,36,38] However, such studies have not identified the extent to which neuromuscular control of the trunk/core affects lower extremity kinematics. This study is an important link in demonstrating the influence of the tonic stabilizers of the core as measured by a common clinical test [18,19,41,20] on the motion of the lower extremities.[10,42,43] A logical next step would be to take the same basic approach of this study and apply it to clinical populations, and another logical next step would be to include electromyography as a means of better understanding how muscular recruitment patterns of the muscles in the axial skeleton influence lower extremity kinematics during functional tasks such as the SLS.

This investigation had a number of limitations. The methods used were aggregated from procedures used in previous research, but the context was different in the present study. For example, the Sahrmann method was used as means of assessing ability to recruit core musculature. As in previous research and as in clinical practice, the authors administered this test with participants positioned in supine and it is unknown how well this assessment technique extrapolates to core activation or stabilizing ability in the standing position used during the SLS task. A related limitation is the subjective nature of the core engagement task during the SLS; no attempts were made to quantify the extent to which participants engaged the core musculature. Rather, the authors used a verbal command, a method widely used by physical therapists when teaching therapeutic exercise in clinical environments. Use of electromyography would address both of these limitations by providing: 1) a comparison between core engagement during standing, in comparison to the supine position used during the Sahrmann test; and 2) an objective assessment of participants ability to intentionally engage the core musculature. Still, the authors believe the current technique has merit since it is similar to what is commonly used in a clinical setting, as well as likely what will be used should practicing clinicians choose to integrate these or other findings into clinical practice.

Another limitation may have been the uniform step height for all participants. While this method is consistent with previous studies using the SLS,[44] joint angles during the SLS may have been influenced by differences in participant anthropometrics. To minimize this effect, all data were normalized for height. The narrow age range of the participants also presented a challenge, limiting the capacity to generalize the findings to a broader population. Similarly, while statistically significant differences were found between a number of factors the study would have been strengthened with a larger sample size. Finally, the present study represents a single assessment of intentional activating the core on lower extremity kinematics. Future studies should employ longitudinal designs to investigate time-dependent effects of core activation, particularly in those who were in the low scoring group on the clinical assessment of core stability.

CONCLUSIONS

In the present study, core activation was associated with improved stability and mobility in terms of lower extremity kinematics during a single leg squat, and intentional core activation had the greatest influence on lower extremity kinematics in individuals with lower core recruitment scores. These results suggest that, with all else being equal, individuals with lower core scores may have more to gain from increasing core stability during lower extremity movement. These findings have clinically meaningful implications for individuals, particularly women, who may be at risk for knee injury during functional activities involving the lower extremities.

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