ORIGINAL RESEARCH

THE RELATIONSHIP BETWEEN LOWER EXTREMITY CLOSED KINETIC CHAIN STRENGTH & SAGITTAL PLANE LANDING KINEMATICS IN FEMALE ATHLETES

Christopher R. Carcia, PhD, PT, SCS, OCS[1]
Ben Kivlan, MPT, SCS, OCS[1]
Jason S. Scibek, PhD, ATC[1]

ABSTRACT

Background: Female athletes continue to injure their anterior cruciate ligaments at a greater rate than males in comparable sports. During landing activities, females exhibit several different kinematic and kinetic traits when compared to their male counterparts including decreased knee flexion angles as well as decreased lower extremity (LE) strength. While open kinetic chain strength measures have not been related to landing kinematics, given the closer replication of movement patterns that occur during closed kinetic chain (CKC) activity, it is possible that lower extremity strength if measured in this fashion will be related to landing kinematics.

Purpose: To determine if unilateral isometric CKC lower extremity (LE) strength was related to sagittal plane tibiofemoral kinematics during a single leg landing task in competitive female athletes. We hypothesized females who demonstrated lesser CKC LE strength would exhibit decreased sagittal plane angles during landing.

Methods: 20 competitive female athletes (age = 16.0 ± 1.8 yrs; height = 166.5 ± 8.3 cm; weight = 59.7 ± 10.2 kg) completed CKC LE strength testing followed by 5 unilateral drop landings on the dominant LE during one test session at an outpatient physical therapy clinic. Closed kinetic chain LE strength was measured on a computerized leg press with an integrated load cell while sagittal plane tibiofemoral kinematics were quantified with an electrogoniometer.

Results: No significant relationships between absolute or normalized isometric CKC strength and sagittal plane landing kinematics were identified.

Conclusions: Closed kinetic chain lower extremity isometric strength tested at 25 degrees of knee flexion is not related to sagittal plane landing kinematics in adolescent competitive female athletes.

Levels of Evidence: **Analytic, Observational**

Key Words: ACL, Closed Kinetic Chain, Female, Kinematics, Strength

CORRESPONDING
Christopher R. Carcia PhD, PT, SCS, OCS
John G. Rangos, Sr. School of Health Sciences
Duquesne University, Pittsburgh, PA, USA
Email: carcia@duq.edu

---

[1] Duquesne University
Pittsburgh, PA, USA

INTRODUCTION

Anterior cruciate ligament (ACL) injury continues to occur at an alarming rate in female athletes.[1] The vast majority of ACL injuries occur during a non-contact or non-collision mechanism.[2,3] Common mechanisms of injury include cutting, pivoting, or while landing from a jump.[4,5] Many risk factors to explain the gender disparity have been proposed.[6] Arguably the most compelling sex specific biomechanical risk factor heightening injury risk is faulty lower extremity (LE) kinematics during landing. Researchers consistently report that females exhibit lesser knee flexion when landing from a jump.[7-9] The ramifications of a more extended knee posture during landing include decreased hamstring activity,[10-13] increased quadriceps activity,[10] increased anterior tibial translation[11-12,14-15] and increased vertical ground reaction forces.[9,16] Furthermore, less total knee joint excursion occurs throughout the maneuver resulting in a shorter time period for the dissipation of loads at the joint.[17] Each of these alterations is thought to heighten ACL injury risk.[15,18-21] The reasons females exhibit lesser knee flexion angles during landing is unclear. Investigators have identified that female athletes are more quadriceps dominant when compared to their male counterparts.[6] Given the action of the quadriceps, dominance of this muscle group would encourage a more extended knee posture during landing. Likewise, it is plausible that female athletes rely more on bony geometry and passive capsuloligamentous restraints to compensate for deficiencies in LE strength.

Traditional LE strength measures quantified in an open kinetic chain (OKC) have been poorly associated with landing kinematics.[22-24] The ability to generate strength in a closed kinetic chain (CKC), similar to a squat maneuver, more closely represents LE kinematics during a landing task compared to OKC strength measures. Hence strength quantified in a CKC may be related to landing kinematics in female athletes. Other authors have identified significant relationships between CKC strength and performance on functional tasks including a single leg hop, vertical jump and a speed/agility test.[25] The authors of the current study, however, were unable to identify previous work specifically exploring the relationship between CKC strength and landing kinematics. Therefore, the purpose of this study was to determine if LE strength as measured isometrically in a CKC via a squat maneuver was related to sagittal plane knee kinematics during a landing task in competitive female athletes. We hypothesized that females who demonstrated lesser LE CKC strength would exhibit decreased sagittal plane angles during landing.

METHODS

Twenty competitive female athletes (age = 16.0 ± 1.8 yrs; height = 166.5 ± 8.3 cm; weight = 59.7 ± 10.2 kg) from the sports of soccer (n = 16), basketball (n = 2) and volleyball (n = 2) were recruited for participation in this University IRB approved study. All enrolled soccer players competed at the cup level signifying elite status for their age. Inclusion criteria consisted of no history of surgery or a LE injury on the dominant side within the last six months which necessitated the use of crutches for more than one day. Subjects were excluded from the study if they were ACL deficient, had undergone ACL reconstruction, or had previously suffered other significant LE trauma (e.g. fracture, patellar dislocation, torn meniscus). Likewise, subjects were excluded if they were unable to perform the drop landing task or CKC strength assessment without pain. None of the subjects in this study wore foot orthotics on a daily basis or for participation in sport activity.

Identical procedures including informed consent were followed for each participant. All data was collected on the dominant LE during a single session at an outpatient physical therapy clinic. First, each subject's height and weight were measured using a mechanical beam scale (Pelstar LLC; Alsip, IL). Then, leg dominance was determined by the LE on which the subject landed from a 40 cm wood box on two out of three trials.[26] Next, the primary investigator performed a physical examination of the subject's dominant lower extremity. The primary investigator has 20 years of experience in an orthopedic, sports medicine rehabilitation setting and has attained ABPTS certification in both the areas of orthopedic and sports physical therapy. The exam consisted of inspection, range of motion assessment, manual muscle testing of the quadriceps and hamstrings and tests for ligamentous and patellar instability. All subjects demonstrated no swelling, full range of motion with normal end feels, strong and painless manual muscle tests, and negative ligamentous tests including the Lachmann and patellar apprehension tests. After the screening, subjects donned their sneakers and performed a 5 minute warm-up on a recumbent bicycle (Ufit Technology; Woodinville, WA) at a self-selected pace. Next, subjects performed practice trials of the drop landing task. The subject stepped onto a box 40 cm in height, placed hands on iliac crests, feet shoulder width apart and toes aligned just over the edge of the platform. The subject was then instructed to lean forward, fall off the platform, and land on her dominant lower extremity. Placing hands on iliac crests minimized the influence of upper extremity motion on landing kinematics. Subjects were instructed to maintain single limb balance until cued by the investigator to place the opposite LE on the floor. Each subject was allowed as many practice trials as necessary to become adept at the task. Subjects usually required only two to three trials to become comfortable with the task. Next, an electrogoniometer (EG) (Model SG-150; Biometrics Ltd; Gwent, UK) was centered over the lateral joint line of the tibiofemoral joint. The proximal block of the EG was aligned with the greater trochanter and the distal block with the lateral malleolus. The EG was secured to the lateral aspect of the knee using double sided tape (Scotch; 3M, St. Paul, MN) and subsequently circumferentially wrapped with 2¾" pre-wrap (Cramer Products Inc.; Gardner, KS). Once the EG was secured, the subject was asked to stand with her knee straight (0°) at which time the EG was zeroed in the sagittal plane. Verification of the zero degree reference angle was made both visually and goniometrically by the same investigator. Subjects were then asked to bend and straighten their knee so that data from the EG could be visually confirmed on a laptop personal computer (PC) (Dell; Austin, TX).

Figure 1. *Subject in standardized position on computerized leg press.*

Subjects were then visually and verbally oriented to the computerized leg press (CDM Sport; Fort Worth, TX). To further familiarize the subject with the leg press and control for a learning effect and potentially reduce fatigue, subjects were asked to use their non-dominant LE to push against the foot plate of the leg press for two sub-maximal repetitions. Once familiar with its operation, subjects assumed a standardized position on the leg press (Figure 1). Specifically foot position was adjusted so the subject's tibial crest was horizontal to the floor and the hip was placed in neutral alignment with respect to the frontal plane. The sled of the leg press was adjusted so the knee was flexed to 25°. Twenty five degrees of knee flexion was chosen to quantify lower extremity strength as this is representative of the angle at which many females land from a jump[27] and the approximate angle at which ACL injuries occur.[28] Upper extremity position was standardized by having each subject grasp the handles on the leg press. Once the subject was properly aligned, the leg press was locked. Locking the leg press ensured the subject maintained the standardized position during testing. Subjects pushed with their foot against the plate of the leg press five times. Each repetition was held for 5 seconds with a 25 second rest between each repetition thereby providing a 1:5 work rest ratio. Subjects performed the first and second repetition at a perceived effort of 50 and 75% respectively. The remaining three repetitions were each performed at 100% effort. Dependent measures derived from the three repetitions performed at 100% included maximal force and average force. Force was sensed by a load cell that was integrated into the leg press by the manufacturer. The load cell was tested at the manufacturer for repeatability, zero balance, creep, non-linearity, and hysteresis and

Figure 2. *Subject performing unilateral landing task from 40 cm box.*

demonstrated $< \pm 0.02\%$ error for the identified variables. Forces were transmitted to a personal computer for display. After testing was completed on the leg press, subjects sat in a chair for a five minute rest period. During this time, a foot switch (Biometrics Ltd; Gwent, UK) was secured to the bottom of the subject's sneaker between the first and second metatarsal heads with pre-wrap. Prior study[29] and pilot work indicated subjects utilized a forefoot strategy when landing from a box 40 cm in height. After the rest period, subjects once again stood on the box. Subjects then completed five trials of the drop landing task with a 30 second rest period between trials (Figure 2). All subjects completed the isometric strength test and landing task in the same order. Foot switch and EG data were recorded synchronously using a commercially available software acquisition program (Run Tech; Laguna Hills, CA). Each trial was collected using a trigger sweep method of one second duration (300 milliseconds [ms] pre-contact and 700 ms post-contact) initiated by the foot switch. The EG sampled at a rate of 2000 hertz and has been shown to be highly reliable ($ICC_{[3,k]} = 0.995$) under test-retest conditions and valid ($ICC_{[3,k]} = 0.991$) when compared to three-dimensional motion analysis when assessing sagittal plane knee kinematics during landing activities.[30] A published systematic review of the literature for use of the EG to quantify tibiofemoral kinematics further substantiated these findings.[31] Upon landing, subjects were asked to 'stick the landing' and maintain single limb balance for one second after each trial. Failure to meet these criteria resulted in negation and a repeat of the trial. After the fifth successful trial, the instrumentation was removed from the subject and the experiment was complete. With only a few exceptions subjects successfully completed each trial. Of these exceptions, subjects needed to complete only an additional one to two trials for a total of five acceptable trials.

Data reduction & statistical analysis

For the leg press, a mean from the last three repetitions for each variable (maximum force, and average force) was used for data analysis. Two additional variables were created by normalizing force to body weight (maximum force/body weight; average force/body weight). Kinematic trials were signal averaged using the software acquisition system. Once signal averaged, an event buffer 100 ms in duration starting at initial contact was established for each subject. A maximum of 100 ms was established as ACL injury has been reported to occur within this window.[4] Within this temporal envelope, sagittal plane knee flexion angles at initial contact and 100 ms were identified (Figure 3). Additionally, the rate of excursion (degrees/second) was calculated by dividing knee excursion by the time difference between knee flexion at 100 ms and initial contact:

$$\frac{\text{Knee Flexion}_{(100\text{ ms})} - \text{Knee Flexion}_{(\text{Initial Contact})}}{\text{Time}_{(100\text{ ms})} - \text{Time}_{(\text{Initial Contact})}}$$

Pearson correlation coefficients were calculated with a statistical software package (SPSS Version 17.0; Chicago, IL). Alpha levels were established a-priori at $P < 0.05$.

RESULTS

Descriptive measures of force and kinematic data appear in Table 1. Pearson correlation coefficients exhibiting the relationship between force and kinematic data appear in Table 2. No significant relationships between maximum force, average force or force normalized to body weight and kinematic data were evident.

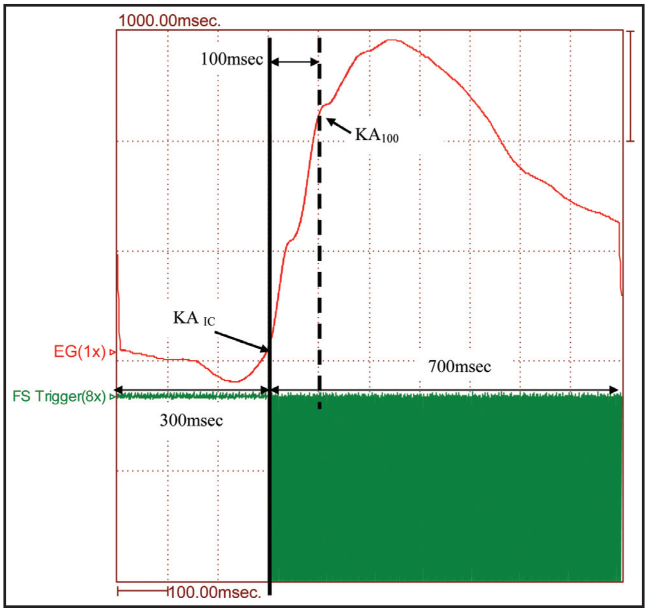

Figure 3. *Illustration of sagittal plane knee motion as measured by the electrogoniometer (EG - red line) and the foot switch (FS Trigger - green line) during a landing trial.* KA_{IC} = *Sagittal plane knee angle at initial contact;* KA_{100} = *Sagittal plane knee angle 100 ms after initial contact.*

DISCUSSION

The primary finding of our study was absolute isometric force and isometric force normalized to body weight quantified in a CKC with the knee in slight flexion were not related to sagittal plane knee kinematics during landing in adolescent female athletes. We theorized that females with lesser CKC sagittal plane LE strength would exhibit lesser knee flexion angles in an attempt to compensate for muscular deficiencies by relying on bony geometry and passive ligamentous supports. While some authors have identified a relationship between isokinetic and isometric LE strength and frontal plane knee angles during a single leg squat[32] or step down maneuver[33] respectively, our findings are in agreement with other investigators who have failed to identify an association between LE strength and sagittal plane landing kinematics.[22-24] Shultz et al examined several neuromuscular thigh parameters including OKC isometric quadriceps and hamstrings strength and electromyographic (EMG) activity of the quadriceps and hamstrings to determine if these variables were predictive of sagittal plane landing kinematics.[23] The investigators however reported very little of the variance during a landing task was explained by the predictor variables of strength and EMG. The biggest difference between this work and the current study

Table 1. *Descriptive Statistics for force and kinematic data.*

Descriptive Statistics	Minimum	Maximum	Mean (SD)
Average Force (N)	362.00	1149.00	628.97 (220.79)
Normalized Average Force (N/kg)	7.03	14.57	10.34 (2.39)
Peak Force (N)	406.00	1324.00	715.79 (254.39)
Normalized Peak Force (N/kg)	8.18	16.64	11.76 (2.72)
Sagittal Plane Knee Angle @ Initial Contact (degrees)	10.77	37.91	23.65 (7.54)
Sagittal Plane Knee Angle @ 100 ms (degrees)	42.31	66.81	53.14 (6.82)
Rate of Excursion (degrees/second)	220.00	470.00	305.50 (58.80)

Table 2. *Pearson correlation coefficients between force and kinematic data.*

	KA_{IC}	KA_{100}	Rate of Excursion
Peak Force	r = -0.14 P = 0.53	r = -0.07 P = 0.76	r = 0.17 P = 0.45
Average Force	r = -0.13 P = 0.58	r = -0.06 P = 0.79	r = 0.16 P = 0.48
Normalized Peak Force	r = -0.03 P = 0.89	r = -0.02 P = 0.92	r = 0.14 P = 0.55
Normalized Average Force	r = -0.004 P = 0.98	r = -0.005 P = 0.98	r = 0.11 P = 0.63

was that Shultz et al quantified isometric strength in an OKC. While an OKC assessment of strength allows for an isolated assessment of muscle groups, muscle groups operate synergistically during tasks such as landing. Though a significant relationship was not apparent, the authors of the current study theorized that if strength were tested in a manner that more closely represented the functional task of interest, a relationship between these two variables would be elucidated. In other work, Mizner et al reported that muscle strength was not predictive of alterations in lower extremity landing mechanics following an instructional session intended to improve landing kinematics and kinetics.[24] Likewise, despite observable increases in force production, Herman et al. noted no change in LE kinematics in female athletes during a stop-jump task following nine weeks of LE strengthening.[34] Collectively, the results of the current study and current literature suggest factors other than strength (e.g. kinematics at other joints or neuromuscular control) may explain the LE kinematic patterns observed in female athletes during landing.

Trunk kinematics have been associated with sagittal plane hip and knee kinematics. Specifically, increased trunk flexion angles are related to increased hip and knee flexion angles during landing.[35] As females tend to demonstrate a more erect trunk posture during landing[36] it is possible that trunk position encourages a decreased sagittal plane knee angle during landing. Why female athletes may adopt a more erect trunk position during landing when compared to males is unknown. Given the function of the gluteus maximus as a trunk stabilizer and decelerator of femoral medial rotation, decreased power of this core muscle may partially explain not only a more erect trunk position but also the characteristic collapse of the femurs into medial rotation following landing. Supporting this premise, anecdotally, it is recognized that individuals with substantial gluteus maximus weakness commonly exhibit a gait deviation known as the 'gluteus maximus lurch'. The gluteus maximus lurch occurs when the trunk quickly moves into excessive extension at heel strike. This gait deviation moves the center of mass posterior to the hip joint thereby minimizing the need for hip extensor activation. Other work has identified females exhibit less gluteus maximus muscle activity when compared to males during a landing task.[37] Additional studies comparing gluteus maximus power production between genders and its relationship to landing kinematics is necessary.

As suggested, the ability to generate force rapidly (muscular power) may be related to tibiofemoral landing kinematics. The authors were unable to identify any studies that have examined this notion. It is unknown if time to peak torque of the hip or knee musculature as investigated either in a closed or open kinetic chain is related to sagittal plane landing mechanics. If related, this may serve as a feasible screening tool to identify female athletes who would benefit from an intervention program aimed at improving muscular power and hence LE landing kinematics. Additional study in this regard is required.

From a statistical perspective, in order to identify a significant relationship between two variables, the data must demonstrate a sufficient range of values. Without enough variation, the data will not adequately identify the actual degree of relationship present between the two variables.[38] In other words, with a limited range or distribution of data, the correlation (r values) will underestimate the true relationship. To avoid this error, the authors examined data both numerically and graphically. As noted by the ranges and standard deviations in Table 1, the authors believe that the data demonstrated sufficient variability such that if a significant relationship between these two variables was present, it would have been detected.

The sagittal plane kinematic data observed in this study are comparable to those reported by other investigators.[27] The authors identified that females displayed an average of 23.6° at initial contact. Similarly, Decker et al. reported a mean initial contact angle of 22.8° in a group of competitive female athletes engaged in the sports of volleyball and basketball during a landing task.[27] The mean knee flexion angle of 53.1° at 100 ms found in the current study, however, was substantially less than the peak knee flexion angle of 98.4° reported by Decker and colleagues. The dissimilarity is partially explained by the fact that the current investigation does not report peak knee flexion angles but rather the knee flexion angle that occurred at 100 ms after initial contact. Peak knee flexion occurred after 100 ms for every subject in the current study. Dissimilarities may also be due to a difference in platform heights. The athletes in the study by Decker et al jumped from a 60 cm platform which was 33% higher than the platform used in the current study. Other studies that have examined female athlete lower extremity kinematics during a unilateral land from a similar height have reported comparable peak sagittal plane knee angles.[39-40]

LIMITATIONS

The data acquired in this study is limited to the population studied, angle and manner of strength testing as well as to the type of landing task performed. Given the temporal window during which ACL injury occurs, it could be contended that quantifying maximal force over a five second duration is incongruous. Further, it should be appreciated that following ground contact, what occurs is an eccentric, not isometric, contraction of the lower extremity musculature. Perhaps sagittal plane landing kinematics would be related to a CKC eccentric measure of force within an arc of motion that is comparable to that experienced by female athletes during landing. The unilateral drop landing task employed in this study is arguably not representative of what occurs on the field of play for many athletes. Athletes rarely if ever land and remain stationary. Whether the results would have been the same or not during a countermovement or drop vertical jump is unknown and requires further study. Though a CKC assessment of muscle strength may be more representative of the manner in which muscle groups work together during function, it is less clear as to the relative contribution of each muscle group as it relates to the total force production during the strength test. As the methods of this study did not include the acquisition of electromyographic data, it cannot be stated with certainty which muscle groups were specifically tested with the CKC strength test. Based on the strength measure, it was the assumption of the authors that the muscle groups primarily responsible for producing the force included the single joint hip extensors, knee extensors, and ankle plantarflexors. Likewise, it is recognized that other muscle groups worked synergistically throughout the kinematic chain to decelerate as well as provide stabilization in the frontal and transverse planes during the landing task.

CONCLUSIONS

Unilateral lower extremity closed kinetic chain isometric strength (absolute or normalized to body weight) tested at 25° of knee flexion is not related to sagittal plane knee kinematics during a single-limb drop landing in adolescent female athletes.

REFERENCES

1. Agel J, Arendt EA, Bershadsky B: Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med.* 2005;33:524-530.
2. Faunø P, Wulff Jakobsen B. Mechanism of anterior cruciate ligament injuries in soccer. *Int J Sports Med.* 2006;27:75-9.
3. Olsen OE, Myklebust G, Engebretsen L, et al. Injury mechanisms for anterior cruciate ligament injuries

in team handball: a systematic video analysis. *Am J Sports Med.* 2004;32:1002-1012.

4. Krosshaug T, Nakamae A, Boden BP, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med.* 2007;35:359-367.
5. Delfico AJ, Garrett WE. Mechanisms of injury of the anterior cruciate ligament in soccer players. *Clin Sports Med.* 1998;17:779-785.
6. Griffin LY, Agel J, Albohm MJ, et al. Noncontact anterior cruciate ligament injuries: Risk factors and prevention strategies. *J Am Acad Orthop Surg.* 2000;8:141-150.
7. Chappell JD, Creighton RA, Giuliani C, et al. Kinematics and electromyography of landing preparation in vertical stop-jump: Risks for noncontact anterior cruciate ligament injury. *Am J Sports Med.* 2007;35:235-241.
8. McLean SG, Lipfert SW, van den Bogert AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Med Sci Sport Exer*. 2004;36:1008-1016.
9. Hewett TE, Stroupe AL, Nance TA, et al. Plyometric training in female athletes – decreased impact forces and increased hamstring torques. *Am J Sports Med.* 1996;24:765-773.
10. Colby S, Francisco A, Yu B, et al. Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury. *Am J Sports Med.* 2000;28:234-240.
11. Li G, Rudy TW, Sakane M, et al. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* 1999;32:395-400.
12. Pandy MG, Shelburne KB. Dependence of cruciate-ligament loading on muscle forces and external load. *J Biomech.* 1997;30:1015-1024.
13. Baratta R, Solomonow M, Zhou BH, et al. Muscular coactivation. The role of the antagonist musculature in maintaining knee stability. *Am J Sports Med.*1988;16:113-22.
14. More RC, Karras BT, Neiman R, et al. Hamstrings – an anterior cruciate ligament protagonist. An in vitro study. *Am J Sports Med.* 1993;21:231-237.
15. Renstrom P, Arms SW, Stanwyck TS, et al. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. *Am J Sports Med.* 1986;14:83-87.
16. DeVita P, Skelly WA. Effect of landing stiffness on joint kinematics and energetics in the lower extremity. *Med Sci Sport Exerc.* 1992;24:108-115.
17. Lephart SM, Abt JP, Ferris CM. Neuromuscular contributions to anterior cruciate ligament injuries in females. *Curr Opin Rheumatol.* 2002;14:168-173.
18. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus laoding of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* 2005;33:492-501.
19. DeMorat G, Weinhold P, Blackburn T, et al. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med*. 2004;32:477-483.
20. Ferretti A, Papandrea P, Contenduca F, et al. Knee ligament injuries in volleyball players. *Am J Sports Med.*1992;20:203-207.
21. Solomonow M, Baratta R, Zhou BH, et al. The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *Am J Sports Med.* 1987;5:207-213.
22. Sigward SM, Cheng W, Souza RB, et al. The relationship between dynamic knee valgus and hip structure, range of motion and strength during a landing and cutting. *J Orthop Sports Phys Ther.* 2010;40:A48-A49.
23. Shultz SJ, Nguyen A-D, Leonard MD, et al. Thigh strength and activation as predictors of knee biomechanics during a drop jump task. *Med Sci Sports Exerc.* 2009;41:857-866.
24. Mizner RL, Kawaguchi JK, Chmielewski TL. Muscle strength in the lower extremity does not predict post-instruction improvements in the landing patterns of female athletes. *J Orthop Sports Phys* Ther. 2008;38:353-361.
25. Negrete R, Brophy J. The relationship between isokinetic open and closed chain lower extremity strength and functional performance. *J Sport RehabiL* 2000;9:46-61.
26. Padua DA, Arnold BL, Carcia CR, et al. Gender differences in leg stiffness and stiffness recruitment strategy during two-legged hopping. *J Mot Behav.* 2005;37:111-125.
27. Decker MJ, Torry MR, Wyland DJ, et al. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech.* 2003;18:662-669.
28. Boden BP, Dean GS, Feagin JA, et al. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* 2000;23:573-578.
29. Lindenberg KM, Carcia CR, Martin RL, et al. The influence of heel height on sagittal plane knee kinematics during landing in females. *J Orthop Sports Phys Ther.* 2010;40:A49.

30. Bronner S, Agraharasamakulam S, Ojofeitimi S. Reliability and validity of electrogoniometry measurement of lower extremity movement. *J Med Eng Technol.* 2010;34:232-242.
31. Piriyaprasarth P, Morris ME. Psychometric properties of measurement tools for quantifying knee joint position and movement: a systematic review. *The Knee.* 2007;14:2-8.
32. Clairborne TL, Armstrong CW, Gandhi V, et al. Relationship between hip and knee strength and knee valgus during a single leg squat. *J Appl Biomech.* 2006;22:41-50.
33. Hollman JH, Ginos BE, Kozuchowski J, et al. Relationships between knee valgus, hip-muscle strength and hip-muscle recruitment during a single-limb step-down. *J Sport Rehabil.* 2009;18:104-117.
34. Herman DC, Weinhold PS, Guskiewicz KM, et al. The effects of strength training on the lower extremity biomechanics of female recreational athletes during a stop-jump task. *Am J Sports Med.* 2008;36:733-740.
35. Blackburn JT. Padua DA, Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clin Biomech*. 2008; 23:313-319.
36. Ireland ML. Anterior cruciate ligament injury in female athletes: epidemiology. *J Athl Train.* 1999;34:150-154.
37. Zazulak BT, Ponce PL, Straub SJ, et al. Gender comparison of hip muscle activity during single-leg landing. *J Orthop Sports Phys Ther.* 2005;35:292-299.
38. Portney LG, Watkins MP. Foundations of clinical research – applications to practice. 3rd ed. Upper Saddle River, NJ: Prentice-Hall, Inc.; 2009.
39. Orishimo KF, Kremenic IJ, Pappas E, et al. Comparison of landing biomechanics between male and female professional dancers. *Am J Sports Med.* 2009;37:2187-2193.
40. Nagano Y, Hirofumi I, Ikai M, et al. Gender differences in knee kinematics and muscle activity during a single limb drop landing. *Knee.* 2007;14:218-223.