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Hip-Abductor Fatigue, Frontal-Plane Landing Angle, and Excursion During a Drop Jump

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Context: The influence of hip-muscle function on knee-joint kinematics during landing has been inadequately investigated. ***Objective:*** To determine the effect of bilateral hip-abductor fatigue on frontal-plane tibiofemoral landing characteristics and vertical ground-reaction force (vGRF) during the landing phase of a drop jump. ***Design:*** Experimental, pretest–posttest. ***Setting:*** Research laboratory. ***Participants:*** 20 recreationally active college-age students. ***Intervention:*** Isometric bilateral hip-abductor-fatigue protocol. ***Main Outcome Measures:*** Frontal-plane tibiofemoral landing angle, excursion, and vGRF during landing from a drop jump under prefatigue, postfatigue, and recovery conditions. ***Results:*** After the fatigue protocol, participants landed in a greater valgus orientation than in the prefatigued state. No differences in frontal-plane excursion or vGRF were noted. ***Conclusions:*** Isolated bilateral hip-abductor fatigue alters frontal-plane lower extremity orientation during a double-leg landing. Because an increase in valgus orientation has been observed at or near the time of noncontact anterior cruciate ligament injuries, we recommend improving hip-abductor muscle performance to lessen the risk of such injuries. ***Key Words:*** anterior cruciate ligament, knee, valgus

Noncontact anterior cruciate ligament (ACL) injuries are commonly associated with deceleration activities such as changing direction[1] or landing from a jump.[2,3] The precise mechanism remains unclear, although has been observed to coincide with maximum tibiofemoral valgus.[4,5] Tibiofemoral valgus results in the weight-bearing lower extremity when the hip is adducted and internally rotated while the tibia is externally rotated and the foot pronated.[6] It is not surprising that increased tibiofemoral varus or valgus moments and vertical ground-reaction forces (vGRF) during landing have been identified as ACL-injury risk factors.[7,8] Improvements in global lower extremity neuromuscular control achieved through training programs have shown promise with decreasing tibiofemoral moments, impact forces, and injury rates.[7,9] The specific training responses that are responsible for altering these risk factors and injury rates are unclear, though increased hamstring torques have

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been documented.[9] Although the hamstrings, because of their angle of pull, are ideally situated to reduce anterior tibial translation between 15° and 80° of knee flexion,[10] they are not as optimally positioned to resist external varus and valgus moments at the tibiofemoral joint. Supporting this premise, the hamstrings have been shown to elicit substantial EMG activity to reduce anterior tibial translation but minimal activity during varus or valgus loading when the knee is flexed.[11,12] Other muscle groups must therefore be responsible for controlling tibiofemoral motion in the frontal plane.

Contemporary theory suggests that the hip musculature plays a critical role in controlling multiplanar femoral motion during sport activities.[4] In fact, videotape analysis has revealed that just before or at the time of ACL injury, the hip is commonly in a position of adduction and internal rotation.[13] Attainment of this position during landing suggests insufficient activity of the hip abductor and external rotators.[14] Weakness or fatigue of the hip abductors or external rotators might contribute to the observed altered lower extremity landing kinematics and further increase injury risk.[15,16] The effect of hip-abductor fatigue on frontal-plane tibiofemoral characteristics and vGRF during landing, however, has not been reported. Therefore, the purpose of our study was to determine the effect of bilateral hip-abductor fatigue on frontal-plane tibiofemoral characteristics including landing angle, excursion, and vGRF during the landing phase of a drop jump. We hypothesized that after fatigue, landing angle would be in more of a valgus orientation and frontal-plane excursion and vGRF would decrease when compared with baseline levels.

Methods

Experimental Design and Setting

A pretest–posttest design was used to examine the effects of bilateral hip-abductor fatigue on frontal-plane tibiofemoral characteristics during the landing phase of a drop jump. All data were collected during 1 test session at the university sports-medicine/athletic training research laboratory.

Participants

Twenty recreationally active college-age students (10 women, 10 men; age 24 ± 2.8 years, height 174.2 ± 7.9 cm, weight 70.9 ± 12.7 kg) volunteered for the study. Inclusion criteria were currently engaging in exercise 3 or more hours per week, being free from unresolved lower extremity injury over the preceding 6 months, having no known medical conditions preventing the ability to perform a drop jump, and being otherwise healthy. Before the commencement of data collection, all participants read and signed an informed-consent form approved by the university's investigative review board for the protection of human subjects.

Instrumentation

We used a wooden box, 0.3 m in height, to establish a fixed height for the drop jump; 2 twin-axis electrogoniometers (Biometrics Ltd, Gwent, UK) to measure frontal-plane tibiofemoral landing angle and excursion; and a force plate (Bertec Corp, Columbus, Ohio) to quantify vGRF (1000 Hz). We used a Kin-Com II isokinetic dynamometer (Chattecx Inc, Chattanooga, Tenn) to establish baseline force of the hip abductors and administer the fatigue protocol. The electrogoniometers, force plate, and KinCom II were interfaced with a personal computer (Compaq Corp, Houston, Tex) and DATAPAC Version 2.42 lab acquisition and analysis software (RUN Technologies, Laguna Hills, Calif) via an A/D board.

Procedures

After recording height, weight, and age, with the subject standing in the anatomical position we centered and secured an electrogoniometer across the lateral joint line of each knee with double-sided adhesive tape. The proximal and distal sensors of each electrogoniometer were aligned with the participant's greater trochanter and lateral malleolus, respectively. To prevent undesired movement of the electrogoniometer sensors, each was circumferentially covered by 3-in prewrap (Cramer Corp, Gardner, Neb) secured by 1.5-in athletic tape (Johnson & Johnson, Princeton, NJ). Participants then stood on the wooden box positioned 12.7 cm behind the force plate with their knees in extension and feet pointing straight ahead and 35 cm apart (distance between medial borders). Once positioned, the electrogoniometers were zeroed and participants completed baseline drop jumps. After baseline data had been collected, subjects completed maximum voluntary isometric hip-abduction contractions (MVICs) and underwent a fatigue protocol before performing 2 repeat series of drop landings—immediately postfatigue and after 2 minutes of recovery.

Baseline Drop Jumps. Participants were instructed to drop onto the force plate with both feet and then immediately perform a maximal-effort vertical jump. Participants were allowed to complete several practice trials immediately before data collection to become adept at the task. Participants then performed 3 acceptable test trials of the drop jump. Trials were disqualified and repeated if a subject exhibited a loss of balance on landing on the force plate or if movement artifact was visible on the software acquisition system. Data were collected on a trigger-sweep mode with a 250-millisecond pretrigger interval elicited by the vGRF.

MVIC and Fatigue Protocol. Next, participants stood adjacent to the Kin-Com dynamometer head with the distal edge of the resistance pad 5 cm above the lateral joint line (Figure 1). The resistance pad of the Kin-Com was fixed with a 3.8-cm-thick dense foam pad, cut with a 6- by 3-cm well to accommodate the electrogoniometer (Figure 2). Participant position was standardized with trunk erect, arms across chest, feet straight ahead and in line vertically with shoulders, and head in neutral. Once he or she had been positioned, to ensure consistent foot placement during the MVICs and fatigue protocol, we marked each participant's position on the floor with white athletic tape. Participants performed three 5-second MVICs of the right hip abductors followed by the left, with a 15-second rest interval between MVIC repetitions. The resistance arm of the dynamometer provided maximal resistance to the test leg, which was held slightly off the ground in a non-weight-bearing fashion, while the contralateral lower extremity supported the body weight and provided stabilization. Baseline maximum hip-abduction force was recorded for left and right sides by averaging the individual peak forces from

Figure 1 — Participant standing adjacent to KinCom II isokinetic dynamometer in standardized position for maximum voluntary isometric contractions and fatigue protocol.

Figure 2 — Custom-designed foam well pad used to accommodate electrogoniometer during fatigue protocol.

the 3 trials of the non-weight-bearing lower extremity as visualized from the Kin-Com II computer display.

For the fatigue protocol, participant position was standardized in the same manner as for the MVIC. All participants initiated the fatigue protocol with the right leg in a non-weight-bearing posture (pushing against the resistance pad of the Kin-Com) and the left leg in a weight-bearing posture. Participants performed maximal isometric efforts against the resistance pad for trials of 15 seconds, alternating sides by turning their body 180° after each trial. Prior work from our laboratory[17] has demonstrated that these procedures are effective for eliciting considerable bilateral (non- and weight-bearing) hip-abductor electromyographic activity simultaneously. Therefore, with the exception of the time necessary to reposition between sides (~2 seconds), the hip abductors did not rest during the fatigue protocol. The isometric mode was chosen for several reasons: (1) Fatigue of the abductors could be accomplished in a simultaneous bilateral manner, (2) the protocol incorporated both weight-bearing and non-weight-bearing conditions, (3) the abductors were fatigued in a controlled and isolated fashion, and (4) the distance subjects could move in the laboratory was limited by the cables from the electrogoniometers to the data-acquisition system. Participants were given verbal cues counting down from 3 to prompt when to alternate. When switching sides, each participant's foot position relative to the tape placed on the floor was monitored, and if necessary, participants were instructed to adjust their feet. Alternating right and left trials were continued until the participant was fatigued. We defined fatigue as the participant's inability to achieve 50% of their ipsilateral baseline force bilaterally for 2 trials.

To validate our MVIC and fatigue data as displayed by the KinCom II monitor, we simultaneously collected data from the KinCom II with DATAPAC during each trial of the MVICs and on the last trial of the fatigue protocol bilaterally.

Postfatigue and Recovery Jumps. Once fatigued, participants immediately went to the box and performed 3 drop jumps as previously described. The elapsed time between completion of the fatigue protocol and performance of the postfatigue jumps was less than 15 seconds. After performing the postfatigue drop jumps, participants were instructed to sit down in a chair for 2 minutes to recover and then perform 3 final drop jumps.

Data Reduction

For the vGRF and electrogoniometer data, event markers within DATAPAC were established to demarcate initial contact of the participant on the force plate, maximum vGRF, and maximum frontal-plane tibiofemoral angle during landing. Frontal-plane landing angle was defined as the angle recorded by the electrogoniometer at initial force-plate contact. Frontal-plane tibiofemoral excursion was defined as the difference between landing angle and maximum tibiofemoral angle achieved during the landing phase (Figure 3). Maximum vGRF (N) was normalized for each participant and expressed in multiples of his or her body weight.

Statistical Analysis

The averages of the 3 trials for each dependent measure (landing angle, excursion, maximum tibiofemoral angle, and vGRF) and condition (prefatigue, postfatigue,

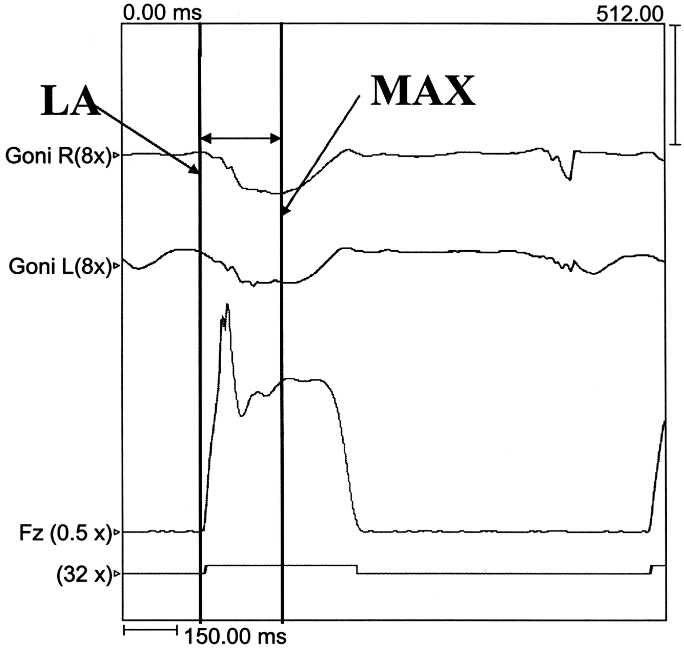

Figure 3 — Depiction of goniometric and ground-reaction-force data from acquisition software with event markers. LA indicates landing angle, and MAX, maximum frontal-plane angle.

and recovery) were used for statistical analysis. Data were entered into and analyzed with SPSS Version 10.1 (SPSS Inc, Chicago, Ill). A repeated-measures analysis of variance (ANOVA) with 2 within factors (force at 2 levels [prefatigue and post-fatigue] and side at 2 levels [right and left]) was used to evaluate the effect of the fatigue protocol on force by side. Separate repeated-measures ANOVAs with 1 within factor (drop jump) at 3 levels (prefatigue, postfatigue, and recovery) were used to analyze the effect of bilateral hip-abductor fatigue on left and right frontal-plane tibiofemoral characteristics (landing angle, maximum angle, and excursion) and vGRF after the drop jump. Alpha levels were set a priori at $P \le .05$. Where indicated, multiple comparisons with Bonferroni correction were performed to identify significant differences between means.

Results

Overall, hip-abductor force was reduced after the fatigue protocol from 120.0 ± 29.28 N to 84.0 ± 21.64 N ($P < .001$). Hip-abductor force did not differ by side ($P = .41$), and there was no side-by-force interaction ($P = .15$). The average time to fatigue was 9.5 ± 3.8 minutes. Table 1 lists the means, standard deviations, and probability (P) values for landing angle, maximum angle, and excursion across

Table 1 Frontal-Plane Tibiofemoral Measures Across the 3 Conditions, Mean ± SD and Probability Values*

	Prefatigue	Postfatigue	Recovery	*P*
LA				
right	–0.19° ± 7.9°	–0.93° ± 7.1°	–1.95° ± 7.2°	<.001†
left	–0.42° ± 4.7°	–1.24° ± 3.9°	–1.59° ± 4.7°	<.028‡
MAX				
right	–10.28° ± 12.4°	–9.63° ± 10.9°	–10.70° ± 11.4°	.286
left	–12.79° ± 9.4°	–12.86° ± 9.7°	–12.86° ± 9.2°	.996
EXC				
right	10.09° ± 7.1°	8.70° ± 6.5°	8.75° ± 6.3°	.065
left	12.37° ± 6.8°	11.62° ± 7.7°	11.27° ± 7.1°	.349

*LA indicates landing angle; MAX, maximum frontal-plane angle; and EXC, frontal-plane excursion (LA – MAX). Negative sign indicates valgus orientation.

†Indicates $LA_{recovery} > LA_{postfatigue} = LA_{prefatigue}$.

‡Bonferroni correction failed to identify differences between means.

the 3 conditions for right and left sides. No difference in vGRF was noted between prefatigue (3.64 ± 0.77 times body weight), postfatigue (3.71 ± 0.76 times body weight) and recovery (3.77 ± 0.84 times body weight) conditions (*P* = .549).

Comments

Although hip-muscle weakness or fatigue has been associated with altered lower extremity kinematics[18] and injury patterns,[4,19] to our knowledge, no research has specifically examined the influence of proximal fatigue on knee kinematics during landing. As it relates to ACL injury, a plethora of research over the last 3 decades has focused on the ability of the hamstrings to restrain anterior tibial translation.[10,11,20,21] Despite these works, the frequency of noncontact ACL injuries has not declined.[22] This might be related to the fact that ACL strain is greater when combined loads (ie, valgus and anterior tibial shear forces) are applied simultaneously.[23,24] Although the hamstrings are ideally situated to control anterior tibial translation,[10,11,20,21] they are not as well situated to control tibiofemoral varus or valgus rotations. Hence, the need to examine other more proximal neuromuscular structures is readily apparent. The primary finding of our study was that bilateral hip-abductor fatigue altered frontal-plane tibiofemoral landing angle but not excursion or vGRF during the landing phase of a drop jump.

Landing Angle

Although several studies have examined tibiofemoral kinematics during landing,[25-27] the focus of these works has been on sagittal-plane motion—frontal-plane kinematic data are seldom reported. Using similar methodology, Ford et al compared frontal-plane tibiofemoral kinematics between genders.[28] At initial contact, both males and females landed with their knees in slight valgus, 3.3° and 5.9° ($P > .05$), respectively. In our study, the mean for the prefatigue condition was close to a neutral varus/valgus position. Although the mean differences between studies seem small, the context of these differences must be considered. Because the quantity of frontal-plane tibiofemoral motion is substantially less than sagittal-plane tibiofemoral motion, a 3° to 5° change in the frontal plane represents a more sizeable percentage than a 3° to 5° change in the sagittal plane. Using this framework, our values were notably smaller than those reported by Ford et al. These differences, however, might be partially explained by differences in methods. We zeroed the electrogoniometers after they were applied to each participant. Therefore, if the participant had a natural predilection for valgus lower extremity alignment, zeroing the goniometer at this time would have negated this predilection. Ford et al used an optical marker system to capture lower extremity kinematic data. If the participant had a natural predilection for varus or valgus alignment, it is possible that this tendency was captured by the motion-analysis system.

After fatigue, participants in our study landed with their knees in more of a valgus orientation bilaterally than in the nonfatigued condition. These findings suggest that the hip abductors might have allowed a greater amount of hip adduction to occur, thereby facilitating a more pronounced valgus orientation of the tibiofemoral joint during landing. Although the differences in frontal-plane tibiofemoral motion were relatively small (<2°), the clinical importance might be considerable. Because the tibiofemoral joint is located at the juncture of 2 substantial bony levers, small changes in varus or valgus alignment might translate into large changes in moments about the joint. Further study is necessary to quantify the effect of hip-abductor fatigue on varus and valgus moments about the tibiofemoral joint.

Tibiofemoral landing angle at initial contact bilaterally was greatest during the recovery jumps rather than immediately after fatigue. The reasons for this phenomenon are unclear. One possible explanation might be related to the accumulation of metabolites, including lactic acid, produced by the working muscles during the fatigue protocol. Lactic acid impairs a muscle's ability to produce force.[29] Furthermore, it has been observed that blood lactate levels do not peak until several minutes after the cessation of heavy exercise.[30] Therefore, it is possible that concentrations of lactic acid after the 2-minute rest period were actually higher than those immediately after the fatigue protocol. If this is true, the elevated concentration of lactic acid might have impaired the ability of the hip abductors to contract and hence control the lower extremity during the landing task. Further study is necessary to clarify this finding, as well as to determine what time frame is necessary after fatigue before kinematics return to baseline.

Excursion

Frontal-plane tibiofemoral excursion for all participants proceeded in a valgus direction on landing. Although frontal-plane tibiofemoral excursion did not significantly differ between conditions, a trend toward less excursion after fatigue was apparent on the right side ($P = .065$). We interpret this to signify that because the knee landed in a greater valgus position, lesser excursion was available before end-range valgus was attained. If so, this might have implications as it relates to force absorption. With less range available to absorb forces, force dissipation might not be as effective, thereby increasing injury risk. This concept has been suggested by other investigators[31-32] examining ground-reaction forces and knee-flexion angles during landing tasks.

Overall, frontal-plane tibiofemoral excursion in our study was similar to the male (12.8°) but much smaller than the female (21.7°) group data reported by Ford et al.[28] Post hoc analysis of our prefatigue excursion data on the right by sex revealed that although women (12.6°) displayed greater excursion than men (7.5°), these differences were not significant (P = .112). Although the lack of statistical significance might be a function of sample size (10 per group), the mean differences do not approach the magnitude of the difference reported by Ford et al. One possible explanation is a difference in populations sampled. Ford et al studied frontal-plane tibiofemoral excursion in high school athletes, and we gathered data from a college population. It is plausible that our women were more physically mature and exhibited less tibiofemoral valgus during dynamic activity because of enhanced neuromuscular control.[9] In fact, 4 of our female participants were off-season NCAA Division I field-hockey athletes. Further study is necessary to clarify this discrepancy.

Vertical Ground-Reaction Force

Despite fatigue of the hip abductors bilaterally, vGRF was not affected. These data suggest that other compensatory strategies[32] (ie, sagittal hip, knee, and ankle excursions) are more critical for absorbing vGRF. Because hip-abductor fatigue influenced frontal-plane tibiofemoral landing angle, however, frontal-plane ground-reaction forces and moments might be more likely to be affected by hip-abductor fatigue because of the orientation of the limb at ground contact.

Clinical Implications

Based on the results of this study, it appears that the proximal hip musculature influences kinematics at the tibiofemoral joint. Alterations in landing kinematics might increase injury risk. Therefore, we recommend incorporating exercise aimed at improving hip-muscle performance for injured and healthy athletes who are engaged in sports in which they often have to land from a jump. Improving hip-muscle performance might help preserve normal kinematics at the tibiofemoral joint, thereby lowering injury risk.

In addition, because hip-abductor fatigue likely does not occur as an isolated entity but rather concomitantly with the fatigue of other muscle groups during activity, the effects within a functional environment might be more pronounced than our laboratory observations. Subsequent study should examine the effects of a global hip-fatigue protocol on frontal-plane landing characteristics at the tibiofemoral joint.

Limitations

The present results are limited to the population represented, for drop landings performed from 0.3 m and after isolated hip-abductor fatigue to the attained level.

It is worth noting that the level of fatigue achieved was actually 30% and not the 50% we seemingly attained at the time of testing. This discrepancy was the result of difficulty in reading the force output on the Kin-Com monitor at the time of testing. Because of the nature of the testing, consistent, steady pressure on the resistance arm and hence load cell was difficult for participants to maintain. This difficulty manifested itself as fluctuations in force output as indicated by the computer monitor, making precise identification of force challenging.

Conclusions

Hip-abductor fatigue influences frontal-plane tibiofemoral landing angle but not excursion or vertical ground-reaction force during the landing phase of a drop jump. Because a valgus position of the tibiofemoral joint at landing has been identified as a risk factor for ACL injury, fatigued hip abductors might influence this risk.

References

1. Noyes FR, Moor PA, Matthews DS, Butler DL. The symptomatic anterior cruciate deficient knee. *J Bone Joint Surg Am*. 1983;65:154-162.
2. Gerberich SG, Lucmann S, Finke C, Priest JD, Beard BJ. Analysis of severe injuries associated with volleyball activities. *Physician Sports Med*. 1987;15(8):75-79.
3. Ferretti A, Papandrea P, Conteduca F, Mariani PP. Knee ligament injuries in volleyball players. *Am J Sports Med*. 1992;20(2):203-207.
4. Ireland ML. The female ACL: why is it more prone to injury? *Orthop Clin North Am*. 2002;33:637-651.
5. Ebstrup JF, Bojsen-Moller F. Anterior cruciate ligament injury in indoor ball games. *Scand J Med Sci Sports*. 2000;10:114-116.
6. Ireland ML. Proprioception and neuromuscular control related to the female athlete. In: Lephart SM, Fu FH, eds. *Proprioception and Neuromuscular Control in Joint Stability*. Champaign, Ill: Human Kinetics; 2000:291-299.
7. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes—a prospective study. *Am J Sports Med*. 2005;33(4):492-501.
8. Hewett TE, Lindenfeld TN, Riccobene JV, Noyes FR. The effect of neuromuscular training on the incidence of knee injury in female athletes—a prospective study. *Am J Sports Med*. 1999;27(6):699-706.
9. Hewett TE, Stroupe AL, Nance TA, Noyes FR. Plyometric training in female athletes—decreased impact forces and increased hamstring torques. *Am J Sports Med*. 1996;24(6):765-773.
10. Hirokawa S, Solomonow Z, Luo Y, D'Ambrosia R. Muscular co-contraction and control of knee stability. *J Electromyogr Kinesiol*. 1991;1(3):199-208.
11. 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;15: 207-213.
12. Buchanan TS, Lloyd DG. Muscle activation at the human knee during isometric flexion-extension and varus-valgus loads. *J Orthop Res*. 1997;15:11-17.
13. Ireland ML. Anterior cruciate ligament injury in female athletes: epidemiology. *J Athl Train*. 1999;34(2):150-154.
14. Simoneau GG. Kinesiology of walking. In: Neumann DA, ed. *Kinesiology of the Musculoskeletal System*. St. Louis, Mo: Mosby, Inc; 2002:523-569.

15. Ireland ML, Leetun DT. Gender differences in core stability as measured by trunk and hip performance. Paper presented at: Research Retreat II—ACL Injuries: The Gender Bias; April 2003; Lexington, Ky.
16. Fredericson M, Cookingham CL, Chaudhari AM, Dowdell BC, Oestreicher N, Sahrmann SA. Hip abductor weakness in distance runners with iliotibial band syndrome. *Clin J Sports Med*. 2000;10:169-175.
17. Bauer AM, Webright WG, Arnold BA, Schmitz RJ, Gansneder BM. Comparison of weight-bearing and non-weight bearing gluteus medius EMG during isometric hip abduction [abstract]. *J Athl Train*. 1999;34:S58.
18. Hruska R. Pelvic stability influences lower extremity kinematics. *Biomechanics*. 1998;5: 23-29.
19. Nadler SF, Malanga GA, DePrince M, Stitik TP, Feinberg JH. The relationship between lower extremity injury, low back pain and hip muscle strength in male and female collegiate athletes. *Clin J Sports Med*. 2000;10:89-97.
20. More RC, Karras BT, Neiman R, Fritschy D, Woo SL-Y, Daniel D. Hamstrings—an anterior cruciate ligament protagonist—an *in-vitro* study. *Am J Sports Med*. 1993;21: 231-237.
21. MacWilliams BA, Wilson DR, DesJardins JD, Romero J, Chao EY. Hamstrings cocontraction reduces internal rotation, anterior translation and anterior cruciate ligament load in weight-bearing flexion. *J Orthop Res*. 1999;17:817-822.
22. 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.
23. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GAM, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res*. 1995;13:930-935.
24. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res*. 1992;10:167-176.
25. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Steadman JR. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech*. 2003;18:662-669.
26. Fagenbaum R, Darling WG. Jump landing strategies in male and female college athletes and the implications of such strategies for anterior cruciate ligament injury. *Am J Sports Med*. 2003;31:233-240.
27. Huston LJ, Vibert B, Ashton-Miller JA, Wojtys EM. Gender differences in knee angle when landing from a drop-jump. *Am J Knee Surg*. 2001;14:215-219.
28. Ford KR, Myer GD, Hewett TE. Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc*. 2003;35:1745-1750.
29. Spagenburg EE, Ward CW, Williams JH. Effects of lactate on force production by mouse EDL muscle: implications for the development of fatigue. *Can J Physiol Pharmacol*. 1998;76:642-648.
30. Astrand I. Aerobic work capacity in men and women with special reference to age. *Acta Physiol Scand*. 1960;49(suppl 169):1-92.
31. Lephart SM, Ferris CM, Riemann BL, Myers JB, Fu FH. Gender differences in strength and lower extremity kinematics during landing. *Clin Orthop*. 2002;401:162-169.
32. Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc*. 1992;24:108-115.