Association between femoral anteversion and lower extremity posture upon single-leg landing: implications for anterior cruciate ligament injury

J. Phys. Ther. Sci.
25: 1213–1217, 2013

Association between Femoral Anteversion and Lower Extremity Posture upon Single-leg Landing: Implications for Anterior Cruciate Ligament Injury

Masaaki Kaneko, PT, MS[1,2)*], Keishoku Sakuraba, MD, PhD[1)]

[1)] *Department of Sports Medicine, Graduate School of Medicine, Juntendo University: 2-1-1 Hongo, Bunkyo-ku, Tokyo 113-8421, Japan*
[2)] *Department of Rehabilitation, Tohoku University Hospital, Japan*

Abstract. [Purpose] Increased femoral anteversion may occur with hip internal rotation and valgus knee alignment upon landing and is considered a risk factor for anterior cruciate ligament injury. We examined the relationship between femoral anteversion and joint motion and muscle activity of the lower extremity in terms of the risk factors for anterior cruciate ligament injury. [Subjects] Sixteen healthy females were divided on the basis of femoral anteversion into low and high groups. [Methods] Femoral anteversion was assessed using Craig's test. We performed kinematic analysis and measured the electromyography activity of the lower extremity upon left single-leg landing. [Results] The high group had a significantly lower hip flexion angle and higher knee flexion and valgus angles than the low group. The rectus femoris showed significantly greater electromyography activities in the high group than in the low group. [Conclusion] These results suggest that increased femoral anteversion results in lower hip flexion angle, higher knee valgus alignment, and greater rectus femoris muscle activity, leading to anterior tibial displacement upon single-leg landing. Increased femoral anteversion may be a potential risk factor for anterior cruciate ligament injury.
Key words: Femoral anteversion, Lower extremity alignment, Single-leg landing

(This article was submitted Mar. 27, 2013, and was accepted May 10, 2013)

INTRODUCTION

Approximately 91% of anterior cruciate ligament (ACL) injuries occur during sports activities[1–4)]; most are caused by noncontact injury mechanisms, e.g., landing from a jump and rapid deceleration[5–8)]. The position of the knee in noncontact ACL injury is characterized by slight flexion (<30°), valgus of the knee, and internal rotation of the tibia[1, 9–11)]. Thus, training programs to prevent noncontact ACL injuries should consider the anatomical and neurophysiological factors for preventing slight flexion and valgus of the knee joint. Recent research has reported an association between the position of ACL injuries and hip joint kinematics[12)]. Hip and knee flexion angles upon landing are important factors determining the load on the knee, with slight flexion angles resulting in a greater load[13)]. Moreover, the decrease in hip flexion angle during rapid deceleration results in a leg position that increases the quadriceps force, leading to anterior tibial displacement and increased risk of ACL injury[14)]. These findings indicate that the shape of the femur may affect both kinematics of the hip and knee joints upon landing as well as muscle activity for protecting the joints.

Anteversion and the neck–shaft angle are morphological factors of the femur that affect the kinematics of the hip and knee joints upon landing. Femoral anteversion is the angle formed by the axis of the femoral neck and the horizontal axis of the femoral condyles. The angle of torsion decreases up to approximately 6 years of age, and the femoral head changes so that it faces more medially in the acetabulum[15)]. Thus, when femoral anteversion is larger, the femoral head faces anteriorly in the acetabulum, resulting in reduced congruity of the hip joint. Improved congruity of the hip joint may occur with excessive internal rotation of the hip[16)] and valgus of the knee[17)]. Thus, femoral anteversion has been considered a risk factor for ACL injury. However, the association between neuromuscular control and hip and knee joint kinematics upon single-leg landing due to differences in femoral anteversion remains unclear.

This study aimed to clarify the relationship between femoral anteversion and hip and knee joint kinematics and muscle activity of the lower extremity upon single-leg landing in terms of risk factors for ACL injury. We hypothesized that increased femoral anteversion leads to an ACL injury-risk position and, therefore, must be considered a risk factor for ACL injury.

SUBJECTS AND METHODS

Sixteen healthy female college students (age, 20.8 ± 1.0 years; height, 160.9 ± 3.8 cm; weight, 54.1 ± 5.8 kg) participated in this study. The purpose of the study and the measurements involved were explained to the subjects beforehand, both verbally and in writing, and their consent

*To whom correspondence should be addressed.
E-mail: kanemasha.1978@gmail.com

was obtained. This study was approved by the Ethics Committee of Juntendo University Graduate School of Health and Sports Science.

Subjects performed a single-leg landing task by standing on both legs on a 30-cm high platform. They then jumped 30 cm forward, landing on the left leg. To minimize the effects of efforts for maintaining balance, subjects were instructed to cross their arms on their chest. Inability to maintain this posture upon landing and allowing the opposite leg to contact the ground were considered failures. The task was concluded when the subject had made three successful attempts.

Femoral anteversion was measured using Craig's test. For measurement, the knee joint of the tested leg was maintained at 90° while in a prone position. The examiner palpated the greater trochanter while passively rotating the hip until the most prominent part of the greater trochanter reached its most lateral position. The angle between the shaft of the tibia and a line perpendicular to the floor was measured using a goniometer[18–20].

The angle of the hip and knee joints upon single-leg landing was measured using the VICON MX three-dimensional (3D) motion analysis system (Vicon Motion Systems, Oxford, UK), eight infrared cameras, and a frequency of 100 Hz. Measurements were performed after the measurement error was confirmed to be 0.7 mm or less.

Ground reaction force was calculated synchronously at 1500 Hz using an AMTI OR6-7 Force Platform (AMTI, Watertown, MA, USA). Infrared reflective markers were affixed to the body according to the marker positions in the Plug-in Gait lower body model (Vicon Motion Systems). Markers were placed at 16 sites: both the anterior superior and posterior superior iliac spines, the center of both thighs externally, the lateral joint line of both knees, the center of both shanks externally, the lateral malleolus of both ankles, the center of both heels, and the head of the second metatarsal of both feet.

Marker trajectories were filtered with a Woltring low-pass filter and a 20-Hz cut-off frequency. Ground reaction force data were filtered with a fourth-order Butterworth low-pass filter with zero lag and a 6-Hz cut-off frequency. The hip and knee joint angles in the sagittal and frontal planes were calculated using a Plug-in Biomechanical Modeler (Vicon Motion Systems). The angle of hip flexion and extension were defined as the angles formed by the pelvic axis and femoral axis in the sagittal plane, with flexion denoted as + and extension denoted as −. The angle of hip adduction and abduction were defined as the angles formed by the pelvic axis and femoral axis in the frontal plane, with adduction denoted as + and abduction denoted as −. The angle of knee flexion and extension were defined as the angles formed by the femoral axis and shank axis in the sagittal plane, with flexion denoted as + and extension denoted as −. The varus and valgus angles of the knee were defined as the angles formed by the femoral axis and shank axis in the frontal plane, with varus denoted as + and valgus denoted as −. The measurement interval was calculated from the initial ground contact upon landing to 100 ms immediately after ground contact. Initial ground contact was defined as the point at which the vertical ground reaction force exceeded 8 N.

Muscle activity of the lower extremity was measured using a TeleMyo 2400 Surface Electromyography System (Noraxon, Scottsdale, AZ, USA) and a 1500-Hz sampling frequency. The electromyography (EMG) data were collected from four muscles: the rectus femoris (RF), gluteus maximus (GM), semitendinosus (ST), and biceps femoris (BF). The skin over the belly of each muscle was prepared for electrode placement by dry shaving and cleaning the area with alcohol to reduce surface impedance. Blue Sensor M-00-S surface electrodes (Ambu, Ballerup, Denmark) affixed in the direction of the muscle fibers with a 2-cm interelectrode distance. The position of the electrode for RF was the midpoint of a line connecting the anterior superior iliac spine and the top of the patella, that for the GM was the midpoint of a line connecting the sacrum and the greater trochanter of the femur, that for the ST was the midpoint of a line connecting the ischial tuberosity and the medial epicondyle of the tibia, and that for the BF was the midpoint of a line connecting the ischial tuberosity and the lateral epicondyle of the tibia. The EMG data were processed with a 50-ms root mean square moving window. The EMG data for each muscle were obtained from 100 ms before ground contact to 100 ms immediately after ground contact.

Muscle activity 100 ms before ground contact indicated preactivity of the muscle. Preactivity plays a key role in maintaining knee joint stability after the impact of landing because the feedback mechanism alone is not enough to provide knee joint stability[21, 22]. Thus, the activity approximately 100 ms before ground contact is crucial[23, 24]. ACL strain upon landing reportedly peaks at approximately 40 ms after landing, and the ground reaction force after landing reportedly peaks at approximately 100 ms after landing[4, 25, 26]. Thus, muscle activity 100 ms after landing was measured because ACL injury most likely occurs within 100 ms of landing. Measurements of the maximum voluntary contraction (MVC) utilized the test positions for manual muscle testing reported by Daniels and Worthingham. Normalization (%MVC) of the EMG data for each muscle was accomplished using the EMG data for 1 s during the waveform of MVC for 5 s. All measurements were performed for the left leg, and the mean of three attempts was used in the analysis.

The mean for the 16 subjects was calculated on the basis of measurements of femoral anteversion; the subjects were divided into two groups: the high group with six subjects at the upper end of the mean and the low group with six subjects at the lower end of the mean. Although the femoral anteversions of the two groups were significantly different, there were no significant differences between the groups for age, height, and weight (Table 1).

All statistical tests were performed using the SPSS 11.0 J for Windows statistical software (SPSS Inc., Chicago, IL, USA). The low and high groups were compared using unpaired t-tests for the kinematics and EMG data. The alpha level for determining statistical significance was set at 0.05.

Table 1. Characteristics of subjects

	Low group (n = 6)	High group (n = 6)
Age (years)	20.3 ±1.4	20.7 ± 0.5
Height (cm)	162.4 ± 3.3	158.5 ± 3.5
Weight (kg)	55.3 ± 4.7	54.0 ± 7.2
Femoral anteversion (deg)	16.1 ± 1.7*	20.7 ± 3.3*

Mean ± SD. *p<0.05

Table 2. Group comparisons of joint motion

Time (ms)	Hip flexion (deg)		Hip abduction (deg)		Knee flexion (deg)		Knee varus and valgus (deg)	
	Low group	High group	Low group	High group	Low group	High group	Low group	High group
0	24.8 ± 3.3*	16.2 ± 6.6*	−8.1 ± 3.1	−9.8 ± 3.3	4.3 ± 2.8	6.1 ± 4.5	2.4 ± 1.5	2.4 ± 3.3
10	25.0 ± 3.3*	16.3 ± 6.4*	−8.5 ± 3.2	−10.1 ± 3.4	5.8 ± 2.7	8.3 ± 4.9	2.8 ± 1.8	2.4 ± 3.3
20	25.4 ± 3.3*	16.6 ± 6.3*	−8.9 ± 3.4	−10.4 ± 3.5	7.8 ± 2.6	10.8 ± 5.2	3.5 ± 2.2	2.5 ± 3.4
30	26.0 ± 3.3*	17.2 ± 6.3*	−9.3 ± 3.6	−10.4 ± 3.6	10.0 ± 2.5	13.7 ± 5.6	4.2 ± 2.6	2.6 ± 3.7
40	26.9 ± 3.3*	18.1 ± 6.4*	−9.6 ± 3.7	−10.3 ± 3.7	12.4 ± 2.5	16.8 ± 6.0	5.1 ± 3.1	2.5 ± 4.0
50	27.9 ±3.4*	19.1 ± 6.5*	−9.8 ± 3.8	−9.8 ± 3.8	15.1 ± 2.6	20.0 ± 6.3	6.0 ± 3.6	2.3 ± 4.4
60	29.2 ± 3.5*	20.4 ± 6.5*	−9.9 ± 3.8	−9.0 ± 3.9	17.8 ± 2.7	23.3 ± 6.4	6.9 ± 4.0	2.0 ± 4.8
70	30.6 ± 3.6*	21.7 ± 6.6*	−9.8 ± 3.7	−8.0 ± 4.0	20.6 ± 2.9	26.7 ± 6.4	7.6 ± 4.4*	1.4 ± 5.2*
80	32.1 ± 3.8*	23.1 ± 6.6*	−9.4 ± 3.6	−6.7 ± 4.2	23.5 ± 3.2*	29.9 ± 6.2*	8.2 ± 4.6*	0.8 ± 5.5*
90	33.7 ± 4.2*	24.4 ± 6.5*	−8.8 ± 3.4	−5.4 ± 4.3	26.3 ± 3.7*	33.3 ± 5.9*	8.5 ± 4.9*	0.1 ± 5.7*
100	35.1 ± 4.6*	25.5 ± 6.5*	−8.1 ± 3.2	−4.1 ± 4.4	29.1 ± 4.2*	36.0 ± 5.8*	8.6 ± 5.1*	−0.5 ± 6.0*

Mean ± SD. *p<0.05. 0 ms, ground contact

Table 3. Muscle activity 100 ms before ground contact

Group	Low group (%)	High group (%)
GM	1.5 ± 0.6*	0.5 ± 0.5*
RF	0.9 ± 0.4*	1.7 ± 0.5*
ST	1.6 ± 0.6	2.1 ± 1.0
BF	2.2 ± 0.9	2.8 ± 1.8

Mean ± SD. *p<0.05. GM, gluteus maximus; RF, rectus femoris; ST, semitendinosus; BF, biceps femoris

Table 4. Muscle activity 100 ms after ground contact

Group	Low group (%)	High group (%)
GM	3.3 ± 1.2	2.1 ± 1.2
RF	4.0 ± 1.4*	5.8 ± 0.7*
ST	1.9 ± 0.8	1.2 ± 0.5
BF	3.5 ± 1.2	3.7 ± 1.9

Mean ± SD. *p<0.05. GM, gluteus maximus; RF, rectus femoris; ST, semitendinosus; BF, biceps femoris

RESULTS

Table 2 shows the mean hip and knee joint angles in the sagittal and frontal planes upon single-leg landing. The hip joint flexion from initial ground contact upon landing to 100 ms immediately after ground contact was significantly lower in the high group than in the low group ($p < 0.05$; Table 2). Significant differences between the two groups for hip joint adduction and abduction were not observed ($p < 0.05$; Table 2). The knee joint flexion from 80 to 100 ms after initial ground contact was significantly higher in the high group than in the low group ($p < 0.05$; Table 2). The knee valgus from 70 to 100 ms after the initial ground contact was significantly higher in the high group than in the low group ($p < 0.05$; Table 2).

Muscle activity 100 ms before ground contact indicated that the %MVC of the GM was significantly less in the high group (0.5% ± 0.5%) than in the low group (1.5% ± 0.6%, $p < 0.05$; Table 3). The %MVC of the RF was significantly greater in the high group (1.7% ± 0.5%) than in the low group (0.9% ± 0.4%, $p < 0.05$; Table 3). Significant differences between the groups in %MVC of the ST and BF were not observed (Table 3). Muscle activity 100 ms after ground contact indicated that the %MVC of the RF was significantly greater in the high group (5.8% ± 0.7%) than in the low group (4.0% ± 1.4%, $p < 0.05$; Table 4). Significant differences between the two groups in %MVC of GM, ST, and BF were not observed (Table 4).

DISCUSSION

Abnormal alignment of the lower extremity has been proposed as a risk factor for acute and chronic lower extremity injuries such as ACL injuries, patellofemoral syndrome, and plantar fasciitis. In addition, it has been suggested that biomedical changes caused by abnormal alignment of the lower extremity may influence joint load and muscles as well as neuromuscular control and function[27].

Increased femoral anteversion causes anterior displacement of the femoral head in the acetabulum and a decrease in congruity of the hip joint. Thus, improvement in the congruity of the hip joint may occur with excessive internal rotation of the hip[16]. Excessive internal rotation of the hip leads to knee valgus alignment as a result of a kinematic chain[17]; thus, an increase in femoral anteversion may cause ACL injury. Our study results showed that the valgus of the knee in the high group increased after single-leg landing. This was because the congruity of the acetabulum and the head of the femur may have improved to protect the hip joint from the impact of landing; excessive internal rotation of the hip may result in a kinematic chain. This study focused on joint kinematics in the frontal plane as well as the kinematics of the hip and knee joints in the sagittal plane. The hip and knee joint kinematics in the high group were characterized by lower hip joint flexion and higher knee joint flexion after single-leg landing.

On the basis of video image analysis of ACL injuries, a recent study reported that ACL injury occurs when the trunk is positioned behind the leg[28]. In other words, body position in the sagittal plane affects kinetic and kinematic components of the leg, presenting a risk of ACL injury. Blackburn and Padua demonstrated that increased trunk extension upon landing increased the ground reaction force and the quadriceps forces[29]. Similarly, Kulas et al. reported that increased trunk extension upon landing caused an increase in quadriceps forces and a decrease in hamstring forces[30]. Increased quadriceps forces lead to anterior tibial displacement, increasing the load on the ACL[28, 31]. Increased trunk extension is associated with a decrease in the angle of hip joint flexion[32]. Thus, increase of the femoral anteversion suggests a physical characteristic that probably leads to ACL injury. The leg position upon landing is maintained by soft tissues such as muscles, the joint capsule, and ligaments. However, the feedback mechanism alone is not enough to provide knee joint stability and prevent injury[21, 22]. Thus, the feedforward mechanism is important for providing joint stability and control. ACL injury typically occurs immediately after ground contact upon landing or deceleration[5].

Once the foot contacts the ground, the ground reaction force and ACL strain peak. Cerulli et al. reported that ACL strain peaks approximately 40 ms after ground contact[25], whereas Schmitz et al. reported that it increased approximately 100 ms after ground contact[4]. Thus, ACL injury occurs 40–100 ms after ground contact, the time during which ACL strain peaks. However, the feedback mechanism cannot provide joint stability at this point; therefore, the feedforward mechanism is important for joint stability[23, 24]. The results concerning EMG activity in this study indicated lower muscle activity of the gluteus maximus and greater muscle activity of the rectus femoris before ground contact in the high group than in the low group.

The gluteus maximus controls excessive hip internal rotation[33] and resists knee valgus moment[34]. Thus, the decrease in gluteus maximus activity before ground contact indicates that soft tissues such as ligaments are involved in limiting excessive hip internal rotation and that the ability to resist valgus of the knee diminished. The rectus femoris resists knee flexion moment[32] and produces anterior tibial displacement[32, 35, 36]. Moreover, this muscle leads to valgus of the knee and internal rotation of the tibia[37]; thus, the rectus femoris increases the load on the ACL. Therefore, increased rectus femoris activity before and after ground contact may cause knee extensor moment and anterior tibial displacement. Therefore, activity in these muscles may promote valgus of the knee and anterior tibial displacement, both of which predispose to ACL injury.

Based on hip and knee joint kinematics upon single-leg landing and muscle activities immediately before and after ground contact, the above findings suggest that increased femoral anteversion may be a physical characteristic that probably leads to ACL injury. Furthermore, femoral anteversion may help to identify individuals who are likely to develop ACL injury and may be considered a risk factor for ACL injury.

There are some limitations to this study. Femoral anteversion was measured on the surface of the skin alone. The kinematics of hip and knee joints upon single-leg landing were analyzed using the Plug-in Gait model. Measurement of femoral anteversion is reportedly more reliable than techniques using X-rays[38]. However, studies using more accurate measurement techniques involving sectional computed tomography images are required. In addition, Kadaba et al. indicated that disparities in the placement of infrared reflective markers in the Plug-in Gait model can lead to errors; i.e., errors in measurement of joint motion other than knee joint flexion/extension[39]. Future studies should validate the knee joint motion in detail using the point cluster technique[40], which allows more accurate measurement of the 3D motion of the knee.

REFERENCES

1) Viskontas DG, Giuffre BM, Duggal N, et al.: Bone bruises associated with ACL rupture, correlation with injury mechanism. Am J Sports Med, 2008, 36: 927–933. [Medline] [CrossRef]
2) Nagano Y, Ida H, Akai M, et al.: Gender differences in knee kinematics and muscle activity during single limb drop landing. Knee, 2007, 14: 218–223. [Medline] [CrossRef]
3) Joseph M, Tiberio D, Baird JL, et al.: Knee valgus during drop jumps in national collegiate athletic association division I female athletes. Am J Sports Med, 2008, 36: 285–289. [Medline] [CrossRef]
4) Schmitz RJ, Kulas AS, Perrin DH, et al.: Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech (Bristol, Avon), 2007, 22: 681–688. [Medline] [CrossRef]
5) Boden BP, Dean GS, Feagin JA: Mechanisms of anterior cruciate ligament injury. Orthopedics, 2000, 23: 573–578. [Medline]
6) Shimokochi Y, Jatin PA, Eric GM, et al.: Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc, 2013, 21: 888–897. [Medline] [CrossRef]
7) Yamazaki J, Muneta T, Ju YJ, et al.: Differences in kinematics of single leg squatting between anterior cruciate ligament-injured patients and healthy controls. Knee Surg Sports Traumatol Arthrosc, 2010, 18: 56–63. [Medline] [CrossRef]
8) Trimble MH, Bishop MD, Buckley BD, et al.: The relationship between clinical measurements of lower extremity posture and tibial translation. Clin Biomech (Bristol, Avon), 2002, 17: 286–290. [Medline] [CrossRef]
9) Blackburn JT, Padua DA: Influence of trunk on hip and knee joint kinematics during a controlled drop landing. Clin Biomech (Bristol, Avon), 2008, 23: 313–319. [Medline] [CrossRef]
10) Hewett TE, Myer GD, Ford KR, et al.: Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruci-

ate ligament injury risk in female athletes. Am J Sports Med, 2005, 33: 492–501. [Medline] [CrossRef]

11) Olsen OE, Myklebust G, Engebretsen L, et al.: Injury mechanisms for anterior cruciate ligament injuries in team handball. Am J Sports Med, 2004, 32: 1002–1012. [Medline] [CrossRef]
12) Philippon M, Dewing C, Briggs K, et al.: Decreased femoral head-neck offset: a possible risk factor for ACL injury. Knee Surg Sports Traumatol Arthrosc, 2012, 20: 2585–2589. [Medline] [CrossRef]
13) Lawrence RK, Kernozek TW, Miller EJ, et al.: Influences of hip external rotation strength on knee mechanics during single-leg drop landings in female. Clin Biomech (Bristol, Avon), 2008, 23: 806–813. [Medline] [CrossRef]
14) Boden BP, Torg JS, Knowles SB, et al.: Video analysis of anterior cruciate ligament injury. Am J Sports Med, 2009, 37: 252–259. [Medline] [CrossRef]
15) MacEwen GD: Anteversiom of the femur. Postgrad Med, 1976, 60: 154–156. [Medline]
16) McClay I, Manal K: A comparison of three-dimensional lower extremity kinematics during running between excessive pronators and normals. Clin Biomech (Bristol, Avon), 1998, 13: 195–203. [Medline] [CrossRef]
17) Lephart SM, Ferris CM, Riemann BL, et al.: Gender differences in strength and lower extremity kinematics during landing. Clin Orthop Relat Res, 2002, 401: 162–169. [Medline] [CrossRef]
18) Bonci CM: Assesment and evaluation of predisposing factors to anterior cruciate ligament injury. J Athl Train, 1999, 34: 155–164. [Medline]
19) Nguyen AD, Shultz SJ: Sex differences in clinical measures of lower extremity alignment. J Orthop Sports Phys Ther, 2007, 37: 389–398. [Medline]
20) Nyland J, Kuzemchek S, Parks M, et al.: Femoral anteversion vastus medialis and gluteus medius EMG amplitude: composite hip abductor EMG amplitude ratios during isometric combined hip abduction-external rotation. J Electromyogr Kinesiol, 2004, 14: 255–261. [Medline] [CrossRef]
21) Medina JM, Valovich TC, Howell SK, et al.: Timing of neuromuscular activation of the quadriceps and hamstrings prior to landing in high school male athletes, female athletes and female non-athletes. J Electromyogr Kinesiol, 2008, 18: 591–597. [Medline] [CrossRef]
22) Palmieri-Smith RM, Wojtys EM, Ashton-Miller JA: Association between preparatory muscle activation and peak valgus knee angle. J Electromyogr Kinesiol, 2008, 18: 973–979. [Medline] [CrossRef]
23) Greenwood R, Hopkins A: Muscle responses during sudden falls in man. J Physiol, 1976, 254: 507–518. [Medline]
24) Santello M, Mc donagh, MJ: The control of timing and amplitude of EMG activity in landing movements in humans. Exp Physiol, 1998, 83: 857–874. [Medline]
25) Cerulli G, Benoit DL, Lamontagne M, et al.: In vivo anterior cruciate ligament strain behavior during a rapid deceleration movement: case report. Knee Surg Sports Traumatol Arthrosc, 2003, 11: 307–311. [Medline] [CrossRef]
26) Shin CS, Chaudhari AM, Andriacchi TP: The influence of deceleration forces on ACL strain during single-leg landing: a simulation study. J Biomech, 2007, 40: 1145–1152. [Medline] [CrossRef]
27) Daneshmandi H, Saki F, Shahheidari S, et al.: Lower extremity malalignment and its liner relation with Q angle in female athletes. Procedia Soc Behav Sci, 2011, 15: 3349–3354. [CrossRef]
28) Sheehan FT, Sipprell WH, Boden BP, et al.: Dynamic sagittal plane trunk control during anterior cruciate ligament injury. Am J Sports Med, 2012, 40: 1068–1074. [Medline] [CrossRef]
29) Blackburn JT, Padua DA: Sagittal plane trunk position, landing forces, and quadriceps electromyographic activity. J Athl Train, 2009, 44: 174–179. [Medline] [CrossRef]
30) Kulas A, Zalewski P, Hortobagyi T, et al.: Effects of added trunk load and corresponding trunk position adaptations on lower extremity biomechanics during drop-landings. J Biomech, 2008, 41: 180–185. [Medline] [CrossRef]
31) Beaulieu ML, Mclean SG: Sex-dimorphic landing mechanics and their role within the noncontact ACL injury mechanism; evidence, limitations and directions. SMARTT, 2012, 4: 1–13. [Medline]
32) Shimokochi Y, Shultz SJ: Mechanisms of noncontact anterior cruciate ligament injury. J Athl Train, 2008, 43: 396–408. [Medline] [CrossRef]
33) Ireland ML, Willson JD, Ballantyne BT, et al.: Hip strength in females with and without patellofemoral pain. J Orthop Sports Phys Ther, 2003, 33: 671–676. [Medline]
34) Leetun DT, Ireland ML, Willson JD, et al.: Core stability measures as risk factors for lower extremity injury in athletes. Med Sci Sports Exerc, 2004, 36: 926–934. [Medline] [CrossRef]
35) Chaudhari AM, Andriacchi TP: The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. J Biomech, 2006, 39: 330–338. [Medline] [CrossRef]
36) Myer GD, Ford KR, Hewett TE: The effects of gender on quadriceps muscle activation strategies during a maneuver that mimics a high ACL injury risk position. J Electromyogr Kinesiol, 2005, 15: 181–189. [Medline] [CrossRef]
37) 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. [Medline] [CrossRef]
38) Ruwe PA, Gage JR, Ozonoff MB, et al.: Clinical determination of femoral anteversion. A comparison with established techniques. J Bone Joint Surg Am, 1992, 74: 820–830. [Medline]
39) Kadaba MP, Ramakrishnan HK, Wootten ME, et al.: Repeatability of kinematic, kinetic, and electromyographic date in normal adult gait. J Orthop Res, 1989, 7: 849–860. [Medline] [CrossRef]
40) Andriacchi TP, Alexander EJ, Toney MK, et al.: A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. J Biomech Eng, 1998, 120: 743–749. [Medline] [CrossRef]