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Immediate Effects of a Knee Brace With a Constraint to Knee Extension on Knee Kinematics and Ground Reaction Forces in a Stop-Jump Task

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Background: A small knee flexion angle in landing tasks was identified as a possible risk factor for noncontact anterior cruciate ligament injuries that are common in sports.

Hypothesis: A specially designed knee brace with a constraint to knee extension would significantly increase the knee flexion angle at the landing of athletic tasks preceded with horizontal movement components, such as stop-jump tasks.

Study Design: Repeated measure design for brace effects.

Methods: Three-dimensional videographic and force plate data were collected for 10 male and 10 female recreational athletes performing a stop-jump task with and without the specially designed brace. Knee flexion angle at landing, maximum knee flexion angle, and peak ground reaction forces during the stance phase of the stop-jump task were determined for each subject with and without the knee brace.

Results: The knee brace increased the knee flexion angle at the landing by 5° for both genders but did not significantly affect the peak ground reaction forces during the landing.

Conclusions: The specially designed knee brace may be a useful device in the prevention and rehabilitation of noncontact anterior cruciate ligament injuries in sports.

Keywords: anterior cruciate ligament (ACL); injury prevention; brace; landing; biomechanics

Acute ACL injuries are common knee injuries in sports participants.[12,25,29] An ACL injury has a devastating effect on the individual, resulting in high levels of short-term disability and increasing the likelihood of secondary knee disorders, such as osteoarthritis, in later life.[19,26,29,30,32] In sports, women have an ACL injury rate 3 to 10 times higher than the rate in men.[2,8,20,22] The primary mechanism of the injury is noncontact in nature; that is, there is no physical contact between the patient and other people at the time of injury.[4,24,26] The noncontact nature of the majority of ACL injuries suggests that the intrinsic forces generated by patients themselves are likely to be an important cause.[10,15,16,30]

Previous studies have shown that noncontact ACL injuries mainly occur in the performance of certain athletic tasks, such as stop-jump, landing, and cutting.[1,2,4,11,14] Previous studies have also shown that female recreational athletes have lower extremity motor controls that may increase the load on their ACLs in specific athletic tasks, in comparison with their male counterparts.[7,21] For our long-term studies on the prevention of noncontact ACL injuries, we therefore hypothesized that women tend to have altered lower extremity motor controls that, in specific athletic tasks, frequently bring them close to positions in which noncontact ACL injuries may occur, thereby increasing their risk for noncontact ACL injuries.

One of the characteristics of female recreational athletes' movement is their small knee flexion angle in landing tasks that are preceded with horizontal movements, such as stop-jump tasks.[7,21] Regarding biomechanics, decreasing the knee flexion angle at landing increases the

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loading on the ACL[5,13,23,28,31] and thereby increases the risk for ACL injuries. As a natural continuation of our preliminary studies, the purpose of this study was to address the effects of constraining knee extension on lower extremity kinematics and kinetics by having recreational athletes wear a specially designed knee brace during a stop-jump task. It was hypothesized for this study that the specially designed knee brace would significantly increase the knee flexion angle at the landing of the stop-jump task and that the maximum ground reaction forces would be reduced as the knee flexion angle at landing increased.

MATERIALS AND METHODS

Twelve male and 12 female healthy recreational athletes between 18 and 28 years of age without known histories of knee disorders were recruited as the subjects for this study (Table 1). A *recreational athlete* was defined as a person who plays sports 2 to 3 times per week regularly without following a professionally designed training scheme. All subjects were recruited through advertising from the general student population on a university campus. The use of human subjects was approved by the institution's internal review board.

The newly designed knee brace with a constraint to knee extension was constructed using an existing functional knee brace (4titude; dj Orthopedics, LLC, Vista, Calif). The brace frame was made of 6061-T6 aluminum with upright upper thigh and lower calf cuffs. Hook-and-loop straps are used to attach the brace to the leg (Figure 1). The medium-size brace weighs 20 oz. The new design uses a spring mechanism to constrain knee extension. The resistance mechanism in the hinge (Figure 1) engages at 40° of knee flexion and applies a gradually increasing resistance to knee extension motion up to 10° of knee flexion, at which point a rigid stop prevents further knee extension. The resistive torque is adjustable with a maximum of 3 N·m at 10° of knee flexion. A total of 8 such braces were made for right and left sides in each of the 4 sizes: small, medium, large, and extra large.

The athletic task tested in this study was a vertical stop-jump task frequently performed in basketball and volleyball games. This task consists of an approach run, with up to 5 steps, and a 2-footed landing followed by a 2-footed takeoff for the maximum height (Figure 2). A recent review of more than 100 ACL injury cases on videotape[3] revealed that 70% of noncontact ACL injuries occurred in stop-jump–related tasks.

TABLE 1
Subject Mean Age, Height, and Weight

Gender	Age, y	Body Weight, kg	Standing Height, m
Male	26.0 ± 2.5	83.5 ± 11.9	1.82 ± 0.1
Female	26.0 ± 2.6	61.9 ± 9.1	1.66 ± 0.1

All subjects underwent testing in the Motion Analysis Laboratory of the Center for Human Movement Science of the University of North Carolina at Chapel Hill. Subjects signed informed consent forms before data collection. Subjects were instructed to have a 10-minute warm-up before data collection. The stop-jump task and the knee braces were described to the subject; demonstration of the task was avoided to minimize coaching effects. All subjects were blinded to the hypothesis of this study.

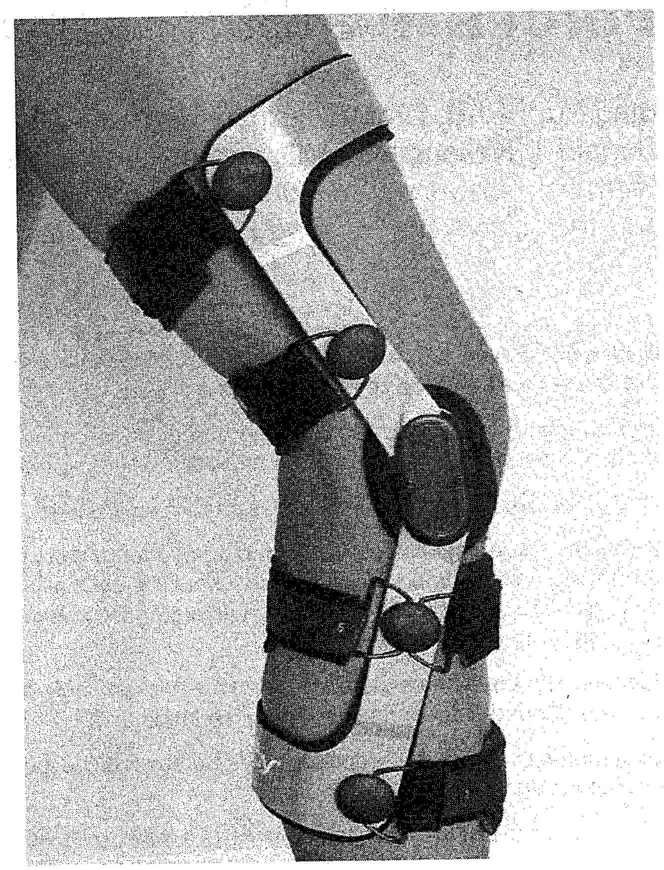

Figure 1. The specially designed knee brace with a constraint to knee extension.

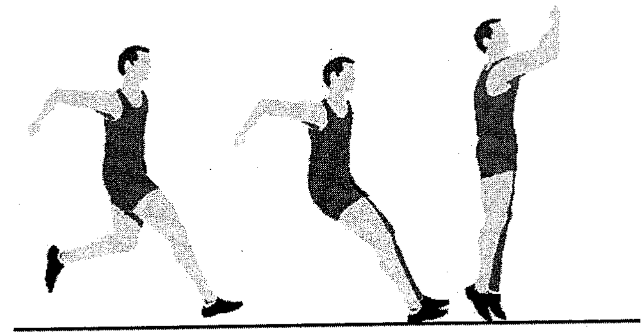

Figure 2. The stop-jump task.

Passive reflective markers were placed on each subject bilaterally at the anterior superior iliac spine, lateral malleolus, upper anterior aspect of the tibia, and lower anterior aspect of the tibia. A marker was also placed on the lower spine at the L4-L5 joint.[17] The subjects performed the stop-jump task with the above-described markers. Each subject performed 5 successful trials of the stop-jump task at the maximum approach run speed and vertical jump effort he/she felt comfortable with for each of the 2 brace conditions: (1) without brace and (2) with the specially designed knee brace with a constraint to knee extension. The order of the 2 conditions was randomized. The newly designed knee brace, in the appropriate size, was applied to the dominant leg of each subject. The *dominant* leg was defined as the leg the subject used for single-leg jumping.

Three-dimensional (3D) videographic and force plate data were collected for each subject in the stop-jump task for the 2 brace conditions. Six infrared video cameras were used to collect the trajectories of reflective markers on the subject at a frame rate of 120 frames/s. The 6 infrared cameras were calibrated for a 2.5 m long × 1.5 m wide × 2.5 m high space (calibration volume), in which the subject performed the stop-jump task. Two Type 4060A Bertec force plates (Bertec Corp, Worthington, Ohio) were used to collect the ground reaction force signals at a sample rate of 1200 samples/channel/s. The videographic and ground reaction force signals were recorded by the Peak Performance Motus videographic and analog data acquisition system (Peak Performance Technology Inc, Englewood, Colo). The videographic and force plate data collection was time synchronized to 1200 frames/s and 1200 samples/channel/s.

Additional 3-D videographic data were collected in a standing trial after all the stop-jump trials. Additional passive reflective markers were placed bilaterally at the medial malleolus and medial and lateral femur condyles. These additional markers were used to estimate the locations of those critical body landmarks that were needed for calculating joint centers but that were not clearly visible when the subjects were performing the stop-jump task. Each subject was asked to stand in the middle of the calibration volume. The 3-D videographic data of all reflective markers were collected.

The collected 3-D coordinates of the markers during each stop-jump trial were filtered through a Butterworth lower-pass digital filter at estimated optimum cutoff frequencies.[33] The 3-D local coordinates of the medial and lateral femur condyles and medial malleolus were estimated from the 3-D coordinates of markers on the tibia in the standing trial. The 3-D coordinates of the hip joint centers in stop-jump trials were estimated from the 3-D coordinates of the reflective markers on the right and left anterior superior iliac spines and L4-L5 joints and on anatomical data.[3] The 3-D coordinates of the medial and lateral femur condyles and medial malleolus in stop-jump trials were estimated from the local coordinates of the corresponding markers in the standing trials, and the direction cosine matrices of the tibia was defined by the 3-D coordinates of the markers on the tibia in stop-jump trials. The *knee joint center* was defined as the middle point between the medial and lateral femur condyles. The *ankle joint center* was defined as the middle point between the medial and lateral malleolus. The 3-D coordinates of the knee and ankle joint centers and medial and lateral malleolus were used to define the *tibia reference frame*. The 3-D coordinates of the knee and hip joint centers and medial and lateral femur condyles were used to define the *femur reference frame*. The knee joint angles were determined as Euler angles of the tibia reference frame relative to the femur reference frame rotated in order of (1) flexion-extension (z-axis), (2) varus-valgus (y-axis), and (3) internal-external rotation (x-axis).[4] The electric signals from the force plates were converted to forces. All signal processing and data reduction were performed using a MotionSoft 3-D motion data reduction program package version 5.5 (MotionSoft Inc, Chapel Hill, NC). The validity and reliability of estimated joint centers and angles can be found in other studies.[3,6,17,18,34]

The *stance phase* of the stop-jump task was defined as the duration from the time of landing to the time of takeoff. The *time of landing* was defined as the time represented by the first frame in which the vertical ground reaction force was greater than zero after the approach run. The *time of takeoff* was defined as the time represented by the first frame in which the vertical ground reaction force was zero after the landing. The entire stance phase of the stop-jump task was divided into 2 phases: landing and jumping phases. The *landing phase* was defined as the duration from the time of landing to the time of the maximum knee flexion angle. The *jumping phase* was defined as the time of the maximum knee flexion angle to the time of takeoff. The approach run speed, knee flexion angle at the landing, maximum knee flexion angle, range of knee flexion motion, and maximum posterior, medial, and vertical ground reaction forces during the landing phase were identified for each trial. The *approach run speed* was defined as the magnitude of the mean horizontal velocity of the hip joint centers at the time of landing. The *range of the knee flexion motion* during the landing phase was defined as the difference between the maximum knee flexion angle during the stance phase of the stop-jump task and the knee flexion angle at the landing. The data from the first 3 successful trials in each condition were used for data analysis.

Analyses of variation with mixed design were conducted to compare the knee flexion angle at the landing, maximum knee flexion angle, and range of knee flexion motion with maximum posterior, medial, and vertical ground reaction forces during the landing phase of the stop-jump task. The brace condition was treated as a repeated measure, whereas gender was considered an independent measure. In case of a significant brace condition by gender interaction effect on a given dependent variable, analyses of variance were conducted to compare the dependent variable between brace conditions as a repeated measure for each gender and between genders as independent groups for each brace condition. A type I error rate of .05 was chosen to indicate statistical significance in each analysis. All statistical analyses were performed using the SYSTAT com-

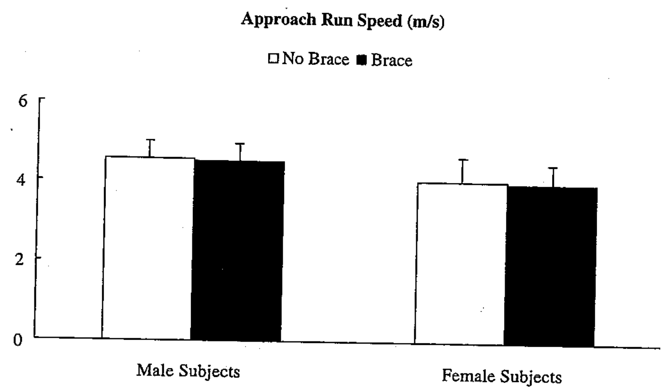

Figure 3. Approach run speed of the stop-jump task with and without the new knee brace. Male subjects had significantly greater approach run speed than did female subjects in both brace and nonbrace conditions (P = .000).

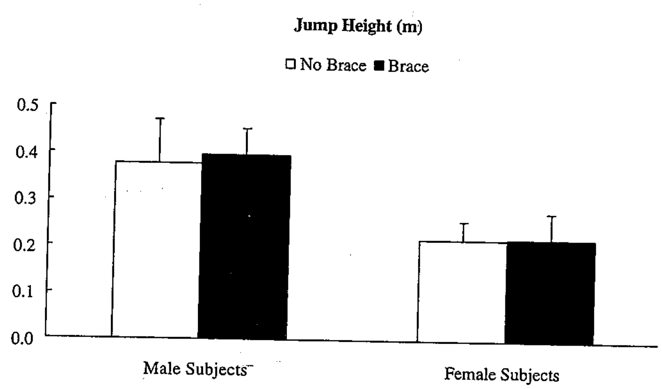

Figure 4. Jump height of the stop-jump task with and without the new knee brace. Male subjects jumped significantly higher than did female subjects (P = .000).

puter program package, version 5.0 (SYSTAT Inc, Evanston, Ill).

RESULTS

There was no significant difference in approach run speed and jump height between brace conditions (P = .825 and .681, respectively). Male subjects' approach run speed and jump height were significantly greater than those of female subjects (P = .000) (Figures 3 and 4).

The specially designed knee brace significantly increased knee flexion angle at the landing in the stop-jump task on average from 27.4° to 32.5° for male subjects and from 22.3° to 27.6° for female subjects (P = .001) (Figure 5). Female subjects had significantly smaller knee flexion angles at the landing in the stop-jump task than did male subjects in both brace and nonbrace conditions (P = .003) (Figure 5).

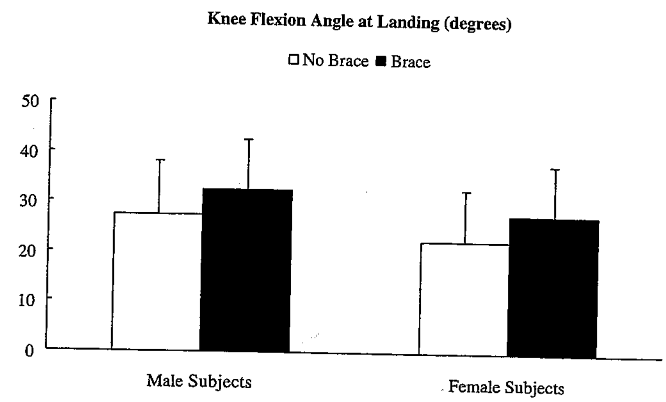

Figure 5. Knee flexion angle at the landing of the stop-jump task with and without the new knee brace. Both male and female subjects had significantly greater knee flexion angles at landing with the brace than without the brace (P = .001). Male subjects had significantly greater knee flexion angles than did female subjects in both brace and nonbrace conditions (P = .003).

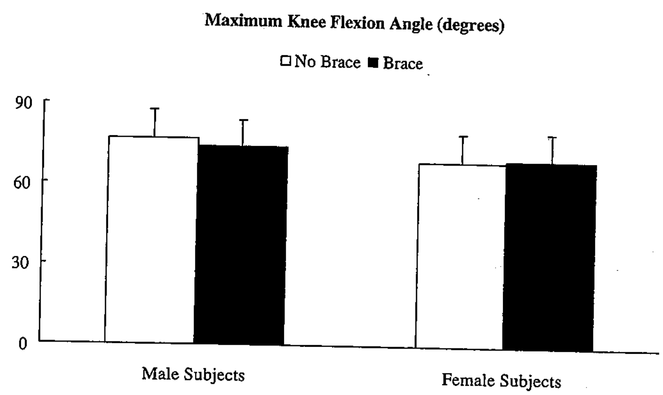

Figure 6. Maximum knee flexion angle during the landing of the stop-jump task with and without the new knee brace. Male subjects had significantly greater maximum knee flexion angles than did female subjects in both brace and nonbrace conditions (P = .001).

There was no significant effect on the maximum knee flexion angle in the stop-jump task (P = .508) in the knee brace condition (Figure 6). Female subjects had significantly smaller maximum knee flexion angles in the stop-jump task than did male subjects in both brace and nonbrace conditions (P = .001) (Figure 6).

The specially designed knee brace significantly reduced the angle of the range of knee flexion motion in the stop-jump task on average from 49.4° to 41.3° for male subjects and from 46.3° to 41.9° for female subjects (P = .015) (Figure 7). There was no significant difference in the range of knee flexion motion in the stop-jump task between male and female subjects (P = .498) (Figure 7).

There was no significant effect on the maximum posterior ground reaction force in the stop-jump task from the

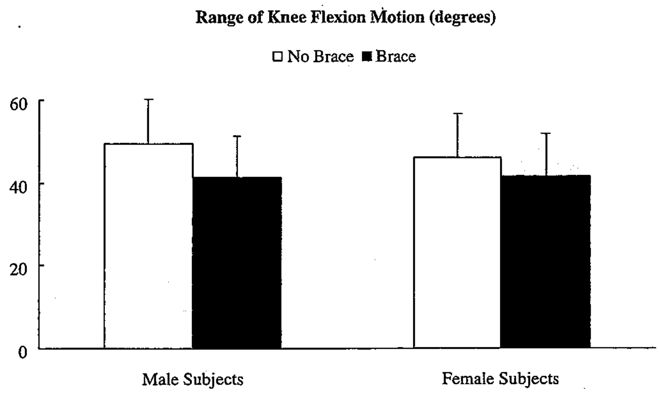

Figure 7. Range of knee flexion motion during the landing of the stop-jump task with and without the new knee brace. Both male and female subjects had significantly smaller ranges of knee flexion motion with the brace than without the brace (P = .015).

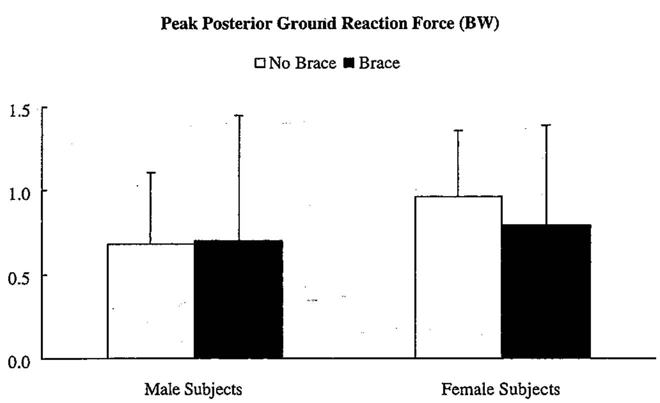

Figure 8. Peak posterior ground reaction force during the landing of the stop-jump task with and without the new knee brace. Female subjects had significantly greater maximum posterior ground reaction force during the landing phase than did male subjects in both brace and nonbrace conditions (P = .007). The peak posterior ground reaction force was normalized to body weight (BW).

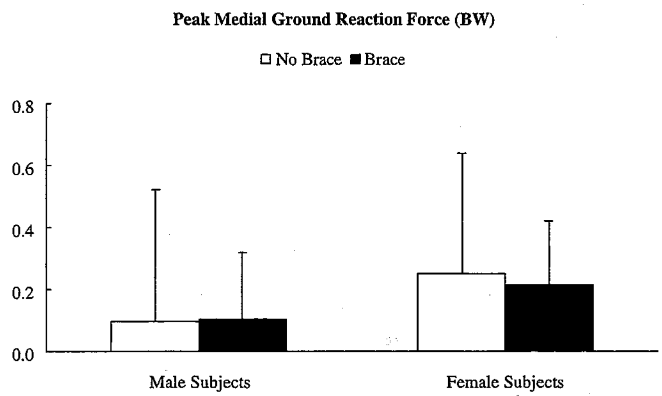

Figure 9. Peak medial ground reaction force during the landing of the stop-jump task with and without the new knee brace. Female subjects had significantly greater maximum medial ground reaction force during the landing phase than did male subjects in both brace and nonbrace conditions (P = .000). The peak medial ground reaction force was normalized to body weight (BW).

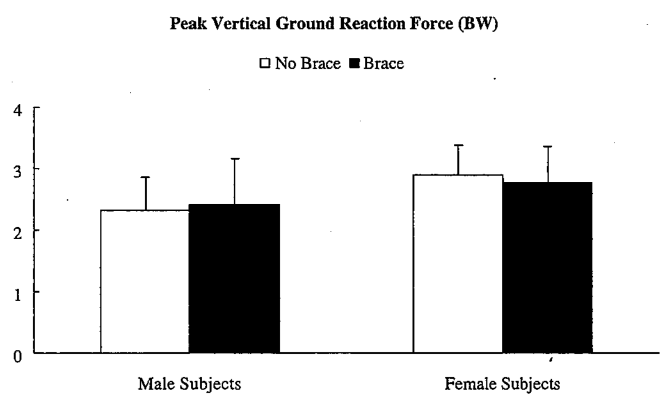

Figure 10. Peak vertical ground reaction force during the landing of the stop-jump task with and without the new knee brace. Female subjects had significantly greater vertical ground reaction force than did male subjects in both brace and nonbrace conditions (P = .003). The peak vertical ground reaction force was normalized to body weight (BW).

specially designed knee brace (P = .588) (Figure 8). Female subjects had significantly greater posterior ground reaction force in the stop-jump task than did male subjects in both brace and nonbrace conditions (P = .007) (Figure 8).

There was no significant effect on the maximum medial ground reaction force in the stop-jump task (P = .708) in the knee brace condition (Figure 9). Female subjects had significantly greater medial ground reaction forces in the stop-jump task than did male subjects in both brace and nonbrace conditions (P = .000) (Figure 9).

There was no significant effect on the maximum vertical ground reaction force in the stop-jump task (P = .708) in the knee brace condition (Figure 10). Female subjects had significantly greater vertical ground reaction forces in the stop-jump task than did male subjects in both brace and nonbrace conditions (P = .003) (Figure 10).

DISCUSSION

The specially designed knee brace did not significantly affect subjects' performances in the stop-jump tasks. The effect on performances and comfort are common concerns when wearing knee braces in sports. The results of this study show that the mean approach run speed and jump height in the stop-jump task were essentially the same with and without the brace for both male and female subjects. These results suggest that the knee brace used in this study did not consistently affect subjects' running and jumping performances in a positive or a negative way. To a certain degree, these results indicate that wearing the specially designed knee brace used in this study was not uncomfortable or that the discomfort of wearing the brace was within the range of tolerance.

The desired function of the mechanism to constrain knee extension while wearing the specially designed knee brace was to modify lower extremity kinematics and kinetics and to reduce the load on the ACL in athletic tasks by increasing the knee flexion angle at the landing. The results of this study supported our hypothesis that the knee brace would significantly increase the knee flexion angle at the landing in the stop-jump task, but they did not support our hypothesis that the knee brace would significantly reduce the maximum ground reaction forces in the stop-jump task. The significant increase in the knee flexion angle at the landing with the specially designed brace was not likely the effect of approach run speed because there was no significant difference in the approach speeds between the brace and nonbrace conditions.

The results of this study also suggested that the specially designed knee brace did not significantly affect the maximum knee flexion angle in the stop-jump task. The significant decrease in the range of knee flexion motion in the stop-jump task with the knee brace was mainly due to the increase in the knee flexion angle at the landing in the knee brace condition. It is likely that the specially designed knee brace did not significantly affect the knee joint resultant forces and moments in this study. These forces and moments are mainly determined by the ground reaction forces and moments because of the relatively small masses and moment of inertia of the foot and shank. There are not likely to be significant differences in knee joint resultant forces and moments if there are no significant differences in ground reaction forces. These results combined together indicated that the specially designed knee brace significantly increased the knee flexion angle at the landing in the stop-jump task, as it was designed to do, but it did not significantly modify other lower extremity kinematics and kinetics in the stop-jump task as it was expected to.

Although the specially designed brace did not significantly reduce the maximum ground reaction forces, it should still have served the overall purpose of the design—to reduce the load on the ACL—because increased knee flexion angle at the landing should assist in reducing the anterior shear force applied on the tibia through the patellar tendon. Studies repeatedly have shown that the anterior shear force applied on the tibia through the patellar tendon is a function of the knee flexion angle.[4,13,23,28,31] The anterior shear force applied on the tibia through the patellar tendon decreases as the patellar tendon–tibia shaft angle decreases, whereas the patellar tendon–tibia shaft angle decreases as the knee flexion angle increases. Therefore, anterior shear force applied on the tibia through the patellar tendon decreases as the knee flexion angle increases. According to the results of a recent study by Nunley et al,[27] the patellar tendon–tibia shaft angle, on average, will be decreased from 19.0° to 17.4° for women and from 13.8° to 12.3° for men when their knee flexion angles increase from 22.3° to 27.6° and from 27.4° to 32.5°, respectively. This means that the anterior shear force applied on the tibia through the patellar tendon, on average, will be reduced by 9% for women and by 13% for men if they increase their knee flexion angles from 22.3° to 27.6° and from 27.4° to 32.5°, respectively. The decrease in the anterior shear force on the tibia should significantly reduce the load on the ACL if other conditions remain the same.

The results of this study indicate that increased knee flexion angle at the landing does not necessarily mean a soft landing. A study by DeVita and Skelly[9] showed that subjects had increased knee flexion angles at the landing and decreased maximum vertical ground reaction forces in a drop landing task when using the soft landing technique in comparison with the knee flexion angles and vertical ground forces when using the hard landing technique. An increase in the knee flexion angle at the landing was recommended to reduce the maximum vertical ground reaction force in the landing. The results of our study, however, indicated that increased knee flexion angle was not likely the cause of the decreased maximum ground reaction force in the study by DeVita and Skelly[9] and, therefore, may not be a critical kinematic characteristic of a soft landing.

The results of this study are in agreement with the literature. Malinzak et al[21] and Chappell et al[7] reported that female recreational athletes had decreased knee flexion angles at the landings of running, cutting, drop landing, and stop-jump tasks in comparison with their male counterparts. The results of the present study showed that female subjects had a smaller knee flexion angle at landing than did male subjects at the landing in the stop-jump task. Chappell et al[7] reported that female recreational athletes showed increased maximum anterior shear force at the proximal tibia at the landing in 3 stop-jump tasks in comparison with their male counterparts. The posterior ground reaction force is a major contributor to the anterior shear force at the proximal tibia. The results of the present study showed that female subjects, on average, had increased posterior ground reaction force at the landing in the stop-jump task. Chappell et al[7] reported that female recreational athletes on average had a valgus moment at the knee, whereas male recreational athletes on average had a varus moment at the knee at the landings of 3 stop-jump tasks. A medial ground reaction force and a valgus knee are contributors to the knee valgus moment. The results of the present study showed that female subjects had increased medial ground reaction force compared with that of male subjects at the landing in the stop-jump task, whereas Malinzak et al[21] reported that female recreational athletes on average had valgus knee and male recreational athletes on average had slightly varus knee at the landings of selected athletic tasks.

Further studies are needed to fully understand the effects of the specially designed knee brace with constraint to knee extension on the lower extremity kinematics and kinetics in athletic tasks and potential clinical applications. Although in the present study the subjects increased their knee flexion angles at the landing of the stop-jump tasks, it is not clear if the increased knee flexion angles were due to the effect of constraining the knee extension or the effect of knee bracing. We did not find evidence in our extensive literature review showing that knee braces without constraint to knee extension assist in reducing knee flexion angle at landings in athletic tasks. Further studies are needed to determine the effects of constraining the knee extension and of purely knee bracing on the lower extremity kinematics and kinetics in athletic tasks. Also, the results of the present study only showed the immediate effects of the specially designed brace on the knee kinematics and kinetics in the stop-jump task. Further studies are needed to determine the long-term training effects of wearing the specially designed knee brace on the lower extremity kinematics and kinetics as compared with not wearing the brace. The present study only investigated the effects of the specially designed knee brace on ground reaction forces. Although the results of the present study showed that wearing the specially designed knee brace did not significantly affect the performance and ground reaction forces in the stop-jump task, and may not affect knee joint resultant moments in the stop-jump task, it is still possible that wearing a knee brace may affect the muscle contraction patterns and techniques to perform athletic tasks. Further studies are needed to compare knee muscle contraction patterns in the stop-jump task with and without the specially designed knee brace. The results of the present study showed a potential to apply the specially designed knee brace in the rehabilitation of ACL injury patients. The present study, however, only tested the immediate effects of the specially designed brace on the knee kinematics and kinetics of healthy recreational athletes without knee injuries. Further studies are needed to determine the effects of the specially designed knee brace on the lower extremity kinematics and kinetics of patients with ACL injuries in postinjury rehabilitation programs.

The results of this study appear to warrant the following conclusions:

1. The immediate application of the specially designed knee brace with the constraint to the knee extension significantly increased knee flexion angle at the landing in the stop-jump task by a mean of 5° for both male and female recreational athletes.
2. The specially designed knee brace with the constraint to the knee extension did not significantly affect the maximum ground reaction forces during the landing phase of the stop-jump task, and the immediate application of the knee brace may not affect the knee joint resultant forces and moments, of recreational athletes.
3. The increased knee flexion angle at the landing of the stop-jump task with the specially designed knee brace may assist in reducing the load on the ACL during the landing in the stop-jump task as well as in other athletic tasks, at least in the immediate application of the knee brace.
4. Further studies are needed to fully understand the potential functions of the specially designed knee brace with the constraint to knee extension in the prevention and rehabilitation programs for ACL injuries.

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REFERENCES

1. Arendt EA, Agel J, Dick R. Anterior cruciate ligament patterns among collegiate men and women. *J Athl Train.* 1999;34:86-92.
2. Arendt EA, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. *Am J Sports Med.* 1995;23:694-701.
3. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23:617-621.
4. Boden BP, Dean GS, Feagin JA, Garrett WE. Mechanisms of ACL injury. *Med Sci Sports.* 1996;28:526.
5. Buff HU, Jones LC, Hungerford DS. Experimental determination of forces transmitted through the patello-femoral joint. *J Biomech.* 1988;21:17-23.
6. Chao EYS. Justification of triaxial goniometer for the measurement of joint rotation. *J Biomech.* 1980;13:989-1006.
7. Chappell JD, Kirdendall DT, Yu B, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med.* 2002;30:261-267.
8. DeHaven K, Lintner D. Athletic injuries: comparison by age, sport, and gender. *Am J Sports Med.* 1986;14:218-224.
9. 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.
10. Emerson RJ. Basketball injuries and anterior cruciate ligament. *Clin Sports Med.* 1993;12:317-328.
11. Feagin JA, Lambert KL. Mechanism of injury and pathology of anterior cruciate ligament injuries. *Orthop Clin North Am.* 1985;16:41-45.
12. Gottlob CA, Baker CL, Pellissier JM, Colvin L. Cost effectiveness of anterior cruciate ligament reconstruction in young adults. *Clin Orthop.* 1999;367:272-282.
13. Grood ES, Suntay WJ, Noyes FR, et al. Biomechanics of the knee-extension exercise. *J Bone Joint Surg Am.* 1994;66:725-734.
14. Gwinn DE, Wilckens JH, McDevitt ER, Ross G, Kao TC. The relative incidence of anterior cruciate ligament injury in men and women at the United State Naval Academy. *Am J Sports Med.* 2000;28:98-102.
15. Huston LJ, Wojtys EM. Neuromuscular performance characteristics in elite female athletes. *Am J Sports Med.* 1996;24:427-436.
16. Johnson RJ. Prevention of cruciate ligament injuries. In: Feagin JA, ed. *The Crucial Ligaments.* New York, NY: Churchill Livingstone; 1988:349-356.
17. Kadaba MP, Ramakrishnan HK, Wootten ME. Measurement of lower extremity kinematics during level walking. *J Orthop Res.* 1990;8:383-392.
18. Kadaba MP, Ramakrishnan HK, Wootten ME, Gainey J, Gorton G, Cochran GVB. Repeatability of kinematics, kinetics, and electromyographic data in normal adult gait. *J Orthop Res.* 1989;7:849-860.
19. Kannus P, Jarvinen M. Conservatively treated tears of the ACL: long-term results. *J Bone Joint Surg Am.* 1987;69:1007-1012.
20. Lindenfeld TN, Schmitt DJ, Hendy MP, et al. Incidence of injury in indoor soccer. *Am J Sports Med.* 1994;22:354-371.
21. Malinzak R, Cobby S, Kirkendail D, Yu B, Garrett W. A comparison of knee motion and electromyography patterns between men and women in selected athletic maneuvers. *Clin Biomech.* 2001;16:438-445.
22. Malone TR, Hardaker WT, Garrett WE, et al. Relationship of gender to anterior cruciate ligament injuries in intercollegiate basketball players. *J South Orthop Assoc.* 1993;2:36-39.
23. Markolf KL, Gorek JF, Kabo JM, et al. Direct measurement of resultant forces in the anterior cruciate ligament. *J Bone Joint Surg Am.* 1990;72:557-567.

24. McNair P, Marshall R, Matheson J. Important features associated with acute anterior cruciate ligament injury. *N Z Med J.* 1993;103:537-539.
25. Miyaskaka KC, Daniel DM, Stone ML, et al. The incidence of knee ligament injuries in the general population. *Am J Knee Surg.* 1991;4:3-8.
26. Noyes FR, Mooar PA, Matthews DS, et al. The symptomatic ACL-deficient knee. *J Bone Joint Surg Am.* 1983;65:154-174.
27. Nunley RM, Wright DW, Renner JB, Yu B, Garrett WE. Gender comparison of patella-tendon tibial shaft angle with weight-bearing. *Res Sports Med.* 2003;11:173-185.
28. Smidt JG. Biomechanical analysis of knee flexion and extension. *J Biomech.* 1973;6:79-92.
29. Smith BA, Livesay GA, Woo SLY. Biology and biomechanics of the anterior cruciate ligament. *Clin Sports Med.* 1993;12:637-666.
30. Traina SM, Bromberg DF. ACL injury patterns in women. *Orthopedics.* 1997;20:545-549.
31. vanEijden TMGJ, De Boer W, Weijs WA. The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion-extension angle. *J Biomech.* 1985;18:803-809.
32. Woo SLY, Lewis JL, Suh JK, Engebretsen L. Acute injury to ligament and meniscus as inducers of osteoarthritis. In: Kuettner KE, Goldberg VM, eds. *Osteoarthritic Disorders.* Rosemont, Ill: American Academy of Orthopaedic Surgeons; 1994:185-196.
33. Yu B, Gabriel D, Noble L, An KN. Estimate of optimum cutoff frequency for a low-pass digital filter. *J Appl Biomech.* 1999;15:318-329.
34. Yu B, Kienbacher T, Growney ES, Johnson ME, An KN. Reproducibility of the kinematics and kinetics of the lower extremity during normal stair-climbing. *J Orthop Res.* 1997;15:348-352.