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Knee and Hip Angle and Moment Adaptations During Cutting Tasks in Subjects With Anterior Cruciate Ligament Deficiency Classified as Noncopers

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Study Design: Two-factor mixed-design study, with factors including group (control and noncoper) and task (sidestep, crossover, and straight).

Objectives: To compare the knee and hip joint angles and moments of control subjects and subjects with an anterior cruciate ligament (ACL) deficient knee classified as noncopers, during a sidestep, crossover, and straight-ahead task.

Background: Subjects with ACL deficiency primarily note difficulty with cutting tasks as opposed to straight-ahead tasks. Yet, previous studies have primarily focused on straight-ahead tasks.

Methods and Measures: Fifteen subjects with ACL deficiency classified as noncopers, based on the number of giving-way episodes (>1) and global question of knee function (<60%), were included in this study. These subjects (10 male, 5 female; age range, 18-49 years) were compared to a healthy control group (7 male, 7 female; age range, 19-47 years). Position data collected at 60 Hz were combined with anthropometric and ground reaction force data collected at 420 Hz to estimate 3-dimensional knee and hip joint angles and moments. All subjects performed 3 tasks including a step and 45° sidestep cut, step and 45° crossover cut, and step and proceed straight. Two-way mixed-model ANOVAs were used to compare peak angle and moment variables between 10% to 30% of stance.

Results: The ACL-deficient noncoper group had 1.8° to 5.7° less knee flexion angle compared to the control group across tasks ($P<.043$). The ACL-deficient noncoper group used 22% to 27% lower knee extensor moment during weight acceptance compared to the control group ($P<.001$). The sagittal plane hip extensor moments were 34% to 39% higher in the ACL-deficient noncoper group compared to the control group ($P<.025$). Hip frontal ($P<.037$) and transverse plane ($P<.04$) moments also distinguished the ACL-deficient noncoper from the control group.

Conclusions: This study suggests that individuals who do not cope well after ACL injury rely on a hip control strategy during cutting tasks.

Key Words: *ACL, biomechanics, knee stability*

Current practice trends suggest that some patients with an anterior cruciate ligament (ACL) deficient knee can cope well with knee instability without surgical reconstruction, while other patients have difficulty.[8,9,13] Because most subjects who experience an ACL rupture intend to return to a high functional level,[6,8,9,33,42,50] most opt for surgical reconstruction to prevent damage to associated knee structures and improve function. A collection of clinical tests has shown the ability to discriminate subjects with ACL deficiency who are able to cope (return to sports) and those considered noncopers.[12] In a prospective study, a combination of self-report scores, knee extension strength tests, and functional tests combined to correctly predict 76% of subjects with ACL-deficient knees that were able to compete in sports without surgery.[12] Common explanations of the unique ability of some subjects with ACL deficiency to return to sports are movement and muscle activation patterns that limit the increased knee laxity associated with loss of the ACL.

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The following review boards approved the study protocol: University of Rochester, Research Subjects Review Board, Ithaca College, All College Review Board for Human Subjects Research; staff and patients of University Sports Medicine who participated in this study.
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Recent research suggests that movement patterns defined by lower extremity angles, moments, and muscle function may distinguish noncopers and copers during various tasks.[20,39-41] Rudolph et al[39-41] in a series of studies suggested that noncopers have lower peak knee flexion angles during weight acceptance (0%-30% of stance) and lower knee extensor moments. Electromyography suggests that hamstring/quadriceps cocontraction is seen in noncopers but not copers,[39-41,43] which supports the assertion that noncopers rely on compensation from the hip musculature. Ferber et al[11] noted that during walking, subjects who had more than 1 episode of giving way used a larger hip extensor moment during early stance compared to controls. In control subjects, Winter[46] described a covariance of the hip and knee moments where a higher hip extensor moment was associated with a lower knee extensor moment. Winter[46] attributed this moment pattern to the biarticular hamstring muscles, suggesting that this hip/knee moment pattern is consistent with a greater contribution of the hamstring muscles. Ferber et al[11] suggested that noncopers may adopt a higher hip extensor moment and lower knee extensor moment, consistent with the covariance pattern described by Winter.[46] It is uncertain whether the moment patterns associated with noncopers during straight-ahead tasks generalize to other more difficult cutting tasks. Because this moment pattern is associated with noncopers, the assumption is that altering this pattern through surgery or rehabilitation is desirable.

Theories implicate both sidestep and crossover cut tasks as problematic for subjects with ACL deficiency. Markolf et al[29] demonstrated that loading in the ACL is increased by knee internal rotation movements. Houck et al[20] demonstrated that the crossover cut results in 4° to 5° larger knee internal rotation angle, suggesting that this task is consistent with the ACL loading described by Markolf et al.[29] Fung et al[14] observed higher isometric knee external rotation strength compared to knee internal rotation strength in subjects with chronic ACL deficiency. Fung et al[14] suggested that higher knee external rotation strength was a compensation to control knee internal rotation. Collectively these studies support a hypothesis that subjects with ACL deficiency may have more difficulty with tasks that cause knee internal rotation, such as the crossover cut. In contrast, a sidestep cut task may result in knee external rotation.[1] In controlled studies it has been suggested that knee external rotation and abduction may cause the ACL to impinge on the lateral femoral condyle, possibly leading to rupture of the ACL.[49] Further, qualitative studies from videotapes of subjects rupturing their ACL implicate knee external rotation and abduction as an injury mechanism.[10,24,27] This gives rise to a competing hypothesis that subjects with ACL-deficiency may experience difficulty with knee external rotation rather than knee internal rotation. Previous studies of movement patterns in individuals with an ACL-deficient knee have not contrasted sidestep and crossover cut tasks.[20,39-41] Further, studies of subjects with an ACL deficient knee performing tasks associated with knee internal and external rotation are necessary to determine muscle control strategies that improve stability during cut tasks.

The purpose of this study was to compare the 3-D knee and hip angles, and moments of control subjects and ACL-deficient noncoper subjects during 3 stepping-down tasks: a step straight, step and 45° sidestep cut, and step and 45° crossover cut. Internal hip and knee moments reflect the minimal agonist contributions, assuming that the joint is not near end range during movement.[45] Differences of the net joint moments, therefore, represent different agonist demands. Similar to studies of straight-ahead tasks,[2,38-41] we hypothesized that noncopers would use lower knee flexion angles and extensor moments. At the hip we hypothesized that the noncopers would use greater hip extensor moments. In the transverse plane we hypothesized that noncopers would show lower knee and hip internal rotator moments during early stance to decrease the stress on the knee. We hypothesized the frontal plane knee and hip angles and moments would be similar between the noncopers and control subjects. Lastly, we hypothesized that the subjects classified as noncopers would demonstrate similar knee transverse plane angles as control subjects, which would suggest that the alterations in knee and hip moments were effective in maintaining knee stability.

METHODS

Subjects

A sample of convenience of 15 subjects with an ACL-deficient knee and 14 control subjects participated in this study (Table 1). The subjects with ACL deficiency included 7 left and 8 right sides, compared to 6 left sides and 8 right sides for the control group. The side selected for testing of the control group was determined using a random sequence. A power analysis using standard deviations from previous studies[20] suggested that samples of 14 subjects per group were sufficient to achieve 80% power. All subjects signed informed consent approved by the Internal Review Boards of Ithaca College and the University of Rochester. The control subjects were between 19 and 47 years of age, were free of lower extremity pain for at least 6 months, and had no previous history of knee injury.

All the subjects classified with ACL deficiency had a greater than 3-mm side-to-side difference on the KT-1000 test. Subjects were excluded if clinical varus/valgus laxity tests were positive or subjects had a

TABLE 1. Demographic and clinical variables (mean ± SD [range]).

Variable	Noncoper	Control
Sample description		
Age (y)	32.6 ± 8.6 (18.0-49.0)	28.4 ± 11.9 (19.0-47.0)
Height (m)	1.8 ± 0.1 (1.6-1.9)	1.7 ± 0.1 (1.6-1.9)
Mass (kg)	79.5 ± 18.0 (60.0-133.6)	71.8 ± 16.0 (54.1-109.2)
Gender (male/female)	10/5	7/7
Clinical		
Injury time (mo)	54.0 ± 68.0 (5.0-216.0)	
Knee laxity (KT-1000 test) (mm)*	4.6 ± 2.2 (3.0-10.0)	
Isometric strength†		
Knee extension (%)	97 ± 14 (75-120)	
Knee flexion (%)	96 ± 25 (54-160)	
Giving way (number since injury)	2.8 ± 1.8 (0.0-5.0)	
Functional ratings‡		
Global rating, overall (%)	67.9 ± 16.0 (35.0-95.0)	
Lysholm scale (%)[32]	80.5 ± 12.0 (54.0-97.0)	
Modified Noyes Questionnaire (%)[48]	75.4 ± 14.0 (45.0-92.0)	
Knee Outcome Survey (KOS), ADL Scale (%)[25]	90.0 ± 7.0 (75-100)	
Knee Outcome Survey (KOS), Sports Scale (%)[25]	61.2 ± 19 (16-92)	
Stride length (m)	1.5 ± 0.2 (1.2-1.9)	1.5 ± 0.1 (1.3-1.8)

* Side-to-side difference, greater laxity on the injured side.
†Involved/uninvolved × 100.
‡Higher scores indicate better function for all scales.

known meniscus involvement that led to surgery. In addition, a difference in knee girth of greater than 2 cm along the joint line, suggesting joint swelling, led to exclusion. Other exclusion criteria included painful knee active range of motion, a leg length discrepancy, and a history of lower extremity pain not related to ACL injury in the last 6 months. Subjects were included as a noncoper if they had more than 1 episode of giving way and/or rated themselves 60% or below on a global rating of knee function.[12] Subject responses to questionnaires[25,28,44] used to characterize their function, along with other clinical measures, are given in Table 1. The functional status of the noncoper group is reflected in the Knee Outcome Survey (KOS) ADL and Sports scales. These scales suggest that noncopers had less difficulty with activities of daily living (mean ± SD KOS-ADL score, 90% ± 7%), compared with more strenuous sports-related tasks (mean ± SD KOS-Sports, 61.2% ± 19%). All subjects with ACL deficiency were at least 5 months postinjury and, therefore, considered representative of a chronic condition.

Knee Isometric Torque

Knee flexor and extensor torque was assessed using a maximal isometric knee flexion/extension effort, with the knee positioned at 60° of flexion. In a seated position subjects were stabilized using the recommended protocol for a Lido MultiJoint II (model 940031-01; Loredon Biomedical, Inc, West Sacramento, CA). Each subject was given 3 warm-up trials prior to performing a maximum isometric effort. Subjects performed a 3-second maximum extension effort followed by a knee flexion effort. Subjects were given visual feedback and verbal encouragement during each contraction. The uninvolved side was tested first followed by the involved side. Control subjects were not tested.

Kinematics and Force Plate Recordings

The infrared diodes (IREDs) of the Optotrak Motion Analysis System (model 3020; Northern Digital, Inc, Waterloo, Ontario, Canada) were tracked at a sampling rate of 60 Hz. Ground reaction forces (GRFs) were recorded at a sampling rate of 420 Hz using a force plate (model 9865B; Kistler Instrument Corp, Amherst, NY) mounted flush with the floor of a 15-m walkway. The force (Fx, Fy, and Fz) and position data (x, y, z) were filtered at a cut-off frequency of 25 Hz and 7 Hz, respectively, using a fourth-order, low-pass Butterworth zero-phase lag filter.

Lower Extremity Modeling

A 4-segment model of the lower extremity, including the foot, leg, thigh, and pelvis, was used to estimate joint angles and moments in 3 dimensions. Rigid-body representations of each segment were achieved by placing 3 IREDs on each segment (Figure 1). The methods used to model the lower extremity are described in published studies[20,22] and are reviewed only briefly here. The IREDs used to represent the pelvis were placed on the right and left anterior superior iliac spine and a hollow aluminum rod extending from the sacrum. The femur was represented by 2 IREDs mounted on a femoral tracking device and a marker placed 10 cm distal to the greater trochanter. The IREDs used to represent the tibia were placed over the anterior border of the tibia. The IREDs used to track the foot were placed on the lateral side of the shoe proximal to the fifth metatarsal head. All subjects were required to wear low-top running-style shoes.

Subsequently, estimated segment inertial properties[48] were combined with the filtered GRF and position data to calculate net joint moments at the ankle, knee, and hip using the Kingait[3] software package (Mishac, Inc, Waterloo, Ontario),[26] which utilizes the same approach as previously published methods.[16,45,47] Net joint moments were subsequently resolved into the local coordinates of the distal segment. The Kingait[3] software package determines 3-dimensional joint angles, including (1) abduction/adduction angles that occur around an anterior/posterior axis, (2) internal/external rotation angles that occur around an inferior/superior axis, and (3) flexion/extension angles that occur around a medial lateral axis consistent with published protocols.[16,47] Evaluation of the femoral and tibia tracking approach used in this study showed root-mean-square errors of ±2° in the transverse plane in a single subject over the first 85% of stance during walking and running.[22]

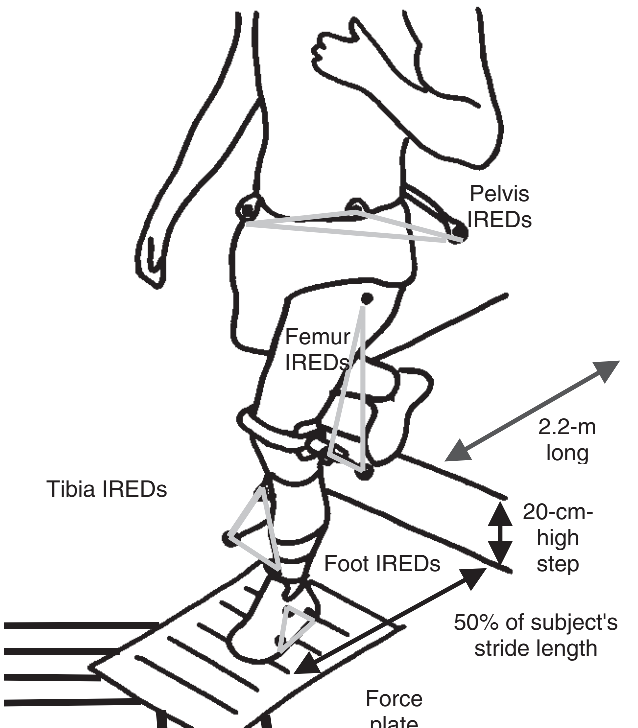

FIGURE 1. Infrared emitting diode (IRED) placements and relationship of the step and force plate. Reprinted from *Houck JR, Yack HJ. Associations of knee angles, moments and function among subjects that are healthy and anterior cruciate ligament deficient during straight ahead and crossover cutting activities. Gait & Posture. 2003:126- 138.*[20] With permission from Elsevier.

A previous study of the reliability of knee joint angles and moments suggested good repeatability (intraclass correlation coefficient greater than 0.8 and standard error of the measurement of 2°) during walking and step and 45° crossover cut activities used in a previous study.[23]

Procedures

Subjects completed 3 different activities: a straight-ahead task (ST), a crossover-cutting task (CC), and a sidestep-cutting task (SC) (Figure 2). The ST included walking and stepping down off a 21-cm platform. The step and 45° CC required subjects to step down off a 21-cm platform and turn 45° using a crossover cut movement (Figure 2). The step and 45° SC required subjects to step down off a 21-cm platform and turn 45° using a sidestep cut movement (Figure 2). The step platform was 2.3 m long, allowing 2 strides prior to stepping down and was positioned so that the distance from the edge of the platform to the center of the force plate was 50% of the subject's stride length during overground walking (Figure 1). The approach velocity was controlled (target velocity, 1.34 m/s) using an infrared timing system (Brower Timing Systems, Draper, UT). The foot-landing strategy (heel first) was manipulated to decrease variability across subjects in the knee kinematics and moments, and subsequently, to enhance power to detect group differences.[17] Colored tape placed adjacent to the force plate provided a target for subjects to achieve the desired cut angle. After at least 10 warm-up trials subjects completed 5 successful trials of each activity. The sequence of testing was randomized.

Four performance variables were collected to ensure that the tasks were performed similarly for each group, allowing differences between groups to be attributed to ACL deficiency. The performance variables included stepping velocity across stance, cut angle, foot position, and stance time. Average velocity across stance was calculated as the distance of the origin of the pelvis segment traveled in the transverse plane from heel strike to toe-off, divided by the stance time. The cutting angle for each task was determined from the peak medial/lateral (M/L) and anterior/posterior (A/P) GRF during late stance.[18] These methods used to calculate the cut angle are described in detail in a different publication.[18] Foot position was the transverse plane angle of the foot in the global coordinate system at foot flat. With respect to global coordinates system, negative values indicate foot external rotation and positive values indicate foot internal rotation. Stance time was determined from vertical GRF using a threshold of 10 N.

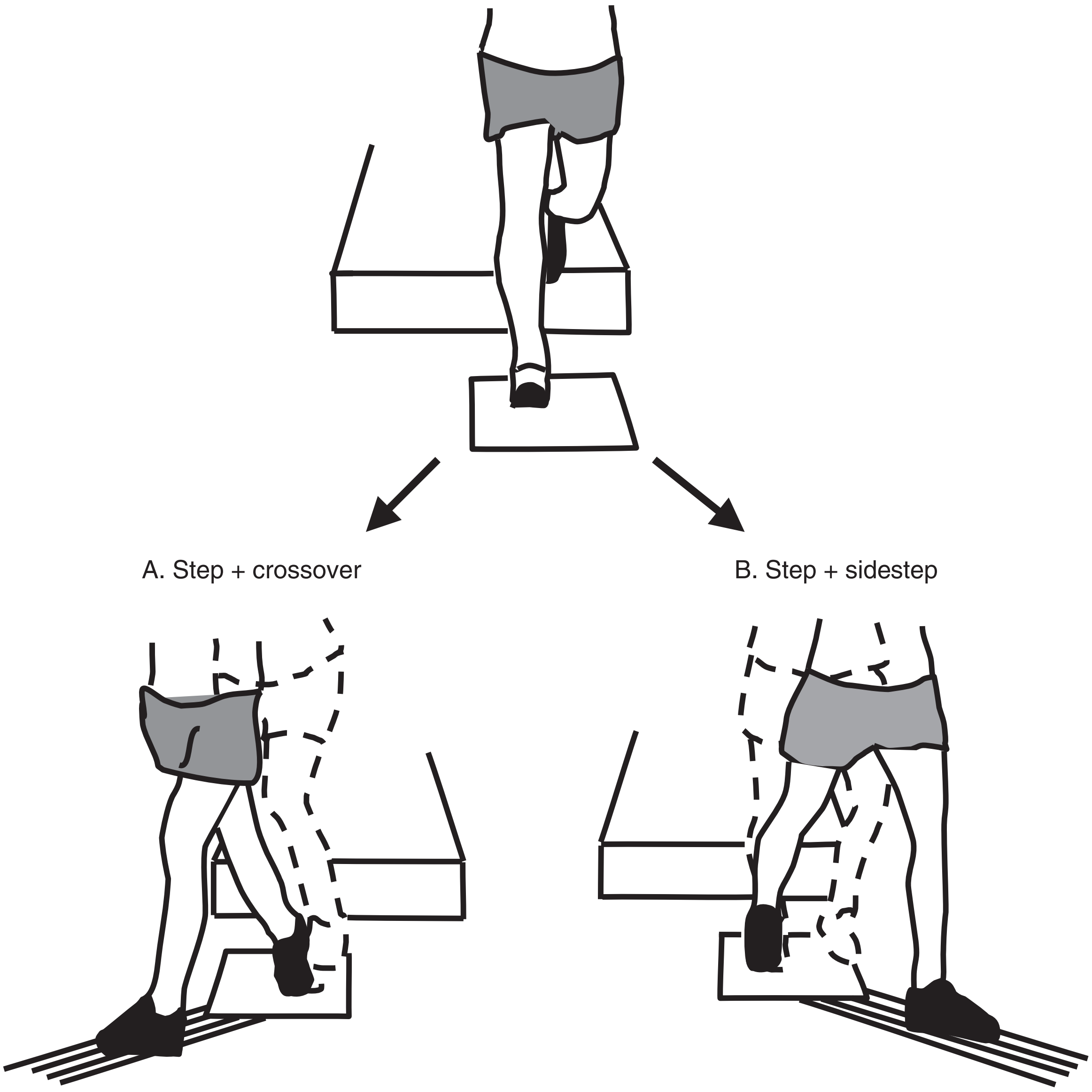

FIGURE 2. The step and 45° crossover cut (CC) and step and 45° sidestep cut (SC) tasks are illustrated above. Reprinted from *Houck JR. Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks. Journal of Electromyography & Kinesiology. 2003:545-554.*[18] With permission from Elsevier.

Analysis

Peak isometric knee torque of the involved side was compared to the uninvolved side using a paired t test with an alpha level of .05. The relative deficit in isometric knee strength was described by reporting the ratio of the involved/uninvolved × 100% for both knee flexion and knee extension. Knee and hip angle and moment patterns for the 5 trials were ensemble-averaged using linear interpolation at 2% intervals to gain a representative pattern for each subject across stance for each task. Because early stance is thought to challenge subjects with an ACL-deficient knee, peak knee and hip angles (3-D) and moments (3-D) were compared from 10% to 30% of stance using a mixed 2-way ANOVA model. One factor was group (fixed factor) with 2 levels, including noncopers (ACL deficient) and controls. The second factor was task (repeated factor), with 3 levels, including ST, CC, and SC.[34] Differences across tasks were ignored in this analysis because they were not related to our hypotheses relating to between-group comparisons (noncoper versus controls). An interaction effect was examined to determine if cutting tasks (SC and/or CC tasks) required greater angle and/or moment adaptations than an ST task. The 2-way ANOVA model was applied to each dependent variable separately using a probability value of less than .05 to indicate significance.

RESULTS

Knee Isometric Torque

There was no significant difference in the isometric knee strength of the involved side compared to the uninvolved side for peak knee extension (P = .311) or knee flexion (P = .39). However, there was a wide range of knee flexion and extension isometric torque across subjects (Table 1).

Performance

For all the performance variables (velocity [P = .505], cut angle [P = .278], foot position [P = .634], and stance time [P = .379]) there were no significant differences between groups. There were significant differences across tasks for mean velocity (range across tasks and groups, 1.33-1.53 m/s) and mean stance time (range across tasks and groups, 0.623-0.715 seconds). The ST task was significantly faster (1.54 ± 0.11 m/s) than both the SC (1.35 ± 0.08 m/s) and CC tasks (1.37 ± 0.14 m/s). The stance time was unique to each task, with the ST task having the shortest time and the CC task the longest time.

Knee and Hip Angles

The peak knee flexion angles were significantly different between the control and noncoper groups (P = .043), while the hip angles (Table 3) and knee angles (Table 2) in the frontal and transverse planes were not significantly different. The differences in peak knee flexion angles between groups did not depend on task (interaction effect, P = .189). The peak knee flexion achieved by the noncoper group was 1.8° to 5.7° lower than the control group from 10% to 30% of stance. There was a notable lack of a significant difference of the peak knee transverse plane angles between groups (P = .534).

Knee and Hip Moments

The sagittal plane knee and hip moments suggested significant differences between groups during early stance that did not depend on task (Tables 4 and 5). The knee extensor moment from 10% to 30% of stance was 0.35 to 0.54 Nm/kg lower (P<.001) in the noncoper group compared to the control group. Consistent with the sagittal plane knee moments during early stance, the hip extensor moment was 0.29 to 0.36 Nm/kg higher (P = .025) across all tasks for the noncoper group (Table 5). There were no significant interaction effects (P>.05), which demonstrates that the differences between groups did not depend on task for any knee or hip moment variable.

TABLE 2. Peak knee angles (in degrees) during 10% to 30% of stance (mean ± SD).

Plane	ST Task		SC Task		CC Task		*P* Value (Group)*
	Control	Noncoper	Control	Noncoper	Control	Noncoper	
Sagittal							
(+) Flexion	38.4 ± 5.2	36.3 ± 6.1	41.3 ± 6.2	35.6 ± 5.8	36.7 ± 5.6	34.9 ± 5.4	.043†
Frontal							
(+) Abduction	–1.9 ± 3.0	1.1 ± 4.7	–1.6 ± 2.8	1.0 ± 4.8	–1.7 ± 3.8	1.7 ± 4.6	.060
Transverse							
(+) Internal rotation	5.9 ± 2.5	4.5 ± 2.6	5.1 ± 2.2	5.7 ± 2.2	6.3 ± 1.5	5.9 ± 2.4	.534

Abbreviations: CC, crossover cut; ST, straight; SC, sidestep cut.
*The P values are the result of the mixed-design 2-way ANOVA (factors included task and group). Group is the main effect for group.
†The main effect for group was significant (P<.05). Task × group interactions were not significant (P>.05).

TABLE 3. Peak hip angles (in degrees) during 10% to 30% of stance (mean ± SD).

Plane	ST Task		SC Task		CC Task		*P* Value (Group)*
	Control	Noncoper	Control	Noncoper	Control	Noncoper	
Sagittal							
(+) Flexion	23.4 ± 4.9	22.4 ± 7.5	25.1 ± 3.5	23.8 ± 7.1	24.0 ± 3.8	25.1 ± 6.6	.838
Frontal							
(+) Abduction	9.2 ± 2.7	8.4 ± 4.0	6.7 ± 3.7	6.7 ± 3.0	7.3 ± 3.4	6.5 ± 5.1	.663
Transverse							
(+) Internal rotation	5.5 ± 3.1	3.9 ± 2.1	4.4 ± 4.7	3.1 ± 5.7	6.4 ± 6.8	8.0 ± 4.0	.727

Abbreviations: CC, crossover cut; ST, straight; SC, sidestep cut.
*The P values are the result of the mixed-design 2-way ANOVA (factors included task and group). Group is the main effect for group. The main effects for group were not significant (P>.05). Task × group interactions were not significant (P>.05).

TABLE 4. Peak knee moments (Nm/kg) during 10% to 30% of stance (mean ± SD).

Plane	ST Task		SC Task		CC Task		*P* Value (Group)*
	Control	Noncoper	Control	Noncoper	Control	Noncoper	
Sagittal							
(+) Flexion	1.71 ± 0.38	1.33 ± 0.24	2.00 ± 0.38	1.46 ± 0.38	1.56 ± 0.35	1.21 ± 0.29	.001†
Frontal							
(+) Abduction	0.79 ± 0.22	0.84 ± 0.23	0.92 ± 0.23	0.96 ± 0.26	0.57 ± 0.25	0.58 ± 0.16	.664
Transverse							
(+) Internal rotation	0.05 ± 0.03	0.05 ± 0.04	0.03 ± 0.02	0.04 ± 0.05	0.09 ± 0.05	0.07 ± 0.04	.884

Abbreviations: CC, crossover cut; ST, straight; SC, sidestep cut.
*The *P* values are the result of the mixed-design 2-way ANOVA (factors included task and group). Group is the main effect for group.
†The main effect for group was significant ($P < .05$). Task × group interactions were not significant ($P > .05$).

TABLE 5. Peak hip moments (Nm/kg) during 10% to 30% of stance (mean ± SD).

Plane	ST Task		SC Task		CC Task		*P* Value (Group)*
	Control	Noncoper	Control	Noncoper	Control	Noncoper	
Sagittal							
(+) Flexion	0.91 ± 0.35	1.27 ± 0.54	0.85 ± 0.36	1.14 ± 0.48	0.84 ± 0.38	1.15 ± 0.36	.025†
Frontal							
(+) Abduction	1.79 ± 0.26	1.57 ± 0.34	1.89 ± 0.27	1.70 ± 0.33	1.43 ± 0.38	1.15 ± 0.29	.037†
Transverse							
(+) Internal rotation	0.55 ± 0.12	0.47 ± 0.11	0.64 ± 0.12	0.52 ± 0.15	0.43 ± 0.12	0.35 ± 0.12	.040†

Abbreviations: CC, crossover cut; ST, straight; SC, sidestep cut.
*The *P* values are the result of the mixed design 2-way ANOVA (factors included task and group). Group is the main effect for group.
†The main effect for group was significant ($P < .05$). Task × group interactions were not significant ($P > .05$).

The transverse and frontal plane moments of the knee were not significantly different between groups (Table 4); however, the hip moments showed significant differences in all 3 planes. During early stance, the noncoper group used a smaller (P = .037) hip abductor moment by 0.19 to 0.28 Nm/kg (Table 5). Coupled with the changes in the sagittal and frontal planes, the hip internal rotator moments of the noncoper group were significantly smaller (P = .04) by 0.08 to 0.12 Nm/kg, compared to the control subjects.

DISCUSSION

The findings of this study suggest that individuals with ACL-deficient knees classified as noncopers demonstrate changes in hip and knee moments during contrasting cut tasks. The hypothesis that the noncoper group would show larger sagittal plane hip and knee moments during either cutting task compared to straight-ahead tasks (an interaction effect) was not supported. This occurred despite comparing contrasting cut tasks known to induce marked differences in muscle activation.[18] The sagittal plane hip and knee moments of the noncoper group were similar to differences described in previous studies of straight-ahead tasks.[11,39-41] The lower knee angles of the noncopers compared to control subjects were also similar to previous studies of straight-ahead tasks.[11,39-41] As hypothesized the hip internal rotator moment during early stance depended on group. Interestingly, the noncoper subjects decreased their hip internal rotator moment during early stance; however, this was not coupled with a change in the knee transverse plane moment. In addition, the ACL-deficient group was similar to the control subjects for transverse plane knee angles across tasks, suggesting that they controlled their knee motion.

Studies of knee transverse plane motion are controversial due to the known errors related to soft tissue artifact that obscures true joint angle data.[5,35,36] Studies comparing bone-mounted markers to skin mounted markers suggest large errors (maximum errors can exceed 10°) in the knee transverse and frontal planes with lower errors in the sagittal plane.[5,35,36] The errors are believed to result from soft tissue deformation relative to the underlying bone.[5,35,36] Previous studies[7,15,32,31,37] reporting tibiofemoral transverse and frontal plane angles using skin-mounted markers are difficult to interpret in light of these errors. This study used a method that showed low errors (<3°) in a single subject when knee motions were compared to bone-mounted markers.[22] The methods used in this study suggest that mild giving-way episodes are characterized by 3° to 4° increases in tibiofemoral internal rotation and increases in tibiofemoral anterior translation instability.[21,23] Brandsson et al[4] compared knee tibiofemoral kinematics using tantalum balls placed in the tibia and femur during a step-up task before and after ACL reconstruction. Brandsson et al[4] found little difference in tibiofemoral knee kinematics except at full knee extension. The results of the present study are therefore best interpreted as suggesting that subjects classified as ACL deficient and noncopers use compensations that are effective in controlling knee internal rotation, except during giving-way episodes.

The differences in sagittal plane knee angles and moments are consistent with previous studies concentrating on straight-ahead tasks. Previous studies[11,39-41] focusing on walking and running suggest that noncopers use a lower knee extensor moment and lower knee flexion angle. New to this study is the observation that this same pattern—decreased knee flexion angle by approximately 2° to 6° and decreased knee extensor moment by 22% to 27%—was utilized during cutting tasks. Differences in peak knee flexion of less than 5°, observed during the ST and CC tasks, are subtle and potentially less meaningful clinically than the approximately 6° decrease observed during the SC task. Coincident with the changes in knee angle and moment, the hip extensor moment during early stance was 34% to 39% higher for the noncopers. Previous studies of straight-ahead tasks reported a similar hip extensor adaptation associated with a noncoper group during walking.[11,39] Winter et al[46] suggested that, if the hip extensor moments increased, the knee moments would shift toward a flexor moment. Our results, which show an increase in the hip extensor and a decrease in the knee extensor moments, are consistent with this pattern.[46] However, it is unclear if the patterns associated with the noncopers in this study were acquired or existed before the injury.

The hip fontal and transverse plane moments were different between groups during early stance. Previous studies suggest that hip frontal plane moments are associated with trunk position and balance control.[30] The 10% to 20% lower hip frontal plane moment observed in the noncoper group suggests less contribution of the hip abductors, which could result from a shift in the trunk center of mass laterally.[30] Coupled with the change in the hip abductor moment during early stance, the internal rotator moment is also lower by 15% to 20% for the noncopers. The noncopers used a smaller hip internal rotator moment across tasks to achieve similar hip angles as the individuals in the control group. The hip internal rotator moment during early stance is responsible for rotating the trunk and pelvis toward the new direction of travel. The hip compensation of the noncoper group is an increased hip extensor moment, which is partially attributable to the gluteus maximus muscle, a strong hip external rotator. Employing a larger hip extensor moment to compensate at the knee may result in a diminished antagonist hip internal rotator moment during early stance. This theory is consistent with the argument that compensations distally (knee) impact proximal muscle control (hip) during dynamic tasks. Clinicians therefore are encouraged to examine both proximal and distal movement patterns after ACL injury for compensations.

The direct impact of alterations in hip and knee transverse plane moments on knee stability is not clear because the net joint moments are unable to distinguish the individual contributions of muscles.[45] Therefore the individual contributions of the hip internal rotators, including the gluteus medius, gluteus minimus, and tensor fascia lata, are uncertain. Because the knee joint transverse plane moments were not unique to the noncopers, a direct association between hip and knee moments in the transverse plane is not possible with data from this study.

The consistency of the velocity, cut angle, foot position, and stance time achieved across groups suggests the comparisons in this study are not confounded by these performance variables. The faster speed recorded for the straight-ahead tasks were similar between the groups, hence, did not affect the group comparison (control versus ACL-deficient groups). However, the differences in speed may affect the differences between tasks and should be taken into account in future studies. The step-and-cut tasks in this study required 30% less knee extensor moments than those required by running-and-sidestep-cutting tasks performed by subjects in a previous study using the same methods.[17] The step-and-cut task requires a landing followed by a cut that theoretically matches the subjective descriptions of the type of task individuals with ACL deficiency suggest is difficult. However, this task is slower than other athletic tasks, such as running and cutting, and therefore may not generalize to faster activities.

Clinical Implications

The results of this study imply that noncopers may depend on a single movement pattern (hip strategy) during a variety of tasks. In the present study the increased hip extensor moments (hip strategy) did not depend on whether the challenge was toward knee internal rotation (crossover cut) or external rotation (sidestep cut). In addition, knee isometric extension torque was high in this group of noncopers (Table 1), suggesting that weakness of the knee extensors was not a strong contributor to the hip pattern adopted by these subjects. The success of a common movement pattern in maintaining knee stability is surprising during such distinct tasks. Progressing individuals to employ more than a single compensatory pattern is potentially important for return to sports, during which a variety of quick movements are expected. For example this finding reinforces the clinical emphasis on utilizing cutting tasks as part of a rehabilitation program seeking to assist subjects in learning motor control compensations. Further research is needed to explore whether hip extensor compensation can be minimized and an alternative stability strategy can be learned. For example, would inducing hip internal rotator moments inhibit the hip extensor compensation identified in the noncopers in this study? Further research is needed to identify the potential for noncopers to use alternative muscle control strategies that would enable them to perform as successfully as subjects identified as copers (ie, ankle strategy).

CONCLUSION

These data extend previous studies by suggesting that the alterations in knee angles and moments attributed to noncopers during straight-ahead tasks also apply to cutting tasks. In addition, the higher hip moments previously associated with subjects classified as noncopers during straight-ahead tasks were also observed in this study, further supporting the premise that noncopers may rely more on a hip control strategy. Unique to this study is the observation that the noncopers also altered their frontal and transverse plane hip moments; however, this was not associated with changes in knee moments in the frontal and transverse planes. Further studies are needed to assess whether changes in transverse and frontal plane hip moments affect knee stability.

REFERENCES

1. Andrews JR, McLeod WD, Ward T, Howard K. The cutting mechanism. *Am J Sports Med*. 1977;5:111-121.
2. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. *J Bone Joint Surg Am*. 1990;72:871-877.
3. Besier TF, Lloyd DG, Cochrane JL, Ackland TR. External loading of the knee joint during running and cutting maneuvers. *Med Sci Sports Exerc*. 2001;33:1168-1175.
4. Brandsson S, Karlsson J, Sward L, Kartus J, Eriksson BI, Karrholm J. Kinematics and laxity of the knee joint after anterior cruciate ligament reconstruction: pre- and postoperative radiostereometric studies. *Am J Sports Med*. 2002;30:361-367.
5. Cappozzo A, Catani F, Leardini A, Benedetti MG, Croce UD. Position and orientation in space of bones during movement: experimental artefacts. *Clin Biomech (Bristol, Avon)*. 1996;11:90-100.
6. Ciccotti MG, Lombardo SJ, Nonweiler B, Pink M. Non-operative treatment of ruptures of the anterior cruciate ligament in middle-aged patients. Results after long-term follow-up. *J Bone Joint Surg Am*. 1994;76:1315-1321.
7. Cross MJ, Gibbs NJ, Bryant GJ. An analysis of the sidestep cutting manoeuvre. *Am J Sports Med*. 1989;17:363-366.
8. Daniel DM, Stone ML, Dobson BE, Fithian DC, Rossman DJ, Kaufman KR. Fate of the ACL-injured patient. A prospective outcome study. *Am J Sports Med*. 1994;22:632-644.
9. Eastlack ME, Axe MJ, Snyder-Mackler L. Laxity, instability, and functional outcome after ACL injury: copers versus noncopers. *Med Sci Sports Exerc*. 1999;31:210-215.
10. Ebstrup JF, Bojsen-Moller F. Anterior cruciate ligament injury in indoor ball games. *Scand J Med Sci Sports*. 2000;10:114-116.
11. Ferber R, Osternig LR, Woollacott MH, Wasielewski NJ, Lee JH. Gait perturbation response in chronic anterior cruciate ligament deficiency and repair. *Clin Biomech (Bristol, Avon)*. 2003;18:132-141.
12. Fitzgerald GK, Axe MJ, Snyder-Mackler L. A decision-making scheme for returning patients to high-level activity with nonoperative treatment after anterior cruciate ligament rupture. *Knee Surg Sports Traumatol Arthrosc*. 2000;8:76-82.
13. Fitzgerald GK, Axe MJ, Snyder-Mackler L. The efficacy of perturbation training in nonoperative anterior cruciate ligament rehabilitation programs for physical active individuals. *Phys Ther*. 2000;80:128-140.
14. Fung DT, Zhang LQ. Modeling of ACL impingement against the intercondylar notch. *Clin Biomech (Bristol, Avon)*. 2003;18:933-941.
15. Georgoulis AD, Papadonikolakis A, Papageorgiou CD, Mitsou A, Stergiou N. Three-dimensional tibiofemoral kinematics of the anterior cruciate ligament-deficient and reconstructed knee during walking. *Am J Sports Med*. 2003;31:75-79.
16. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng*. 1983;105:136-144.
17. Hanford S, Houck JR. Foot landing strategies affect the lower extremity transverse plane angles and moments during a sidestep cutting task [abstract]. *Proceedings Gait and Clinical Movement Analysis Society Annual Meeting*. Chattanooga, TN: Gait and Clinical Movement Analysis Society; April, 2002.
18. Houck J. Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks. *J Electromyogr Kinesiol*. 2003;13:545-554.
19. Houck J, Lerner A, Gushue D, Yack HJ. Self-reported giving-way episode during a stepping-down task: case report of a subject with an ACL-deficient knee. *J Orthop Sports Phys Ther*. 2003;33:273-282; discussion 283-276.
20. Houck J, Yack HJ. Associations of knee angles, moments and function among subjects that are healthy and anterior cruciate ligament deficient (ACLD) during straight ahead and crossover cutting activities. *Gait Posture*. 2003;18:126-138.
21. Houck J, Yack HJ. Giving way event during a combined stepping and crossover cutting task in an individual with anterior cruciate ligament deficiency. *J Orthop Sports Phys Ther*. 2001;31:481-489; discusssion 490-485.
22. Houck J, Yack HJ, Cuddeford T. Validity and comparisons of tibiofemoral orientations and displacement using a femoral tracking device during early to mid stance of walking. *Gait Posture*. 2004;19:76-84.

23. Houck JR, Yack HJ. Reliability of knee angles and kinetics during a stepping and cutting activity [abstract]. *Gait Posture*. 2000;11:149-150.
24. Ireland MK. Anterior cruciate ligament injury in female athletes: epidemiology. *J Athl Train*. 1999;34:150-154.
25. Irrgang JJ, Snyder-Mackler L, Wainner RS, Fu FH, Harner CD. Development of a patient-reported measure of function of the knee. *J Bone Joint Surg Am*. 1998;80:1132-1145.
26. Ishac M. KinGait3. Waterloo, Ontario, Canada: University of Waterloo; 1995. 27. Kirkendall DT, Garrett WE, Jr. The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clin Orthop Relat Res*. 2000;64-68.
28. Lysholm J, Gillquist J. Evaluation of knee ligament surgery results with special emphasis on use of a scoring scale. *Am J Sports Med*. 1982;10:150-154.
29. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res*. 1995;13:930-935.
30. McKinnon CD, Winter DA. Control of whole body balance in the frontal plane during human walking. *J Biomech*. 1993;26:633-644.
31. McLean SG, Myers PT, Neal RJ, Walters MR. A quantitative analysis of knee joint kinematics during the sidestep cutting maneuver. Implications for non-contact anterior cruciate ligament injury. *Bull Hosp Jt Dis*. 1998;57:30-38.
32. McLean SG, Neal RJ, Myers PT, Walters MR. Knee joint kinematics during the sidestep cutting maneuver: potential for injury in women. *Med Sci Sports Exerc*. 1999;31:959-968.
33. Noyes FR, Mooar PA, Matthews DS, Butler DL. The symptomatic anterior cruciate-deficient knee. Part I: the long-term functional disability in athletically active individuals. *J Bone Joint Surg Am*. 1983;65:154-162.
34. Portney LG, Watkins MP. *Foundations of Clinical Research: Applications to Practice.* 2nd ed. Upper Saddle River, NJ: Prentice Hall Health; 2000.
35. Reinschmidt C, Van den Bogert AJ, Lundberg A, et al. Tibiofemoral and tibiocalcaneal motion during walking: external vs. skeletal markers. *Gait Posture*. 1997;6:98-109.
36. Reinschmidt C, van den Bogert AJ, Nigg BM, Lundberg A, Murphy N. Effect of skin movement on the analysis of skeletal knee joint motion during running. *J Biomech*. 1997;30:729-732.
37. Ristanis S, Giakas G, Papageorgiou CD, Moraiti T, Stergiou N, Georgoulis AD. The effects of anterior cruciate ligament reconstruction on tibial rotation during pivoting after descending stairs. *Knee Surg Sports Traumatol Arthrosc*. 2003;11:360-365.
38. Roberts CS, Rash GS, Honaker JT, Wachowiak MP, Shaw JC. A deficient anterior cruciate ligament does not lead to quadriceps avoidance gait. *Gait Posture*. 1999;10:189-199.
39. Rudolph KS, Axe MJ, Buchanan TS, Scholz JP, Snyder-Mackler L. Dynamic stability in the anterior cruciate ligament deficient knee. *Knee Surg Sports Traumatol Arthrosc*. 2001;9:62-71.
40. Rudolph KS, Axe MJ, Snyder-Mackler L. Dynamic stability after ACL injury: who can hop? *Knee Surg Sports Traumatol Arthrosc*. 2000;8:262-269.
41. Rudolph KS, Eastlack ME, Axe MJ, Snyder-Mackler L. 1998 Basmajian Student Award Paper: Movement patterns after anterior cruciate ligament injury: a comparison of patients who compensate well for the injury and those who require operative stabilization. *J Electromyogr Kinesiol*. 1998;8:349-362.
42. Scavenius M, Bak K, Hansen S, Norring K, Jensen KH, Jorgensen U. Isolated total ruptures of the anterior cruciate ligament—a clinical study with long-term follow-up of 7 years. *Scand J Med Sci Sports*. 1999;9:114-119.
43. Steele JR, Brown JM. Effects of chronic anterior cruciate ligament deficiency on muscle activation patterns during an abrupt deceleration task. *Clin Biomech (Bristol, Avon)*. 1999;14:247-257.
44. Wilk KE, Romaniello WT, Soscia SM, Arrigo CA, Andrews JR. The relationship between subjective knee scores, isokinetic testing, and functional testing in the ACL-reconstructed knee. *J Orthop Sports Phys Ther*. 1994;20:60-73.
45. Winter DA. *Biomechanics and Motor Control of Human Movement.* 2nd ed. New York, NY: Wiley & Sons, Inc; 1990.
46. Winter DA. Human balance and posture control during standing and walking. *Gait Posture*. 1995;4:193-214.
47. Wu G, Cavanagh PR. ISB recommendations for standardization in the reporting of kinematic data. *J Biomech*. 1995;28:1257-1261.
48. Yeadon MR, Morlock M. The appropriate use of regression equations for the estimation of segmental inertia parameters. *J Biomech*. 1989;22:683-689.
49. Zhang LQ, Nuber GW, Bowen MK, Koh JL, Butler JP. Multiaxis muscle strength in ACL deficient and reconstructed knees: compensatory mechanism. *Med Sci Sports Exerc*. 2002;34:2-8.
50. Zysk SP, Refior HJ. Operative or conservative treatment of the acutely torn anterior cruciate ligament in middle-aged patients. A follow-up study of 133 patients between the ages of 40 and 59 years. *Arch Orthop Trauma Surg*. 2000;120:59-64.

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2. Jeff R. Houck, Kenneth E. De Haven, Mike Maloney. 2007. Influence of Anticipation on Movement Patterns in Subjects With ACL Deficiency Classified as Noncopers. *Journal of Orthopaedic & Sports Physical Therapy* **37**:2, 56-64. [Abstract] [PDF] [PDF Plus]
3. Gregory D. Myer, Mark V. Paterno, Kevin R. Ford, Carmen E. Quatman, Timothy E. Hewett. 2006. Rehabilitation After Anterior Cruciate Ligament Reconstruction: Criteria-Based Progression Through the Return-to-Sport Phase. *Journal of Orthopaedic & Sports Physical Therapy* **36**:6, 385-402. [Abstract] [PDF] [PDF Plus]