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Movement patterns after anterior cruciate ligament injury: a comparison of patients who compensate well for the injury and those who require operative stabilization

Katherine S. Rudolph [a], Marty E. Eastlack [b], Michael J. Axe [c,d], Lynn Snyder-Mackler [e,*]

[a] *Doctoral Candidate, Department of Physical Therapy, 303 McKinly Lab, University of Delaware, Newark, DE 19716, USA*
[b] *Assistant Professor, Department of Physical Therapy, Beaver College, 450 S. Easton Rd., Glenside, PA 19038-3295, USA*
[c] *Orthopedic Surgeon, First State Orthopedics, 4745 Ogletown-Stanton Rd., Suite 225, Newark, DE 19713, USA*
[d] *Clinical Associate Professor, Department of Physical Therapy, University of Delaware, Newark, DE 19716, USA*
[e] *Associate Professor, Department of Physical Therapy, 309 McKinly Lab, University of Delaware, Newark, DE 19716, USA*

Abstract

The purpose of this study was to describe kinematic and kinetic differences between a group of ACL deficient subjects who were grouped according to functional ability. Sixteen patients with complete ACL rupture were studied; eight subjects had instability with activities of daily living (*non-copers*) and eight subjects had returned to all pre-injury activity without limitation (*copers*). Three-dimensional joint kinematics and kinetics were collected from the knee and ankle during walking, jogging and going up and over a step. Results showed that both groups mitigated the force with which they contacted the floor but non-copers consistently demonstrated less knee flexion in the involved limb. The copers used joint kinematics similar to those of their uninvolved knees and similar to knee motions reported in uninjured subjects. The reduced knee motion in the involved knee of the non-copers did not correlate directly with quadriceps femoris muscle weakness.

The data suggest that the non-copers utilize a stabilization strategy which stiffens the knee joint which not only is unsuccessful but may lead to excessive joint contact forces which have the potential to damage articular structures. The copers use a strategy which permits normal knee kinematics and bodes well for joint integrity.

Keywords: Anterior cruciate ligament; Knee; Gait

1. Introduction

The principal function of the anterior cruciate ligament is to prevent anterior translation of the tibia relative to the femur. Anterior cruciate ligament rupture typically results in loss of knee joint stability, strength of the surrounding musculature, and function [6,15]. Patients often complain of the knee giving way after anterior cruciate ligament rupture, a symptom of instability. Patients usually require reconstructive surgery to reestablish functional stability of the knee [2,5,19]. Several recent studies, however, have reported successful outcomes after non-operative management in patients who have ruptured the anterior cruciate ligament, but generally, only after they have adapted their lifestyles by mitigating activity levels [1,2,4,5].

There are anterior cruciate ligament deficient individuals who can maintain high activity levels, experiencing neither instability, loss of function or weakness despite complete rupture of the anterior cruciate ligament [5,18]. The knee does not give way, even under stressful conditions like jumping and pivoting. These individuals are able to return to all pre-injury activities without reconstructive surgery [5] and often without the use of a brace

* Corresponding author: Tel: (302) 831-3613; Fax: (302) 831-4234; E-mail: smack@brahms.udel.edu

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[7]. We have categorized these people as *copers*, because they appear to have either an intrinsic, or rapidly developed mechanism of compensating for the ruptured ligament.

Historically, the effects of anterior cruciate ligament rupture on muscle strength and function have been compared to those of healthy subjects without regard for how well or how poorly the patients compensate for the ruptured ligament; inclusion in these studies involved the ligament rupture alone (e.g. ‘All subjects were anterior cruciate ligament deficient’). Studies involving free speed walking have shown mixed results. Some authors report significant abnormalities [1,28], whereas others do not [9,12,16,27]. Most authors agree that measurable disturbances in kinetics, kinematics and patterns of muscle activation exist during more stressful activities [4,9,12,28]; however, there is no consensus about what constitutes a typical response to stressful activities in the face of anterior cruciate ligament deficiency [2,3,8]. *Copers* comprise a small percentage of the patient population with anterior cruciate ligament rupture but their ability to stabilize the ACL deficient knee during high level activities suggests that their movement patterns are different from those who cannot stabilize their knees. Failure to distinguish the copers from non-copers in studies of ACL deficient individuals may have resulted in the inconsistencies in movement patterns found in the literature.

It was expected that the *non-copers* in our sample would demonstrate gait abnormalities much like those previously reported in the literature. Since *copers* seem to function as if their ligaments were still intact we anticipated that their gait patterns would be more like those reported in healthy subjects [18]. We also predicted that to maintain normal gait patterns, copers would have stronger quadriceps femoris muscles than the non-copers. The purpose of this study was to describe the kinematic and kinetic differences between a group of subjects who compensated well for anterior cruciate ligament injury (*copers*) and a group who did not (*non-copers*).

2. Procedures and instrumentation

2.1. Subjects

Sixteen subjects with complete rupture of the anterior cruciate ligament (documented via arthroscopy or magnetic resonance imaging), served as subjects for this study. Eight had reported instability with activities of daily living and were scheduled for reconstructive surgery; these subjects were categorized as non-copers. Eight subjects met our operational definition of *coper* by having returned to all pre-injury activity without limitation and by rating their current level of knee function as 85% or greater as compared to their pre-injury level of function. The copers had reported one or fewer episodes of giving way since their injuries. The mean age of the non-copers was 28 years old (± 9 years), and of copers was 31 years old (± 9 years). The mean time from injury to testing was 16.9 months (± 20.6 months) for the non-copers and 66.3 months (± 73 months) for the copers. All subjects underwent a test of anterior tibiofemoral laxity using a KT-2000 arthrometer, using maximum manual anterior force with the knee flexed 20–30 degrees over a bolster (Lachman’s position). All subjects had side-to-side maximum manual Lachman test differences of more than three millimeters, indicating a complete ACL rupture [5]. Difference in side to side laxity measurement averaged 6.7 mm (± 2.3 mm) for the non-copers and 6.4 mm (± 3.3 mm) for the copers. All subjects had an uninvolved knee that was healthy, had full knee range-of-motion bilaterally, and had no concomitant ligamentous injury to the involved knee. No subject had a knee effusion at the time of testing.

2.2. Strength testing

The maximum voluntary isometric contraction force of the subject’s quadriceps femoris muscles was determined using a burst-superimposition technique described previously [23,24]. The subjects were seated and stabilized in an isokinetic dynamometer (KIN COM, Chattanooga Group, Inc., Chattanooga, TN) with the hips and knees flexed 90°. Pre-gelled, self adhesive, 4×6 inch electrodes were placed over the proximal aspect of the vastus lateralis, and distal aspect of the vastus medialis. A supramaximal burst of electrical stimulation (100 pulses per second, 600 microsecond pulse duration, 10-pulse tetanic train) was superimposed on a maximum voluntary isometric contraction, and the force was recorded on the dynamometer. When a true maximum voluntary contraction is performed, no increase in torque is seen superimposed on the maximal contraction force. If an increase in torque was observed the test was repeated up to 4 times, with a 2 minute rest between tests. All subjects were able to perform this test properly within the 4 trials. This technique gives an accurate and reliable measure of isometric force producing capability.

2.3. Motion analysis

Motion analysis was accomplished using a passive, three dimensional motion analysis system (VICON, Oxford Metrics, London, England). Five cameras were calibrated to a calibration volume of 1.67 cubic meters. The cameras provided a field of view of the stance phase of gait (heel strike to toe off). Calibration errors were held below three millimeters. The kinematic variables of interest were collected at a sampling frequency of

120 Hz and recorded on a VAX Workstation 3100, Model 48 (Digital Equipment Corporation, Boston, MA). Kinematic data were acquired using the AMASS™ data acquisition software (VICON, Oxford Metrics, London, England). Kinetic data were collected using a six-component force platform (Bertec Corporation, Worthington, OH) and integrated with the kinematic data using ADG (ADG, ADTECH, Adelphi, MD), which is an analog data capture program. Force data were sampled at 480 Hz.

Data were collected unilaterally. Small rigid thermoplastic shells with three or four retro-reflective markers per segment were attached to the shank, thigh and foot of one limb. Anatomical calibration markers were placed on the greater trochanter, lateral femoral condyle and lateral malleolus to locate the hip, knee and ankle joint centers. An additional marker was placed on the athletic shoes at the level of the fifth metatarsal head to estimate the distal end of the foot segment and a standing calibration file was collected. These anatomical markers were subsequently removed and data collection began. Five trials of each of the following conditions were collected with a three minute rest interval between each activity: 1) free-speed walking; 2) free-speed jogging; and 3) ascending and descending a ten inch step. This procedure was then repeated with the other limb.

Subjects walked and jogged along a thirteen meter walkway at a self-selected speed which was calculated using the time measured by photoelectric beams located 286.5 centimeters apart along the walkway. Velocity was held to within 5% on all trials. Subjects were allowed several practice trials to ensure foot contact with the center of the force plate without the subjects adjusting their stride length to hit the force plate (targeting). A more stressful step activity involving stepping up and over a 25 cm step was performed (Fig. 1). For step trials, the step was positioned approximately 60 cm from the force platform. The subject stood facing the step at a self-selected distance and was told to step up with one limb (referred to as the 'supporting limb') and step over the step with the opposite limb (referred to as the 'step-over limb'). It was the 'step-over limb' which contacted the force plate.

For walking and jogging trials, data collection began just prior to the foot hitting the force platform. At least one stance phase was collected per walking and jogging trial. The contact phase (initial contact to toe off for walking and jogging) of five trials were normalized to 100% of stance and averaged for each condition. Data collection for the step trials began prior to the subject stepping up and ended when the contralateral limb progressed off of the force plate. Knee angles were chosen at discrete points in the activity and averaged over the five trials.

2.4. Data management

Laxity measurements are reported as the difference between laxity in the uninvolved limb and the involved limb measured by a KT-2000 arthrometer using a maximum manual Lachman test. The strength of the involved limb is reported as a Quadriceps Index which is defined as the involved maximum voluntary isometric contraction (*MVIC*) divided by the *MVIC* measured on the uninvolved limb. This is converted to a percentage and represents the involved limb's strength as a percent of the strength of the uninvolved limb.

Kinetic and kinematic data were filtered (kinetic data at forty-five hertz; kinematic data at six hertz) and initial contact and toe off of the stance phase were identified for walk and jog trials (Events, NIH Biomechanics Laboratory, Bethesda, MD). Kinematic and kinetic data were calculated using Move3d rigid-body analysis software (Move3d, NIH Biomechanics Laboratory, Bethesda, MD) which calculates six degrees of freedom (three translations and three rotations) in an inertial reference system. Segment coordinate systems for the thigh, shank and foot segments are generated from the shell markers and oriented in the laboratory using the anatomical markers collected during the standing trial. Move3d uses

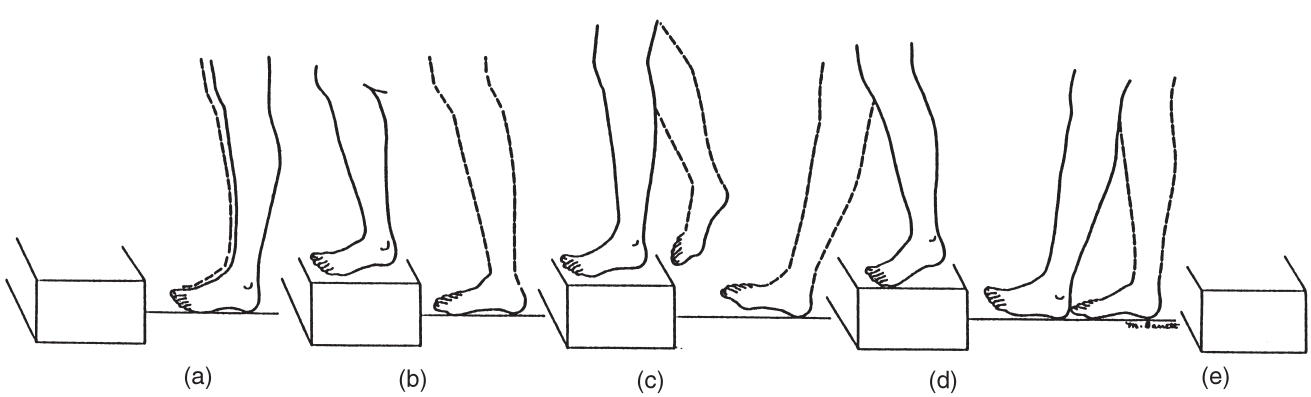

Fig. 1. Step trial variables. (a) Starting position; (b) Peak knee flexion during ascent; (c) Maximum knee extension on the step; (d) Knee angle of supporting limb at heel strike of step-over limb.

Euler angles to calculate three dimensional angular motions between segments, and joint kinetics are calculated using a six degree of freedom, inverse dynamics model as described in detail by Holden et al. [10]. Only sagittal plane kinematics and kinetics were reported since it was thought that the knee flexion angle would be most affected by ACL rupture. Velocity measurements were normalized to leg length, as measured from the greater trochanter to the ground, with shoes on and are reported in meters/second/leg length. The vertical ground reaction forces were normalized to body weight in Newtons and reported as per cent body weight. Calculated kinetic data were normalized to body mass in kilograms. Net internal moments are reported in Newtons $\times$ meter/kilogram and net muscle powers are reported in Watts/kilogram. The height of our step remained constant regardless of the height of the subject. In order to account for differences in leg length during step trials, knee angle during step up and knee angle at initial contact were also normalized by leg length and are reported as degrees $\times$ leg length.

With the exception of initial contact (IC), sub-phases of the gait cycle represent an interval of time [20]. In order to compare movement patterns between subjects, discrete kinematic events of the gait cycle were chosen based on knee position, since the knee joint position would be likely to show the greatest differences in the light of the ACL deficiency. The discrete events, shown in Fig. 2, included initial contact, peak knee flexion during loading response (*LR*) and peak knee extension during mid-stance (*M*st). Variables analyzed during the early stance for walking can be seen in Table 1. The peak vertical ground reaction force during weight acceptance was also used to determine the force with which the subjects contacted the ground.

Only kinematic data and vertical ground reaction forces were available for analysis for jogging and step trials, due to difficulties encountered during data collection. The discrete events for jog trials included initial contact and peak knee flexion during stance and peak knee extension during stance. Jog variables can be seen in Table 2.

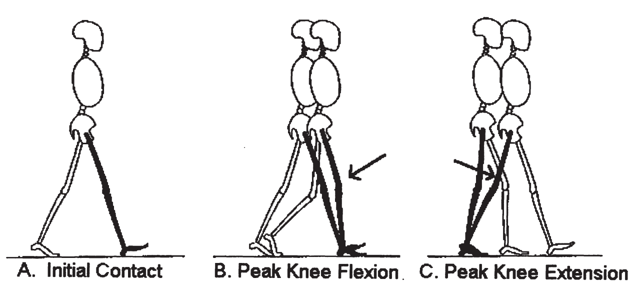

Fig. 2. Gait cycle variables: (A) Knee angle at initial contact; (B) Peak knee flexion during loading response; (C) Maximum knee extension during terminal stance. (Redrawn, with permission, from [20], Figures 2.1; 2.2; 2.4 and 2.5; pages 12–14).

Table 1
Variables for walk trials

Discrete event	Kinematic variable	Kinetic variable (moments are internal)
Initial contact (IC)	knee flexion angle	
	ankle flexion angle	
Peak knee flexion (*LR*)	knee flexion angle	knee extensor moment
		knee power absorption
	ankle flexion angle	ankle plantar flexor moment
		ankle power absorption
Peak knee extension (*M*St)	knee flexion angle	knee extensor moment
		knee power absorption
	ankle flexion angle	ankle plantar flexor moment
		ankle power absorption
		peak vertical ground reaction force during loading response

Table 2
Jog variables

Discrete event	Kinematic variables	Kinetic variable
Initial contact	knee angle	
	ankle angle	
Peak knee flexion	knee angle	
	ankle angle	
Peak knee extension	knee angle	
	ankle angle	
		peak vertical ground reaction force during stance

Since our data were collected unilaterally, the choice was made to collect data from the limb which stepped up on the step because it was thought that the ACL deficient knee of the supporting limb would be most challenged during step up and while supporting the body's weight during the step down from the step. It was also thought that if the subject had difficulty controlling the supporting limb's knee, the vertical ground reaction force might be higher as the subject stepped off the step and the 'step over' limb contacted the force plate. The variables for step trials included peak knee flexion of the supporting limb on the step, peak knee extension of the supporting limb on the step, knee flexion angle of the supporting limb as the 'step-over' limb contacted the force plate and peak vertical ground reaction force as the 'step-over' limb accepted the body's weight on the force plate.

2.5. Data analysis

Student's t-tests were used to compare the group differences in laxity and Quadriceps Index. Kinematic and kinetic variables were analyzed for side-to-side and group differences using an analysis of variance (ANOVA) with two repeated measures (side and condition). When ANOVA's revealed significant side-by-group interactions, within group, paired t-tests were used for post-hoc comparisons. Quadriceps femoris muscle strength (Quadriceps Index) was correlated with the measured gait variables using a Pearson Product Moment Correlation (SYSTAT, SYSTAT, Inc., Evanston, IL). The level of significance for statistical measures was $p < 0.05$.

3. Results

There was a trend toward less quadriceps muscle strength in the ACL deficient limbs of the non-coper group whose average quadriceps index was 88.3% (± 12.5%) compared to 98.9% (± 9.8%) in the coper group (t = 1.844, p = 0.088). The self rating scores, reported as a percentage of knee function prior to injury, were significantly different with the non-copers rating involved knee function at 53.6% (± 9.4%) whereas the copers rated involved knee function at 92% (± 8.4%) (t = 8.259, p = 0.000).

There was a significant correlation between the quadriceps index and the subjects' self-report of functional ability (R = 0.521, p = 0.05). There was no correlation, however, between the quadriceps index and the amount of knee flexion during weight acceptance on the involved side in the copers (R = 0.207, p = 0.209) or non-copers (R = 0.147, p = 0.730).

3.1. Walking

The copers and non-copers' walking speeds were comparable for all trials. The non-copers walked at an average normalized rate of 1.87 meters/second/leg length while collecting data from the involved and uninvolved sides. The copers and the copers walked an average rate of 2.14 meters/second/leg length while collecting data from the involved side and 2.13 meters/second/leg length while collecting data from the uninvolved side. These were not statistically different by side (F = 0.001, p = 0.97) or by group (F = 2.649, p = 0.126). Kinematic and kinetic differences were most apparent during early stance as the limb accepted the weight of the body. The involved knees of the non-copers displayed different kinematics than those of their own healthy knees and both knees of the copers; the latter three of which were indistinguishable from one another (Fig. 3). The non-copers landed at initial contact with significantly less flexion on the involved side (side × group F = 5.048, p = 0.041, t = 1.809, p = 0.113). The non-copers also showed a trend toward less knee flexion on the involved side during loading response (side × group F = 3.578, p = 0.079, t = 2.545, p = 0.038). Ankle kinematics were no different between groups or sides throughout early stance.

Both groups showed similarities in kinetic variables on the involved sides. Copers and non-copers landed with lower peak vertical ground reaction forces on the involved side (F = 4.630, p = 0.049), with no difference seen by group (F = 0.029, p = 0.867) (Fig. 4). The knee extensor moment was lower in both copers and non-copers on the involved side (F = 5.296, p = 0.044), with no difference by group (F = 0.010, p = 0.921) (Fig. 5). Knee extensor power absorption was also lower in the involved sides of the copers and non-copers (F = 8.476, p = 0.016) (Fig. 6) with no difference seen by group (F = 0.329, p = 0.579). The involved knee power absorption at peak knee flexion, in the copers was nearly half that absorbed by the involved limb of the non-copers. The net ankle moment at peak knee flexion was lower in both limbs of the copers, though this was not statistically different (Fig. 7).

At mid- and terminal-stance, the differences between copers and non-copers were not as striking. The ankle moments at peak knee extension were no different by side or by group. The knee moments at peak knee extension were net flexor moments, again, with no differences in magnitude by side or by group. There was, however, a trend toward greater ankle plantar flexion power absorption at the point of peak knee extension on the involved side for both groups (Fig. 8).

3.2. Jogging

During jogging trials the non-copers jogged at an average normalized speed of 3.587 meters/second/leg length while collecting data from the involved leg, and 3.66 meters/second/leg length while collecting data from the uninvolved side, and the copers jogged at a rate of 3.823 meters/second/leg length while collecting data from the involved leg, and 4.051 meters/second/leg length on the uninvolved side. The non-copers' speed appears slightly slower, although their speeds were not statistically different from the non-copers by side (F = 3.887, p = 0.069) or by group (F = 0.964, p = 0.343).

The differences in knee kinematics during loading response were even more pronounced during jogging (Fig. 9). The non-copers flexed the involved knee less at initial contact (side × group F = 4.452, p = 0.053, t = 2.450, p = 0.044) whereas the copers contacted the floor with essentially the same knee flexion angles on the uninvolved and involved sides (t = –0.536, p = 0.608). The non-copers had less peak knee flexion during the stance phase of jogging in their involved limbs (side × group F = 4.659, p = 0.049, t = 3.350, p = 0.012) as

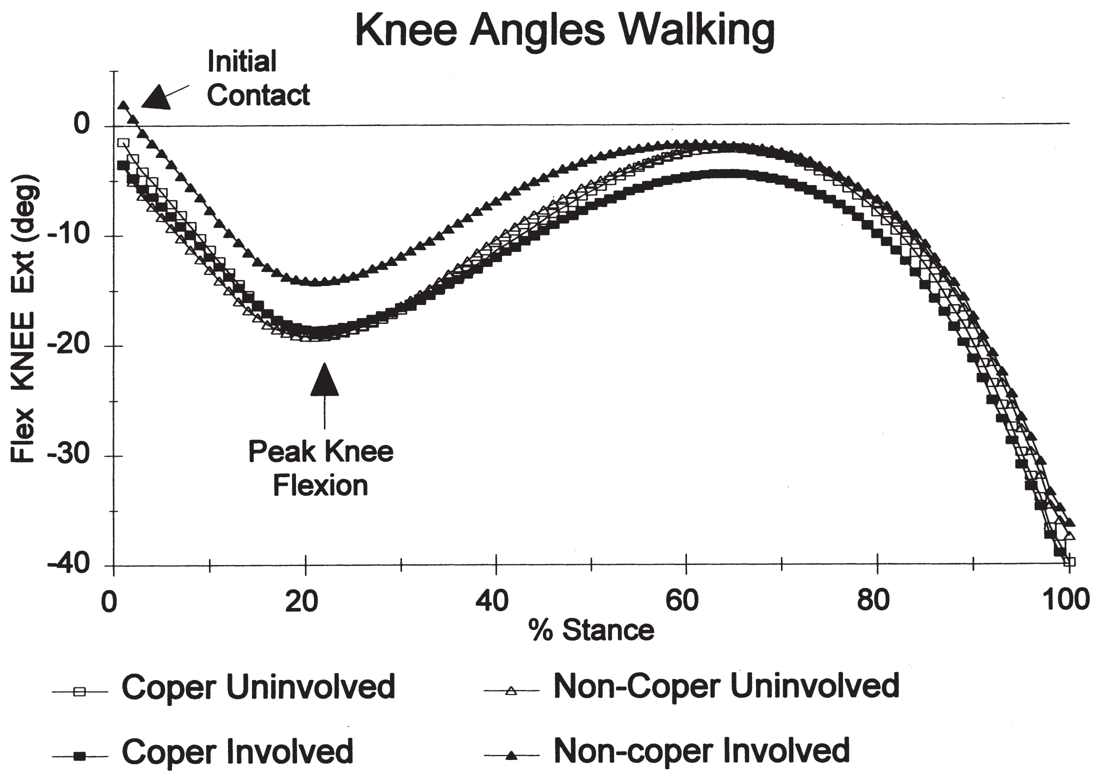

Fig. 3. Knee joint angles during walking (degrees). Flexion is negative.

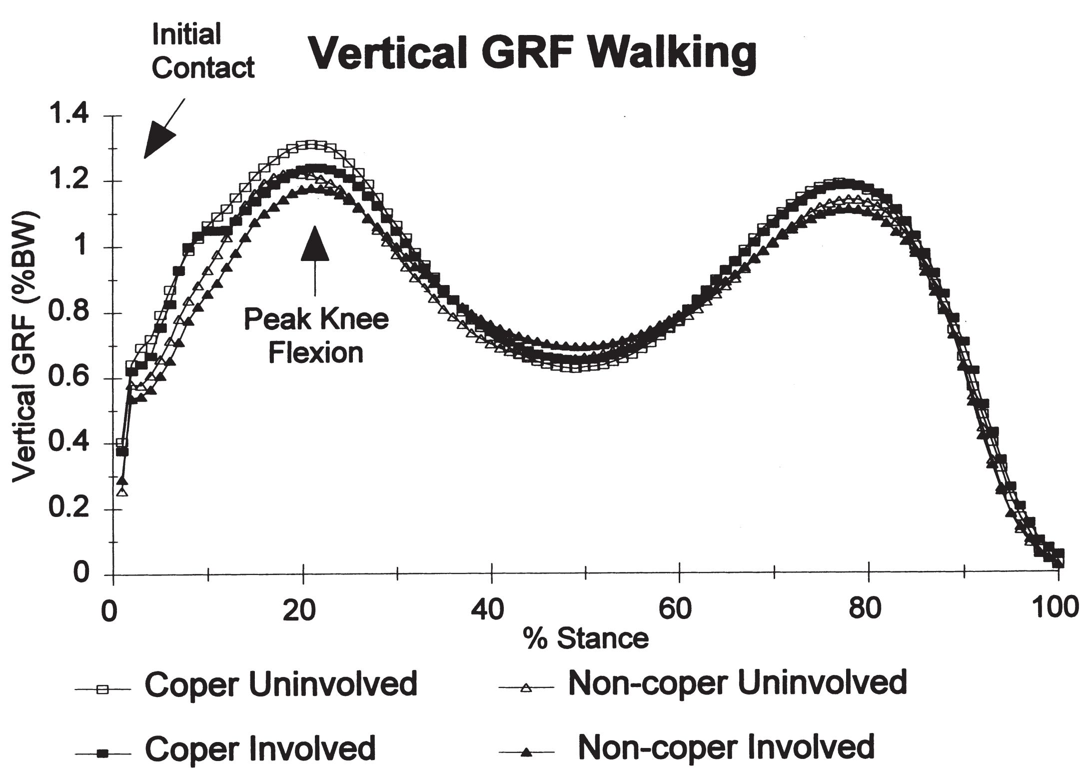

Fig. 4. Vertical ground reaction force during walking, normalized to body weight (% body weight).

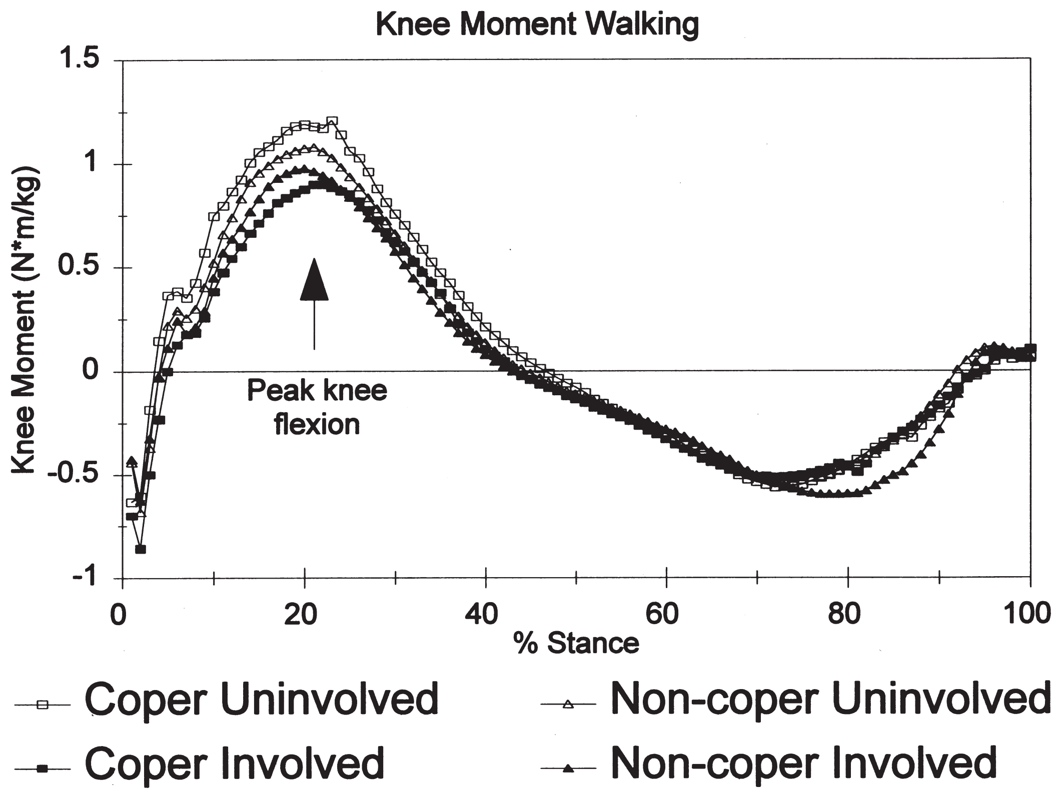

Fig. 5. Net internal knee moment during walking, normalized to body mass (N × m/kg). Positive values indicate the greater relative contribution of the knee extensors; negative values indicate the greater relative contribution of the knee flexors.

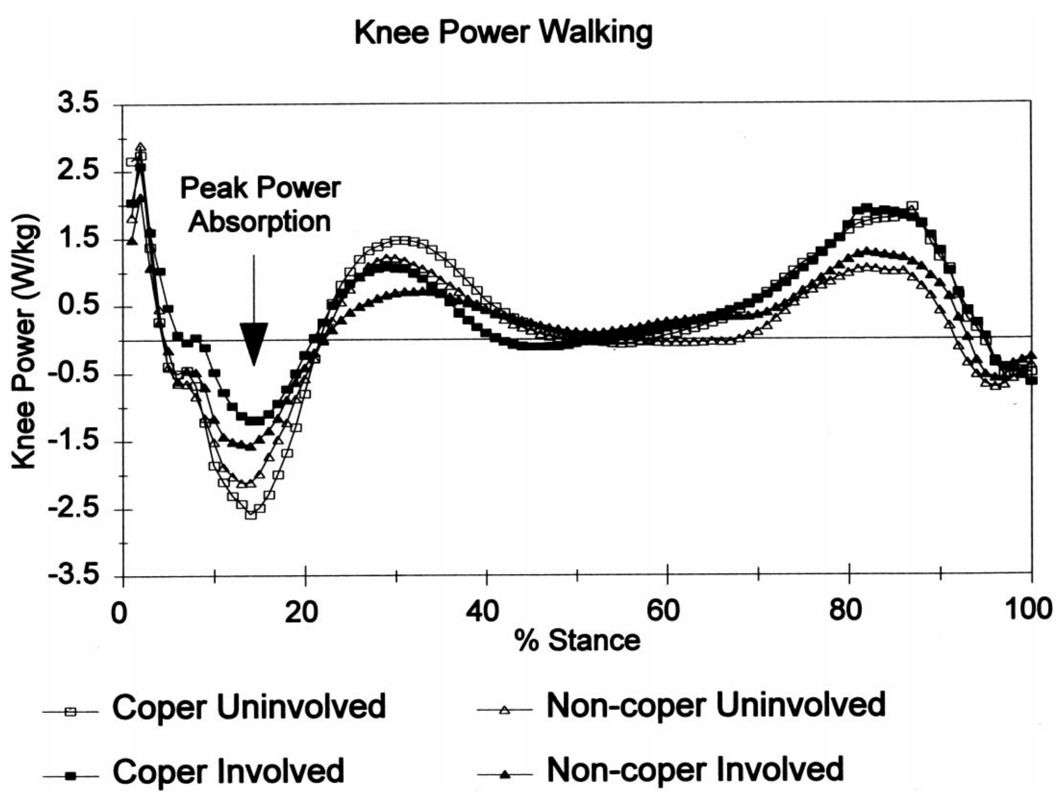

Fig. 6. Knee power during walking, normalized to body mass (Watts/kg). Positive values indicate power generation; negative values indicate power absorption.

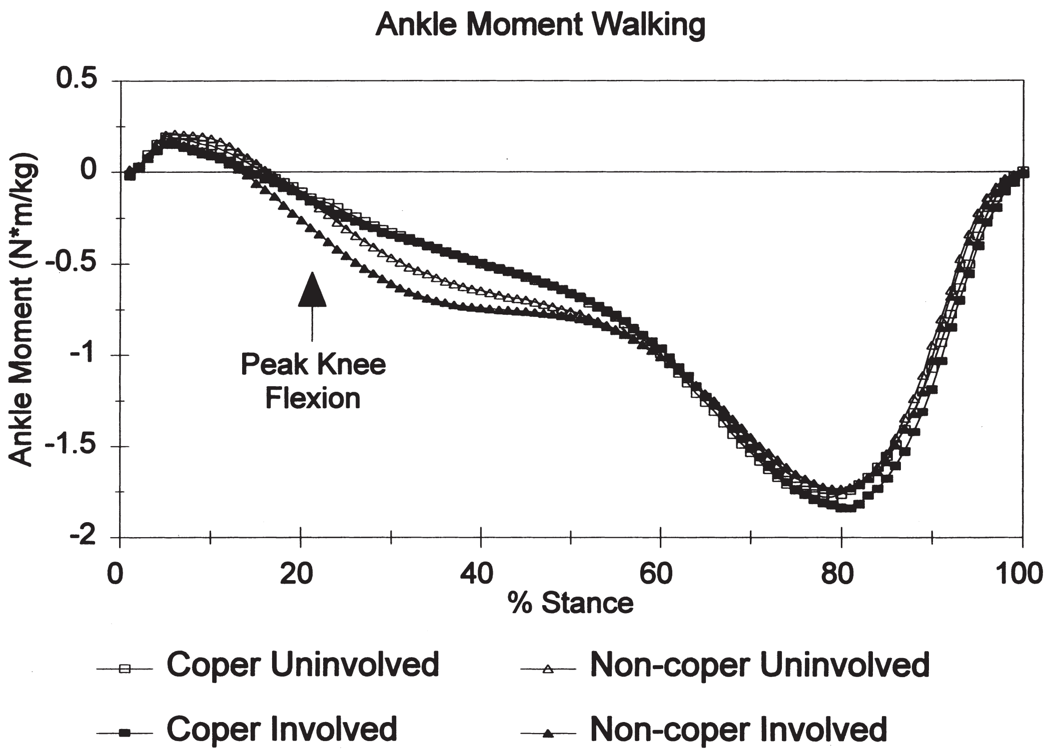

Fig. 7. Net internal ankle moment during walking, normalized to body mass (N × m/kg). Positive values indicate the greater relative contribution of the ankle dorsiflexors; negative values indicate the greater relative contribution of the ankle plantar flexors.

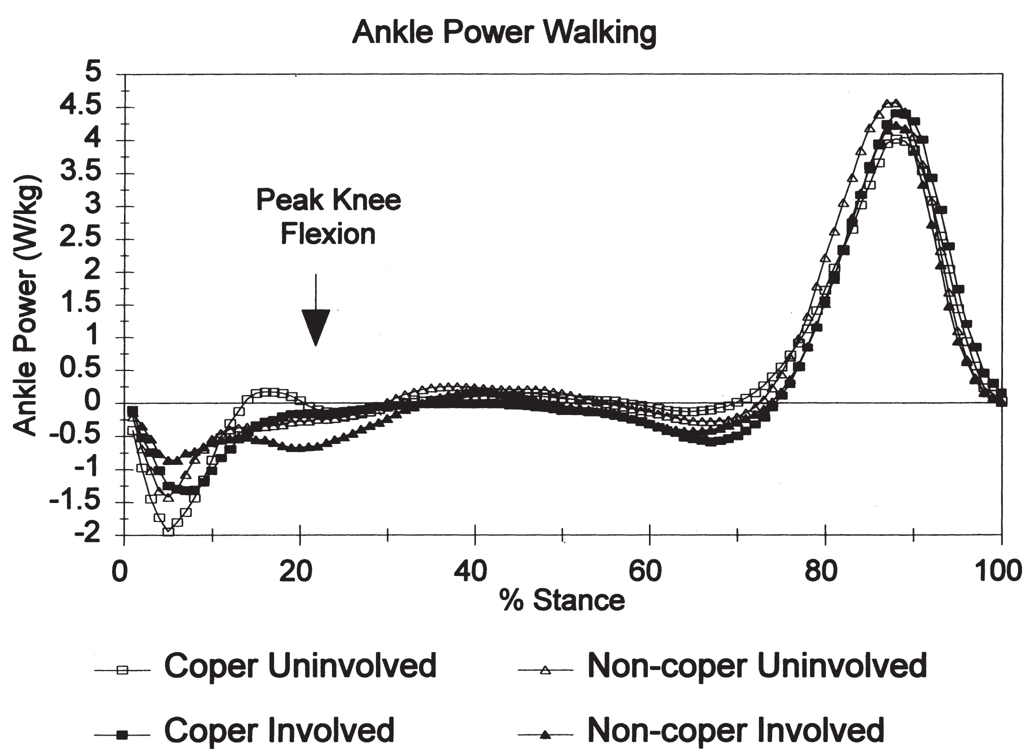

Fig. 8. Ankle power during walking, normalized to body mass (Watts/kg). Positive values indicate power generation, negative values indicate power absorption.

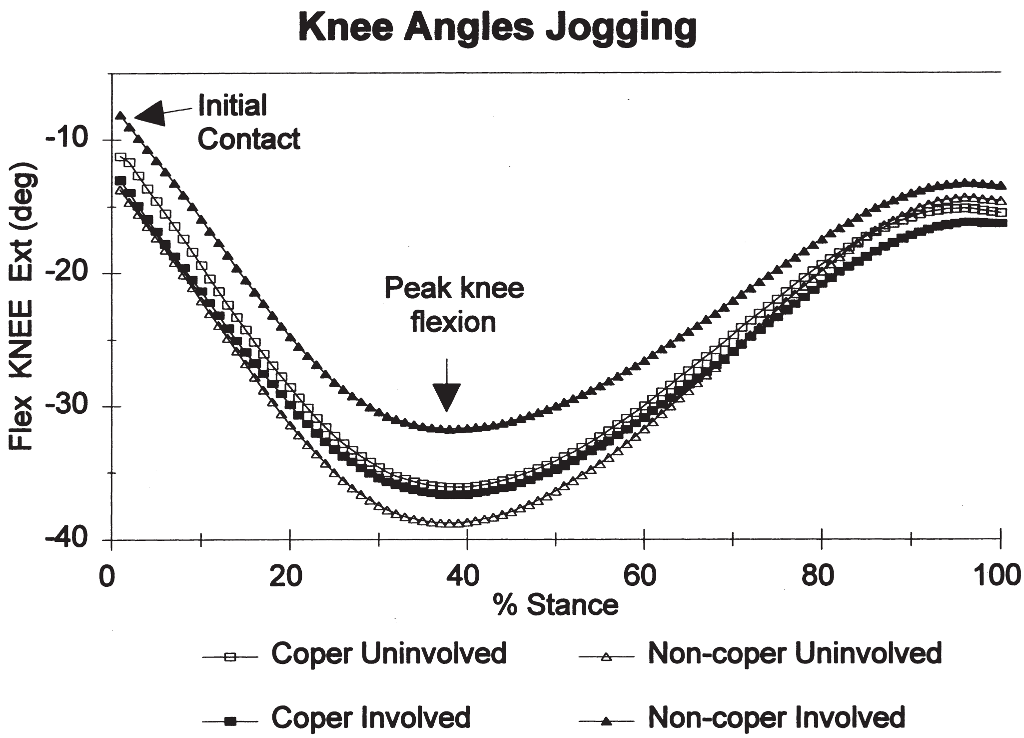

Fig. 9. Knee joint angles during jogging (degrees). Flexion is negative.

well, whereas the copers peak knee angles were no different by side ($t = 0.384$, $p = 0.712$). The ankle kinematics were no different between groups or limbs.

The vertical ground reaction during jogging was lower in the non-coper group overall (group $F = 9.077$, $p = 0.009$) and lower in the involved limbs of both groups (side $F = 7.696$, $p = 0.015$) (Fig. 10). There was a trend toward less vertical ground reaction force on the involved side of the non-copers which was observed in the post-hoc tests ($t = 3.174$, $p = 0.016$). No other differences were observed in the knee or ankle during jogging.

3.3. Step

Results of step trials can be seen in Table 3. During the step trials, the non-copers flexed their knees less to ascend the step (position B, Fig. 1) with the involved limb than with the uninvolved limbs and both limbs of the copers. Both groups demonstrated more knee extension on the step (side $F = 9.353$, $p = 0.009$) when the involved limb supported the body's weight on the step (position C, Fig. 1). The knee angle of the supporting limb at the point when the 'step over' limb contacted the force plate (position D, Fig. 1) was significantly less on the involved limbs of both groups ($F = 5.526$, $p = 0.034$). The peak vertical ground reaction force was lower in the non-coper group as a whole ($F = 6.862$, $p = 0.020$).

4. Discussion

Our hypothesis that the coper and non-coper samples would demonstrate different gait patterns that were unrelated to the amount of joint laxity was supported by the data. Neither group, however, walked 'normally'. The classification of anterior cruciate ligament deficient subjects into copers and non-copers allowed us to clarify the gait abnormalities in this population. The copers in this study were selected based on their being 'the best of the best' and served as a template for how successful compensation occurs in the face of this injury. The non-copers, however, were studied as they presented to our clinic and represent a cross section of non-copers who require surgical stabilization. The time since injury in the copers was much greater than in the non-copers. The laxity values were the same in both groups; both had the same anatomic probability of instability. The non-copers in our sample had symptoms of instability, even with activities of daily living immediately following their injury. The copers had managed to maintain knee stability during high level sports for many years. This provides further evidence of the highly successful stabilization strategy used by the copers who had put their knees in harm's way yet had still avoided further damage to their knees.

The quadriceps muscles of the non-coper group were slightly weaker, although not statistically so and there-

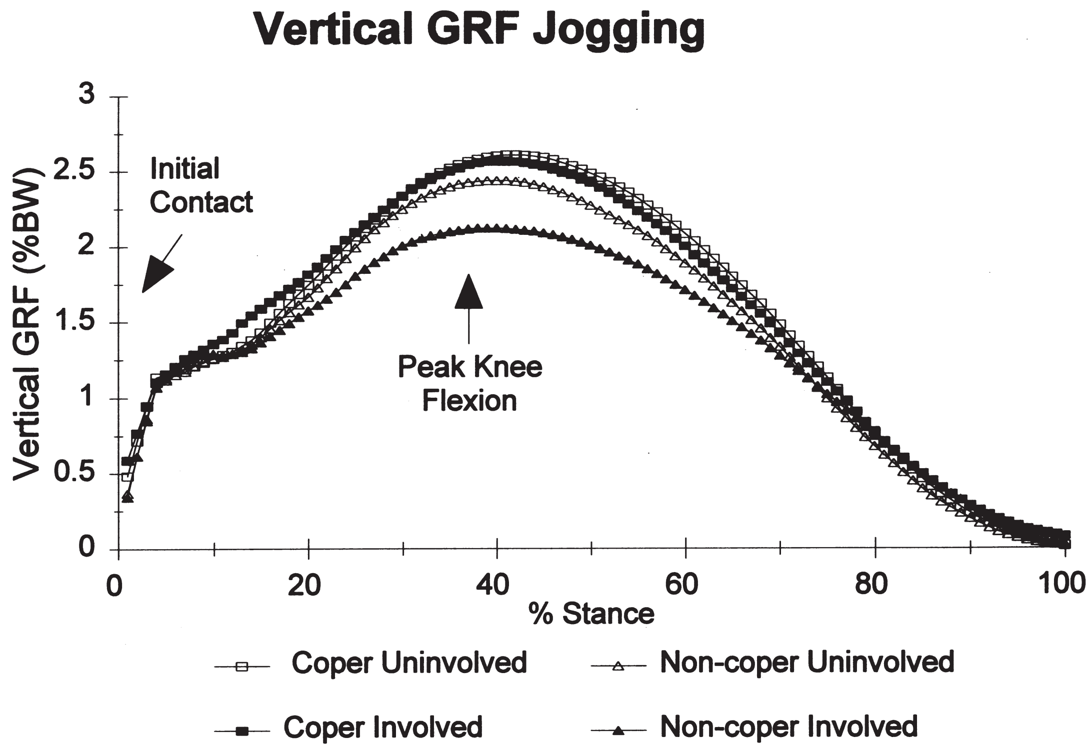

Fig. 10. Vertical ground reaction force during jogging, normalized to body weight (% body weight).

Table 3
Knee joint kinematics and ground reaction forces, step trials

	Coper Involved	Uninvolved	Non-coper Involved	Uninvolved
Flexion on step (degrees × Leg length)	− 90.283	− 86.903	− 73.571*	− 86.797
* (F = 11.352, p = = 0.005)	(8.636)	(9.737)	(12.511)	(4.737)
Extension on step (degrees)	− 25.024**	− 35.169	− 26.586**	− 32.205
** (F = 9.353, p = 0.009)	(12.874)	(9.985)	(11.996)	(8.954)
Stance limb knee flexion at contralateral initial contact (degrees × Leg length)	− 42.817***	− 55.963	− 41.643***	− 58.724
*** (F = 61.278, p = 0.000)	(11.343)	(10.507)	(± 9.071)	(12.812)
Vertical ground reaction force at contralateral heel strike (% body weight)	2.682	2.630	2.418****	20.069****
**** (F = 6.862, p = 0.020)	(0.349)	(0.402)	(0.400)	(0.423)

fore, this group was stronger than most non-copers [13]. The non-copers had obvious alterations in the knee flexion angles on their involved sides during walking, jogging and step trials, whereas the copers maintained normal knee motion. The mechanism(s) by which the copers adapt to anterior cruciate ligament deficiency cannot be completely elucidated from these data; however, subtle, but clear contrasts in the kinetics of walking suggests some evidence for a mechanically efficient, dynamic compensation mechanism adopted by the copers.

The most striking differences in the movement patterns between copers and non-copers appeared during weight acceptance. As the heel hits the floor, a strong eccentric quadriceps muscle contraction follows and is necessary to control the normal knee flexion, which could displace the tibia anteriorly [3]. The potential for knee instability at this point in the gait cycle is substantial when the anterior cruciate ligament is ruptured. The non-copers in this study stiffened their knees, landing in less flexion and accepting weight with less flexion in both walking and jogging, whereas the copers maintained equal knee motion on both sides.

Both copers and non-copers had lower peak vertical ground reaction force and had lower internal knee extensor moments on their involved sides. The lower knee moments observed in the involved limbs of these subjects can represent less activity by the quadriceps, the so called 'quadriceps avoidance gait' suggested by Berchuck and Andriacchi [1]. The moments are net moments, however, and the reduced knee extensor moment more likely reflects a greater relative contribution of the knee flexors, including hamstrings and gastrocnemius muscles. Although the net knee extensor moments are lower in the copers and non-copers, the mechanism behind the reduction appears to be different. In non-copers, the lower knee extensor moment can be almost wholly explained by some combination of decreased knee flexion during weight acceptance and lower vertical ground reaction forces. At the point of peak knee flexion on the involved side, the non-copers demonstrated the greatest power absorption by the plantar flexors. Increased activation of the plantar flexors can act to limit the anterior tibial translation as the body advances over the stance limb [20] and could result in the limited knee flexion we found in the non-coper group.

The copers are able to subtly alter the knee moments and powers in order to control the knee joint during weight acceptance while maintaining joint motions similar to those reported for adult healthy subjects [18]. The copers showed a much more pronounced attenuation of knee power absorption than in the non-copers; the involved limb's value was less than half that of the uninvolved limb (Fig. 6). This reduction in knee power absorption may represent the transfer of power away from the knee. At peak knee flexion the copers show the least ankle power absorption by the plantar flexors on their involved sides. In contrast to the non-copers, the copers do not appear to be relying on their ankle plantar flexors for controlling tibial advancement. This low ankle power absorption might also represent a more finely tuned stabilization strategy whereby the pre-tibial muscles act to draw the tibia forward during loading response [20] thus balancing any increased plantar flexor activity. Without electromyographic data these speculations cannot be confirmed; however our data provide evidence that different stabilization strategies are at play.

We theorize that a pattern of muscular co-contraction is responsible for the kinematic and kinetic data seen in the non-copers. Altered muscle activation in ACL deficient subjects has been described by other investigators [2,3,14,22]. Limbird et al. [14] found that anterior cruciate ligament deficient subjects had less vastus lateralis, rectus femoris and gastrocnemius muscle activity, along with greater biceps femoris activity during knee loading. Sinkjaer and Arendt-Neilson [22] found that subjects who had reported 'good/excellent' knee stability, activated the medial head of the gastrocnemius and lateral hamstrings earlier than did those with poor stability. Cicotti et al. [3] found increased tibialis anterior activity during both terminal swing and terminal stance.

If knee joint stiffening seen in the non-copers resulted from a strategy involving generalized muscle co-contraction, this would lead to increased joint compression forces [26]. Generalized joint stiffening is a more crude compensation tactic which would be incapable of stabilizing the knee under all conditions, particularly in response to sudden, unexpected perturbations. The greater incidences of giving way experienced by the non-copers would lead to more shear forces in the knee. These two forces, compression and shear, contribute significantly to the biochemical and metabolic changes that characterize degeneration of articular cartilage. Recent work by Setton et al. [21] demonstrated that even small shear forces lead to changes in the viscoelastic behavior of articular cartilage after anterior cruciate ligament transection in a canine model. Episodes of giving way would be likely to cause abnormal shear forces on the articular cartilage of the knee. The joint stiffening stabilization strategy seen in the non-copers in this study may be one which is not only unsuccessful in stabilizing the knee but may lay the unfavorable ground work for further articular damage.

The normal joint motion that was seen in the copers occurs despite side-to-side knee laxity measurements that were no different than those of the non-copers [18]. The copers' ability to maintain normal joint motion should minimize the compression and shear forces. Daniel et al. [5] suggested that the development of arthritic changes after anterior cruciate ligament injury is *not* inevitable and that there are sub-populations that are stable and free of arthritic changes despite their anterior cruciate deficiency. The ability of the copers to mitigate contact forces without compromising joint motion may bode well for favorable knee function in the long term.

Quadriceps femoris muscle strength does not appear to be the primary influence on the copers' ability to stabilize their knees in the absence of an anterior cruciate ligament, since the non-coper group in this study did not have significant quadriceps weakness. Quadriceps muscle weakness did not correlate well with the amount of knee flexion or knee extensor moments during weight acceptance. Other studies have correlated quadriceps femoris muscle strength with the amount of knee flexion [23] and knee extensor moment [11] in subjects following anterior cruciate ligament reconstruction. In a previous study performed in our laboratory, the gait pattern of individuals after anterior cruciate ligament reconstruction did correlate with isometric quadriceps muscle strength of the involved limb [25]. As the quadriceps muscle strength in the involved limb became more equal to that in the uninvolved limb larger knee flexion excursions were observed. Hurwitz et al. [11] found that ACL reconstructed subjects had lower knee extensor moments which correlated positively with isokinetic quadriceps femoris muscle strength. Those authors reported that the ACL reconstructed subjects had quadriceps strength comparable to the uninjured subjects and suggested that the subjects with ACL reconstructed knees require greater strength to maintain normal function. Their method of evaluating the muscle strength is unclear. The failure of the investigators to find a difference in strength in the ACL reconstructed subjects may have resulted from the failure to account for side-to-side strength deficits. Both the copers and non-copers in the current study had lower knee extensor moments in their involved limbs despite quadriceps indices of 98% and 88% respectively. Only one of the eight coper subjects had a quadriceps index below 90%, and three had quadriceps indices over 100%, yet they also showed the same reduced knee extensor moment as seen in the non-copers. If quadriceps femoris muscle strength was the primary factor influencing the knee moment, the copers should have had involved knee extensor moments comparable to those of the uninvolved knees. It appears that the reduction in knee extensor moment is part of an intricate compensation mechanism used by ACL deficient individuals rather than a result of inadequate quadriceps muscle strength.

There was no statistically significant difference in walking and jogging speed between the groups, although the copers did walk and jog slightly faster than the non-copers. Faster walking and jogging velocities should result in *higher* forces when the foot makes contact with the ground [17]. The kinematic and ground reaction force data during jogging trials suggest that the differences in control strategies only become more pronounced as the demand on the knee increases. Knee flexion is reduced overall in the involved limb of the non-copers, and the vertical ground reaction force is substantially lower in the involved limb of the non-copers. Despite the more stressful activity, the copers show essentially identical side-to-side joint motions (Fig. 9) and vertical ground reaction forces (Fig. 10). The copers, again, appear to use a strategy in which they are able to dynamically stabilize the knee during weight acceptance while preserving normal knee motion. The stiffening seen in the non-copers during walking is even more pronounced in jogging.

The pattern of decreased knee flexion which manifested itself during walking and jogging in the non-coper population also occurred in ascent and descent from the step and we found some similar kinematic differences in both groups. The non-copers used less knee flexion when ascending the step with the involved limb. Both increased hip abduction and/or ankle plantar flexion on the contralateral side could compensate for the decrease in knee flexion to ascend the step. These trials appeared to be challenging for both the copers and non-copers. Both groups showed greater knee extension when their involved knees supported the body's weight on the step. Extending the knee to a greater extent may serve to increase the mechanical advantage of the quadriceps femoris muscle in an attempt to put the knee in a more stable position. During descent from the step, when the involved limb supported the body's weight and controlled the descent, both copers and non-copers flexed their knee less than when their uninvolved knee was in control. If the ankle of the descending limb was positioned in more plantar flexion this would result in a functionally longer limb, which could reduce knee flexion at initial contact. Simultaneous kinematic data from the other limb is, unfortunately, not available. The non-copers as a whole, had significantly lower vertical ground reaction forces than the copers, particularly when the involved limb was controlling the descent from the step. It is possible that these individuals descended more slowly, a feature that we were unable to capture during these trials.

This study clearly indicates that, in a small number of individuals, anterior cruciate ligament deficiency does not necessarily lead to functional deficits which require surgical stabilization. The copers are able to induce an appropriate compensatory pattern to achieve a dynamic functional stability. Further investigation of joint kinematics and kinetics, with simultaneous electromyography, is necessary to more fully elucidate these differences.

5. Conclusions

Copers are a distinct subset of the anterior cruciate ligament deficient population who are able to fully compensate for the absence of the ligament. Despite ligamentous laxity that was greater and chronicity of injury that was longer than that of the non-copers, the copers in this study had quadriceps femoris muscle strengths, and knee joint motion in walking and jogging indistinguishable from their uninvolved knees. Non-copers compensate for the absence of the anterior cruciate ligament through a crude strategy of decreasing the force with which the foot hits the ground and stiffening the knee, forsaking mobility for stability. Copers use a more precisely coordinated stabilization strategy which maintains knee stability while maintaining normal joint motion.

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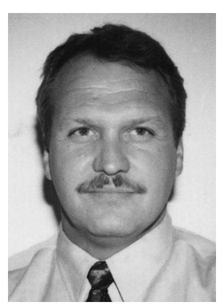
Katherine Rudolph received a B.S. in Geology from Syracruse University, and an M.S. in Physical Therapy from Boston University in 1989. She joined the staff of the Physical Therapy Department of the Massachusetts General Hospital in 1989 and in 1991 became the Clinical Research Physical Therapist in the Orthopedic Biomechanics Laboratory of the Shriner's Hospital, San Francisco Unit. Ms. Rudolph is currently a doctoral candidate in the Interdisciplinary Program in Biomechanics and Movement Science at the University of Delaware and is a member of the American Association for Physical Therapy and the North American Society for Gait and Clinical Motion Analysis.

Martha Eastlack received a B.S. in Biology from Oglethorpe University, an M.S. in Physical Therapy from Boston University, and a Ph.D. in Physiology and Anatomy from the University of Delaware. She has had clinical experience at the National Rehabilitation Hospital in Washington, D.C. and in orthopedics and sports medicine in Newark and Wilmington, DE. The focus of her research is in the areas of muscle morphology and muscle performance. Dr. Eastlack is a member of the Sections for Research, Orthopedics and Education of the APTA and is a member of the American College of Sports Medicine.



Michael Axe, a partner in First State Orthopedics, received his M.D. from Jefferson Medical College in Philadelphia. He completed his orthopedic residency at the University of Pittsburgh and a fellowship in sports orthopedic medicine at the Hughston Sports Medicine Hospital in Columbus, Georgia. He is a clinical associate professor and course director in the physical therapy department at the University of Delaware. He is a member of the American College of Sports Medicine and a fellow of the Anerican Academy of Orthopedic Surgeons. He is the regional physician for the United States Weightlifting Federation and has made numerous presentations and published many articles related to orthopedics and sports medicine.

Dr. Snyder-Mackler is an Associate Professor in the Department of Physical Therapy at the University of Delaware. She also holds a faculty appointment at Allegheny University and is the Director of Sports Medicine Research for the Orthopedic Sports Medicine Fellowship at Thomas Jefferson University in Philadelphia. She maintains an active Sports Physical Therapy practice at the University of Delaware and serves as a rehabiltation consultant to several professional teams. She is a Board Certified Sports Physical Therapist and concentrates her clinical practice and research in the areas of knee, back and shoulder rehabilitation, and electrical stimulation of muscle. She has authored a textbook on electrotherapy and many research publications in the areas of knee, shoulder and back rehabiltation and neuromuscular electrical stimulation. She served as Head Trainer for the beach volleyball venue at the '96 Olympic Games in Atlanta.