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# Dynamic stability in the anterior cruciate ligament deficient knee

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**Abstract** Some individuals can stabilize their knees following anterior cruciate ligament rupture even during activities involving cutting and pivoting (copers), others have instability with daily activities (non-copers). Movement and muscle activation patterns of 11 copers, ten non-copers and ten uninjured subjects were studied during walking and jogging. Results indicate that distinct gait adaptations appeared primarily in the non-copers. Copers used joint ranges of motion, moments and muscle activation patterns similar to uninjured subjects. Non-copers reduced their knee motion, and external knee flexion moments that correlated well with quadriceps strength. Non-copers also achieved peak hamstring activity later in the weight acceptance phase and used a strategy involving more generalized co-contraction. Both copers and non-copers had high levels of quadriceps femoris muscle activity. The reduced knee moment in the involved limbs of the non-copers did not represent „quadriceps avoidance“ but rather represented a strategy of general co-contraction with a greater relative contribution from the hamstring muscles.

**Keywords** Anterior cruciate ligament · Knee · Dynamic stability · Quadriceps avoidance

## Introduction

Anterior cruciate ligament (ACL) injury usually leads to knee joint instability that often prevents people from participating in sports and daily activities. Historically, the degree of passive knee laxity was thought to be a good indicator of the need for surgical reconstruction; however, research has shown that passive anterior knee laxity is unrelated to knee stability during *dynamic* activities [5, 11, 18, 29]. We have identified a cohort of ACL deficient individuals, called "copers", who have no symptoms of knee instability even with sports involving cutting and pivoting [9]. Copers are rare and they provide the unique opportunity to study individuals who have clearly developed a successful knee stabilization strategy in the absence of an ACL.

The mechanism by which ACL deficient individuals attempt to stabilize their knees is unclear despite the myriad of research studies performed on the ACL deficient population. Evidence in the literature suggests that quadriceps femoris muscle strength is related to functional outcome [12, 27], but quadriceps strength alone does not fully characterize dynamic stability of the knee after ACL rupture [11]. Some differences in joint kinematics and kinetics [3, 6, 16] and muscle activity patterns [4, 16, 17, 19, 24, 25] in ACL deficient subjects have been found, yet the conclusions from studies are often contradictory. The lack of consistent findings is due, in part, to a failure of investigators to account for the presence of copers in a general sample of patients. During low level activities, such as walking, ACL deficient individuals who are highly successful at stabilizing their knees dynamically (copers) display no alteration in their gait pattern [22]. If copers are included in the mix of all subjects, then genuine differences in movement patterns will be obscured.

Methodological differences also account for inconsistency in research findings. Investigators who report kinematics and kinetics have identified some common alterations in movement patterns [3, 12, 32] but often infer muscle activation from kinetic data without electromyography to confirm those inferences. Some investigators who use kinetics to infer muscle activity have suggested that ACL deficient people reduce their quadriceps activation to avoid knee instability [3, 12] yet electromyographic (EMG) studies have shown the contrary [4, 16, 24]. Similarly, investigators who report muscle activation often do not include kinematic and kinetic data [4, 17, 19, 24] so the effect of altered muscle activation on movement patterns is unknown.

The purpose of this study was to investigate knee stabilization strategies in two groups of ACL deficient subjects with distinctly different functional abilities (copers and non-copers). We postulated that non-copers would reduce their knee motion and external knee flexion moment through a strategy of generalized co-contraction of knee flexors and extensors, and that copers would move like uninjured subjects. We also postulated that no patients would reduce quadriceps muscle activity, and that the ability to stabilize the knee was unrelated to passive joint laxity and partially related to quadriceps femoris muscle strength.

## Materials and methods

### Subjects

Thirty-one active subjects, ten uninjured and 21 ACL deficient subjects gave informed consent, approved by the Human Subjects Review Board of our institution, to participate in this study. Eleven of the ACL deficient subjects (two female, nine male; ages 22–43 years; mean 30.7) were classified as „copers“ using the following operational definition: high level athletes who were ACL deficient for at least 1 year; no symptoms of knee instability during regular participation in level I (involving jumping, pivoting and hard cutting) and II (involving lateral motions) sports [7]; and had experienced no more than one episode of giving way since injury. The copers were identified from the community by local physicians, coaches, physical therapists and other athletes. All coper individuals who were identified were recruited for and participated in this study. Ten individuals, who had been participating in level I or II sports prior to injury and were within 8 months of ACL rupture, were recruited as non-copers with consent from their physician. In order to be classified as a non-coper, subjects had to demonstrate substantial knee instability during activities of daily living, since the non-copers had not yet returned to sports. To demonstrate knee instability objectively, potential non-copers underwent a pre-screening rehabilitation program (including edema control, range of motion, strengthening, and agility activities for the injured knee) and then underwent a screening evaluation of dynamic knee function based on the work of Fitzgerald et al. [10]. The screening tests (with cut-off criteria) were: four hop tests [20] (<80% timed hop); Knee Outcome Survey – Activities of Daily Living Scale [13] (<80%); global rating scale (<60% of prior knee function); episodes of giving way (>1 since injury). Subjects were classified as non-copers if they met any one of the criteria. Ten non-copers (four women, six men; ages 16–43 years, mean 28.1) were recruited for the study. Ten uninjured individuals, who were matched by age and activity level to the coper subjects (ages 23–41 years, mean 32.2), also participated in the study. All ACL deficient subjects had an uninvolved knee that was uninjured and had full range of motion in both knees. ACL deficient subjects who had undergone previous meniscectomies in the involved knee were included.

### Quadriceps strength and knee joint laxity testing

Quadriceps femoris muscle strength was measured bilaterally using a maximum voluntary isometric contraction (MVIC) with burst superimposition as described previously [26, 27, 28]. The MVIC of each leg was recorded and a quadriceps index was calculated (involved MVIC/uninvolved MVIC×100). In tests on ten healthy subjects repeated testing of the MVIC revealed interclass correlation coefficients (ICC 2, 1) of 0.98 [27]. Passive anterior knee joint laxity was measured using a KT-2000 arthrometer (KT-2000, Medmetrics, San Diego, Calif., USA). The anterior laxity was reported as the difference in laxity from side to side (involved-uninvolved). Daniel et al. [8] reported that a side-to-side difference over 3 mm confirmed ACL rupture in 99% of cases. Reliability of the KT-2000 has been shown to be high with inter-tester reliability coefficients for the injured limb of 0.92 injured limb and 0.75 for the uninjured limb. The day-to-day reliability coefficient for the injured limb is 0.94 and 0.75 for the uninjured limb [21].

### Electromyographic testing

Electromyographic data were recorded from the lateral hamstrings, vastus lateralis, soleus, and medial head of the gastrocnemius muscles of both limbs using surface electrodes and an eight channel FM radio telemetry system (B & L Engineering, Santa Fe Springs, Calif., USA) at 960 Hz. The timing and magnitude of the EMG from the gastrocnemius and soleus muscles were inspected visually for cross-talk and were found to have none (see Appendix 1). A linear envelope (LE) of each muscle was created using full wave rectification and low pass filter (2nd order, phase corrected, Butterworth filter) with cut-off frequency of 20 Hz. Electromyographic data were collected from each muscle group during two, 2-s maximum voluntary isometric contractions and a 1-s resting trial. Data from the dynamic trials were normalized to a maximum EMG, which was defined as the highest level of EMG over a 30-ms interval during any of the isometric, walking or jogging trials. Using this normalization technique all normalized linear envelope data during walking and jogging trials were at or below 100% of maximum activation.

Muscle timing variables (muscle onset and termination of activity) were determined based on a threshold of 2.5 times the average resting EMG level. The timing of peak EMG activity was determined with respect to initial contact. Reliability of the timing variables was assessed in our laboratory in a previous study using the coefficient of variation (CV). The variation of the time of peak muscle activity ranged from 0.44 to 0.59. The muscle onset and termination variables variable had coefficients of variation ranging from 0.12 to 0.33. Muscle contraction magnitude was evaluated by integrating the LE-EMG curves over a „weight acceptance interval“ defined as the range from 100 ms prior to initial contact (to account for electromechanical delay [30]) to the point of peak knee flexion (PKF). This interval was chosen because it is the interval during which strong quadriceps femoris contractions occur, thereby creating the greatest potential for knee instability. Muscle co-contraction (operationally defined as the simultaneous activation of two muscles) was assessed between the quadriceps and hamstrings (VL-LH) and quadriceps and medial gastrocnemius (VL-MG). Co-contraction was determined using the following equation developed in our laboratory: EMGS/EMGL×(EMGS+EMGL). EMGS was the level of activity in the less active muscle and EMGL was the level of activity in the more active muscle (to avoid division by zero errors). This ratio was multiplied by the sum of the activity found in the two muscles. This method provided an estimate of the relative activation of the pair of muscles as well as the magnitude of the co-contraction. The resulting curve was integrated over the weight acceptance interval and used in the analysis.

### Three-dimensional motion analysis

Five trials each of walking and jogging were performed bilaterally with 1- to 3-min rest intervals between each trial. Kinematic data were collected at a sampling rate of 120 Hz (Vicon, Oxford Metrics, London, UK) and force data were recorded from a force platform at a rate of 960 Hz (Bertec Corporation, Worthington, Ohio, USA). Subjects were instructed to walk and jog at their self-selected speed and practice trials were performed until the subject could contact the force platform with only one foot without targeting and velocity was consistently within ±5%. The six cameras were calibrated to a calibration volume of 1.67 $m^3$ and calibration errors were held below 1.5 mm. Joint motions and moments were calculated from markers (located on the pelvis, and bilaterally on the thigh, shank and foot) and force plate data using Move3d software (MOVE3D, NIH Biomotion Laboratory, Bethesda, Md., USA). Inter-subject variation in body size was minimized by normalizing ground reaction force to body weight and joint moments to body mass and velocities were normalized to the subject’s leg length. Previous work in our laboratory has revealed good reliability for kinematic variables with intraclass correlation coefficients (ICC) ranging from 0.6343 to 0.9969. The ICCs for the kinematic variables used in the present study were 0.9969 for hip flexion, 0.9721 for knee flexion, and 0.9932 for ankle flexion. Hip, knee and ankle joint moments were used to calculate the “support moment” which is the sum of the internal extensor moments at the hip, knee and ankle [31]. Winter postulated that the distribution of support moment about the hip, knee and ankle indicates the ability of the nervous system to adapt to different external demands on the body [31]. The relative contribution of each joint to the support moment (% support moment) was determined by dividing each internal extensor moment by the total support moment. Data from the five trials were averaged for each subject and used to identify variables at initial contact and peak knee flexion during stance.

Repeated measures analysis of variance (ANOVA), Pearson’s product moment, and multiple regression analyses were used to determine group and side difference and to determine relationships among variables (Systat, Evanston, Ill., USA). A probability level of $P<0.05$ was used to show statistical significance.

## Results

There were no differences in age ($F$=0.774, $P$=0.471) among the groups or in passive knee joint laxity between the copers and non-copers. The non-copers were the only group with quadriceps that were significantly weaker on the involved side (Table 1).

**Table 1** Joint laxity and quadriceps strength

| | Non-coper | Coper |
|---|---|---|
| Joint laxity ($t$=0.894, $P$=0.382) | 6.0 mm (±3.0 mm) | 5.1 mm (±1.5 mm) |
| Quadriceps strength ($t$=4.033, $P$=0.001)* | 75.3% (±11%) | 97.1% (±12.7%) |

*Significantly different

### Walking

Kinematic and kinetic data during the walking trials can be seen in Table 2. No differences in walking velocity, or stride lengths were found by group or by side during walking. During weight acceptance both copers and non-copers landed with less vertical force on both sides compared to control subjects. Non-copers flexed the knees less on the involved sides compared to their uninjured knees and both knees of the copers and control group. The hip and ankle angles were no different by side or by group. The knee moment at peak knee flexion was lower in the non-copers on the involved side. No differences were observed in the total support moment at peak knee flexion among the three groups; however, the contributions of the hip and knee extensors to the total support moment did differ among the groups. There was a significantly lower contribution from the knee with a concomitant higher contribution from the hip in the involved knee of the non-copers only (Fig. 1).

### Muscle activity

The activity patterns of the muscles crossing the knee joint were altered in the involved limbs of the non-copers (Fig. 2). The medial gastrocnemius had a significantly earlier onset ($F$=6.856, $P$=0.004), and longer total duration ($F$=4.614, $P$=0.019) in the non-copers. The earlier onset of the medial gastrocnemius resulted in the longer duration from onset-to-peak activity ($F$=4.840, $P$=0.016) seen in the involved limb of the non-copers, whose peak activity and termination occurred at essentially the same point in the stance phase as the other subjects. The only difference observed in the vastus lateralis was in the shorter onset-to-peak activity ($F$=9.374, $P$=0.014) in the involved limbs of the copers. The onset of the lateral hamstring was no different in the three groups, but the electromyographic activity showed delayed termination ($F$=3.655, $P$=0.041), longer total duration ($F$=6.622, $P$=0.005) and a trend toward a delayed peak activity ($F$=3.766, $P$=0.065) during

**Table 2** Walking data. *LL* leg length, *GRF* ground reaction force, *BW* body weight

| | Coper | | Non-coper | | Control | |
|---|---|---|---|---|---|---|
| | Involved | Uninvolved | Involved | Uninvolved | Left | Right |
| Velocity m/s per LL ($F$=0.897, $P$=0.420) | 2.099 (±0.09) | 2.053 (±0.09) | 2.056 (±0.09) | 2.077 (±0.09) | 2.234 (±0.09) | 2.199 (±0.08) |
| Stride length m/LL ($F$=1.982, $P$=0.160) | 2.09 (±0.05) | 2.12 (±0.06) | 2.13 (±0.05) | 2.19 (±0.06) | 2.28 (±0.05) | 2.23 (±0.06) |
| Vertical GRF at loading response ($F$=8.499, $P$=0.017)* | 1.25%BW (±0.030) | 1.23%BW (±0.022) | 1.22%BW (±0.031) | 1.26%BW (±0.033) | 1.31%BW* (±0.031) | 1.29%BW* (±0.033) |
| Peak knee flexion angle (negative=flexion) ($F$=8.499, $P$=0.017)** | −22.8° (±1.9) | −24.5° (±1.8) | −21.9°** (±1.9) | −25.9° (±1.9) | −26.5° (±1.9) | −26.5° (±1.9) |
| Knee moment at PKF (N*m/kg) ($F$=6.212, $P$=0.034)** | 0.368 (±0.07) | 0.437 (±0.68) | 0.314** (±0.071) | 0.542 (±0.071) | 0.558 (±0.074) | 0.601 (±0.071) |
| Soleus integral over weight acceptance ($t$=2.894, $P$=0.020)** | 8.655 (±1.292) | 7.489 (±0.679) | 9.811** (±1.362) | 6.626 (±0.716) | 8.302 (±1.292) | 7.612 (±0.679) |

*Control group different from copers and non-copers ($P$<0.05)
**Non-copers' involved side different from all others ($P$<0.05)

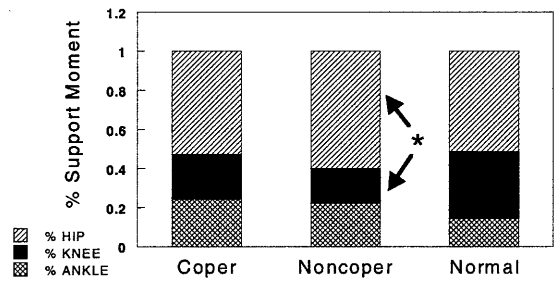


**Fig. 1** Distribution of support moments on the involved side during weight acceptance, walking. Non-copers* had lower knee moments ($F$=5.402, $P$=0.045) and higher hip moments ($F$=3.979, $P$=0.056) than copers or uninjured subjects

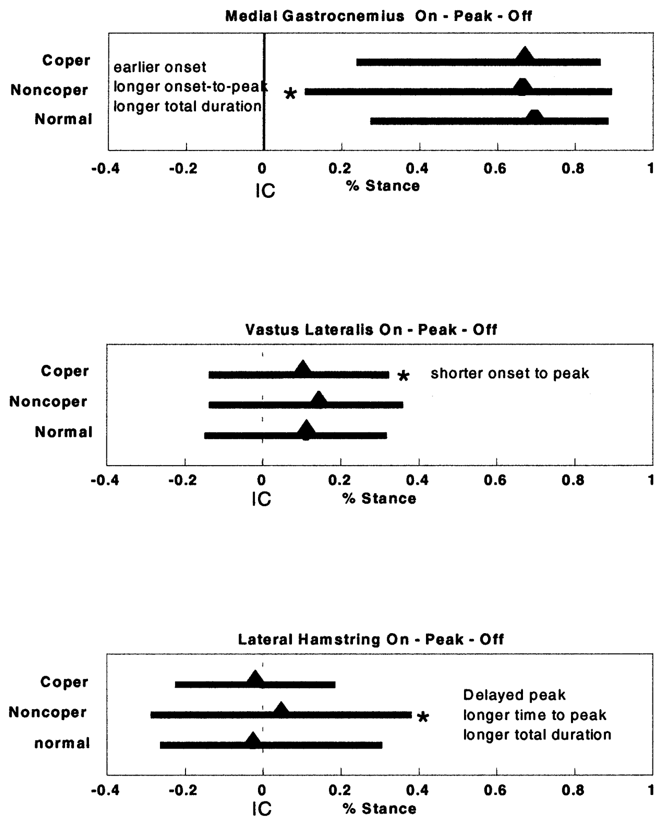


**Fig. 2** Muscle timing of medial gastrocnemius, vastus lateralis and lateral hamstrings during walking. *IC* indicates initial contact

stance phase in the involved limb of the non-copers. The time from onset-to-peak lateral hamstring activity was also significantly longer in the non-copers ($F$=8.668, $P$=0.002). The peak lateral hamstring activity occurred after initial contact in the involved side of the non-copers while it occurred before initial contact in the uninvolved limb of the non-copers and both limbs of the copers and control subjects.

The non-copers had the most pronounced differences in the groups with respect to the magnitude of muscle activation. During weight acceptance, the activation levels were no different in the lateral hamstrings or medial gastrocnemius muscles but the non-copers had significantly higher activation of the soleus on their involved sides (Table 2).

### Effect of laxity and strength on movement patterns

Several trends in the data were found in the correlations between passive joint laxity and knee movement patterns (Table 3). No relationship was found between the amount of passive knee joint laxity and peak knee flexion or external knee flexion moment during weight acceptance in the copers or non-copers. A significant correlation was found, however, between quadriceps strength and peak knee flexion during weight acceptance in the involved limbs of the non-copers only. The copers had significant correlations between lateral hamstring onset-to-peak and peak knee flexion, and knee moment at peak knee flexion, which the non-copers did not have.

Regression analyses revealed that 79.5% of the variability in the knee moment at peak knee flexion was accounted for by the variability in the onset-to-peak of the hamstrings and quadriceps muscles ($F$=6.009, $P$=0.030) in the copers only. The non-copers and control subjects showed no such relationship (non-copers: $F$=1.196, $P$=0.316; control: $F$=0.394, $P$=0.550).

**Table 3** Correlations between involved limb variables – walking

| | Knee flexion angle | External knee flexion moment |
|---|---|---|
| Passive laxity | Copers<br>$r$=–0.246, $P$=1.000 | Copers<br>$r$=0.543, $P$=0.253 |
| | Non-copers<br>$r$=–0.570, $P$=0.257 | Non-copers<br>$r$=0.247, $P$=1.000 |
| Quadriceps index | Copers<br>$r$=0.029, $P$=0.932 | Copers<br>$r$=0.135, $P$=0.693 |
| | Non-copers<br>$r$=0.933, $P$=0.000* | Non-copers<br>$r$=0.716, $P$=0.030* |
| Lateral hamstrings Onset-to-peak EMG | Copers<br>$r$=0.672, $P$=0.030* | Copers<br>$r$=0.765, $P$=0.010* |
| | Non-copers<br>$r$=0.095, $P$=0.824 | Non-copers<br>$r$=0.408, $P$=0.316 |

*Statistically significant correlation, $P$<0.05

### Jogging

Both copers and non-copers jogged more slowly than control subjects (see Table 4) and stride lengths were shorter in both ACL deficient groups during involved and uninvolved side trials.

### Kinematics and kinetics

Differences in kinematics and kinetics were similar to those seen in walking but were more pronounced. Non-copers landed with less force on the involved side (Table 4) and showed a strong trend toward less knee flexion during weight acceptance in the involved limb (Fig. 3), flexing their knees less at initial contact and achieving less peak knee flexion. The knee moment at peak knee flexion was lower in the non-copers on the involved side compared to their own uninjured leg and both legs of the copers and control subjects (Fig. 4). No differences were found in the total support moment in the different groups; however, the distribution of support moments was comparable to that seen in walking trials, i.e. lower knee and higher hip contributions (Fig. 5). No differences were seen in the ankle's contribution to the total support moment.

### Muscle activity

The differences between the groups in the muscle activity patterns were less pronounced during jogging. The only muscle timing variable that was different between the three groups was a shorter total duration of lateral hamstrings in the copers (Fig. 6). Differences were observed, however, in the magnitude of muscle activity during weight acceptance in jogging. The magnitude of electromyographic activity was significantly higher in the in-

**Table 4** ogging data. *LL* leg length

| | Copers | | Non-copers | | Controls | |
|---|---|---|---|---|---|---|
| | Involved | Un-involved | Involved | Un-involved | Involved | Un-involved |
| Velocity m/s per LL ($F$=4.00, $P$=0.03)* | 4.041 (±0.23) | 4.00 (±0.19) | 4.137 (±0.24) | 4.236 (±0.21) | 4.745* (±0.23) | 4.885* (±0.19) |
| Stride length m/LL ($F$=4.30, $P$=0.029)* | 3.089 (±0.13) | 3.034 (±0.11) | 3.194 (±0.18) | 3.297 (±0.15) | 3.575* (±0.15) | 3.592* (±0.13) |
| Vertical ground reaction force ($F$=2.849, $P$=0.075) | 2.172 (±0.07) | 2.204 (±0.07) | 2.084 (±0.07) | 2.156 (±0.07) | 2.322 (±0.07) | 2.357 (±0.07) |

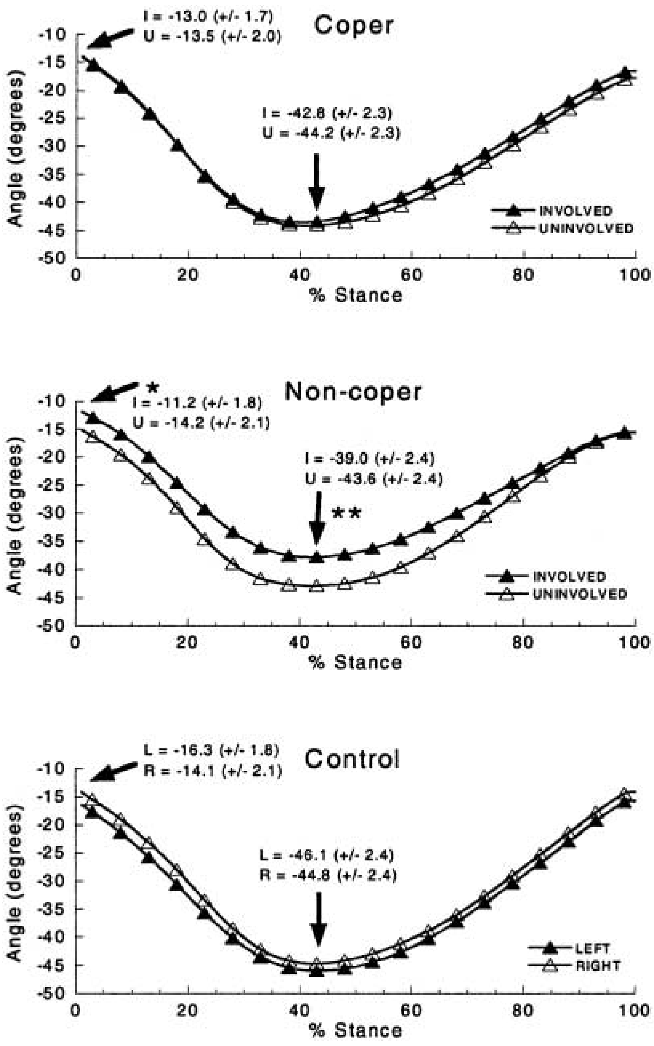


**Fig. 3** Knee flexion during jogging. There was a trend toward less knee flexion at initial contact (*$F$=4.653, $P$=0.059) and less peak knee flexion during weight acceptance (**$F$=5.112, $P$=0.050) in the involved limbs of the non-copers. *I* involved side, *U* uninvolved side

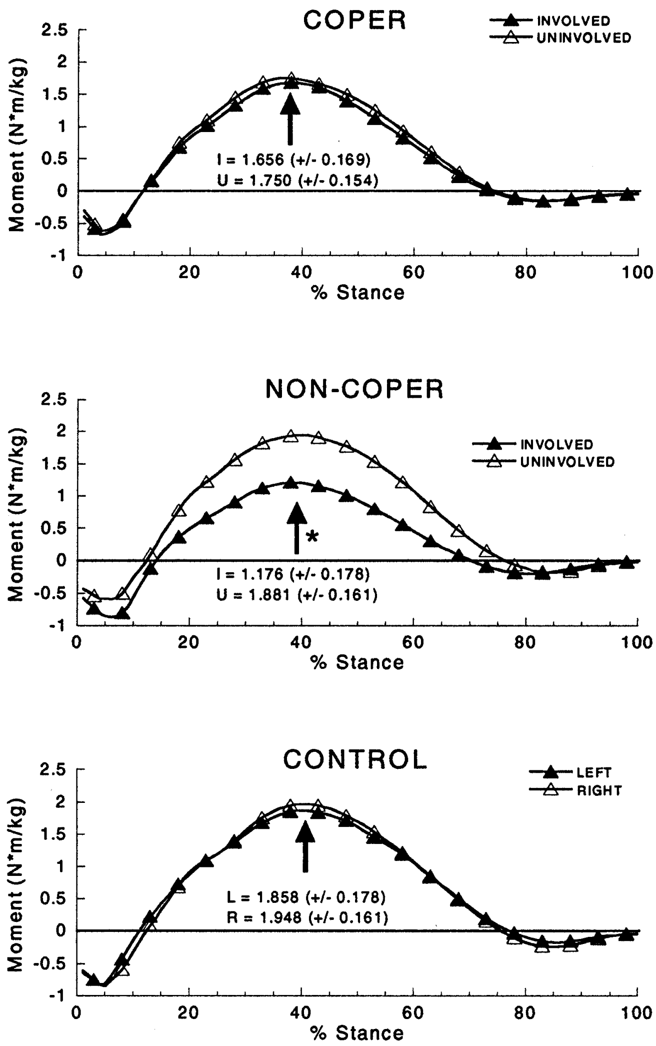


**Fig. 4** Knee moment during jogging. Non-copers had a lower knee extensor moment on the involved side than their own uninvolved sides or either sides of the copers and control subjects (*$F$=4.436, $P$=0.021). *I* involved side, *U* uninvolved side

volved hamstrings in the non-copers (Fig. 7) and there was lower VL-MG co-contraction in the copers (Fig. 8).

## Effect of laxity and strength on movement patterns

Strong correlations were found between quadriceps strength and peak knee flexion, and there was a trend toward a correlation between passive laxity and peak knee flexion in the non-copers only (see Table 5). Non-copers had a significant negative correlation between VL-LH co-contraction and knee angle at peak knee flexion. Copers showed no such correlations. In the non-copers, 83.5% of the variability in the knee moment was accounted for by the variability in the amount of VL-LH and VL-MG co-

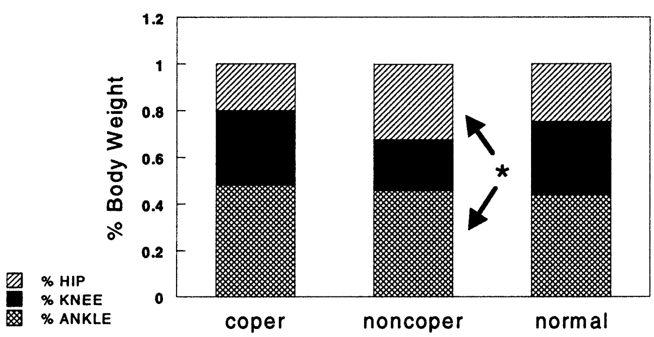


**Fig. 5** Distribution of support moments on the involved side during weight acceptance, jogging. Non-copers had significantly greater hip ($F$=3.3994, $P$=0.030) and less knee ($F$=4.727, $P$=0.017) extensor moments on the involved sides

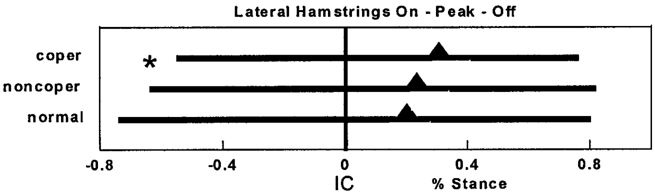


**Fig. 6** Lateral hamstring timing during jogging. Copers' hamstrings were active for a shorter duration of time than non-copers or control subjects (*$F$=3.267, $P$=0.056). *IC* indicates initial contact

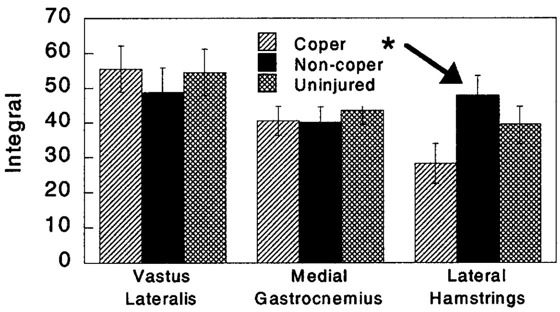


**Fig. 7** Intensity of EMG activity of the involved (or left) limbs during jogging. Non-copers had significantly greater activation of their lateral hamstrings during jogging (*$F$=4.815, $P$=0.017)

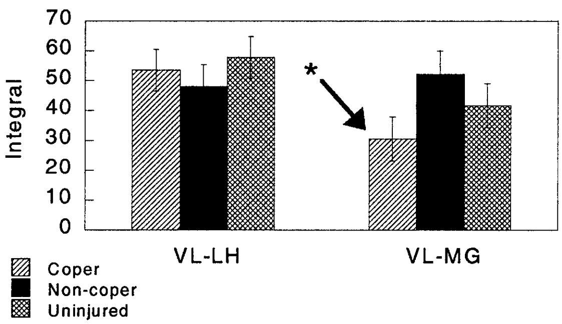


**Fig. 8** Co-contraction between vastus lateralis and lateral hamstrings (*VL-LH*) and vastus lateralis and medial gastrocnemius (*VL-MG*) during weight acceptance, jogging, in the involved limbs Non-copers had significantly greater co-contraction between vastus lateralis and medial gastrocnemius in the involved limb (*$F$=3.609, $P$=0.041)

contraction during weight acceptance ($F$=15.231, $P$= 0.004). No such relationship was found in the copers ($F$=0.585, $P$=0.582) or control subjects ($F$=0.173, $P$= 0.845).

## Discussion

The non-copers in our study were chosen because they represented the "worst of the worst" in terms of knee stability. They had instability even during their usual everyday activities and would therefore be the best subjects to reveal an unsuccessful stabilization strategy that could be

**Table 5** Correlations between involved limb variables – jogging. *NS* Not significant

| | Peak knee flexion angle | Knee moment at peak knee flexion |
|---|---|---|
| Passive laxity | Copers<br>$r$=0.203, $P$=1.000 | NS |
| | Non-copers<br>$r$=-0.866, $P$=0.015* | NS |
| Quadriceps index | Copers<br>$r$=–0.133, $P$=1.000 | NS |
| | Non-copers<br>$r$=–0.798, $P$=0.060** | NS |
| VL-LH co-contraction | Copers<br>$r$=–0.417, $P$=0.231 | NS |
| | Non-copers<br>$r$=–0.670, $P$=0.048* | NS |

*Statistically significant correlation at $P$<0.05
**Trend toward statistical significance

addressed in rehabilitation. The results of this study underscore the importance of accounting for functional ability when studying the compensation strategies of ACL deficient people. The copers moved like uninjured people with normal knee motions and moments and a knee stabilization strategy that involved less muscle co-activation than observed in non-copers. The compensations were found to be directly related to quadriceps femoris muscle strength and passive knee laxity in the non-copers only. The non-copers were shown to limit the motion at the knee joint and the limited knee flexion was related to quadriceps strength and, to some degree, passive knee joint laxity.

The copers and non-copers in this study were no different with respect to age, activity level prior to injury or the amount of passive anterior knee laxity yet they were clearly different with respect to the ability to stabilize the knee and in their movement and muscle activity patterns. The lack of relationship between quadriceps strength and laxity with knee motion and moments in the coper subjects suggests that other factors are involved in successful dynamic knee stability.

Our hypothesis that the movement patterns in ACL deficient people was related to quadriceps strength was confirmed in the non-copers who were weaker, on average, than the copers. Other investigators [11, 12, 27] have reported a significant correlation between quadriceps femoris muscle strength and knee function following ACL injury. Hurwitz et al. [12] reported significant correlations between quadriceps strength and functional deficits in people who had undergone ACL reconstruction. Snyder-Mackler et al. [28] found that knee joint motion returned to normal when quadriceps strength improved in a group of ACL reconstructed subjects. Harilainen et al. [11] found, however, that quadriceps strength only marginally correlated with functional ability in people with chronic ACL deficiency and suggested that quadriceps strength alone did not account for the differences in compensation strategies in ACL deficient individuals. In our study, two non-copers had difficulty stabilizing their ACL deficient knees despite normal quadriceps indices. This suggests that muscle activation (timing and magnitude) is an important factor in successful stabilization strategies along with quadriceps strength.

The movement patterns in the non-copers (reduced knee flexion and reduced external knee flexion moment) are consistent with those reported by other investigators [4, 17]; however, our EMG data allow us to verify the interpretation of our kinetic data. In one of the most often cited papers on ACL deficient gait, Berchuck and Andriacchi [3] explained the reduced external knee flexion moment found in 75% of their ACL deficient subjects as representing less quadriceps femoris activity and coined the term „quadriceps avoidance gait“. They postulated that in the absence of an ACL a strong quadriceps contraction may lead to an anterior shift of the tibia so the ACL deficient individuals avoided knee instability by reducing quadriceps activity. They had no EMG data to verify their interpretation. This interpretation is dangerous in that it implies a motor control strategy in which the ACL deficient individual should reduce their quadriceps activity. Although some investigators have found lower levels of quadriceps electromyographic activity during weight acceptance in ACL deficient individuals [19], most have not [4, 17, 24]. The prolonged knee flexor activity seen in our non-coper subjects has been reported by others [4, 16, 17]. Our data suggest that the reduced external knee flexion moment seen in the non-copers more accurately reflects a greater relative contribution from the knee flexors and that strong quadriceps contractions are essential for adequate joint stability.

The support moment data in this study provide evidence that non-copers attempt to stabilize their knees by transferring control away from the unstable knee to the hip. The hamstrings would provide the perfect mechanism to transfer control from the knee to the hip. Just such a transfer has been demonstrated by other investigators. Jacobs et al. found that during jogging [14], jumping and sprint starting [15] the hamstrings do indeed contribute to the transfer of power from the knee to the hip joint [15].

The reduced knee motion during weight acceptance “stiffens” the knee and suggests that the non-copers are using a knee stabilization strategy that may be putting them at risk for future joint degeneration. Less joint range of motion and muscle activity may lead to less shock absorption and increased compressive and shear forces in all joints during weight acceptance. In a canine model, Setton et al. [23] found that transection of the ACL resulted in changes in the behavior of articular cartilage in response to shear forces. The authors stated that “mechanical changes in the solid matrix of articular cartilage may be an important factor in promoting the progression of human osteoarthritis”. Benedetti et al. [2] state that co-contraction of knee flexors and extensors is a common strategy to reduce shear forces at the knee but increases joint compressive forces and joint loading. A stiffer knee joint may also prevent appropriate responses to unexpected disturbances of knee stability. The copers have no symptoms of the knee giving way thus they avoid abnormal shear forces and are able to maintain a normal knee motion, knee extensor moments and muscle timing. These characteristics bode well for the copers with respect to long-term knee joint integrity.

The non-copers had longer onset-to-peak in both the lateral hamstrings and medial gastrocnemius which may signify delayed peak hamstring muscle activation in the non-copers at a time when rapid control is needed to accept the weight of the body. Slower reflexive activation was found in ACL deficient subjects by Beard et al. [1], who found longer hamstring latency in the ACL deficient limb in response to an anteriorly directed force on the tibia. They also found a significant correlation between hamstring latency and episodes of giving way. We did not measure muscle contractile speed in our study. However, if the delay in the peak lateral hamstring timing and the longer onset-to-peak activity in the medial gastrocnemius muscles of the non-copers represents delayed force production, this would signify that the non-copers cannot balance the external moments quickly enough during weight acceptance to maintain normal joint motions and moments. In contrast, the relationship between more rapid onset-to-peak muscle activation in the lateral vastus and hamstrings and normal knee flexion moment at peak knee flexion in the copers may signify part of the successful coping mechanism involving rapid responses to changing external cues to maintain mobility and dynamic stability in the knee. Whether a more rapid onset-to-peak activity is learned or an inherent characteristic of the copers cannot be discerned from our data. However, we now have strong evidence to support the development of training programs to elicit rapid responses to varying external loads to provide more stability in the ACL deficient knee.

The reduced knee flexion moment, which has been found by us and other investigators [3, 16, 22] is present only in the non-copers and appears to be ubiquitous in individuals with ACL deficiency and substantial dynamic joint instability. We propose that reduced knee flexion moment is the hallmark of the non-coper. The reduced knee moment may be useful not only in identifying subjects with knee instability but may also prove useful in monitoring progress during rehabilitation. Our EMG data clearly demonstrate that the reduced knee flexion moment in no way represents “quadriceps avoidance”. The non-copers in this study were all recently injured and most had significant quadriceps strength deficits. Further studies are needed to clarify the relationship between quadriceps strength and functional ability and how chronicity of injury would affect the pattern of reduced external knee flexion moment in non-copers.

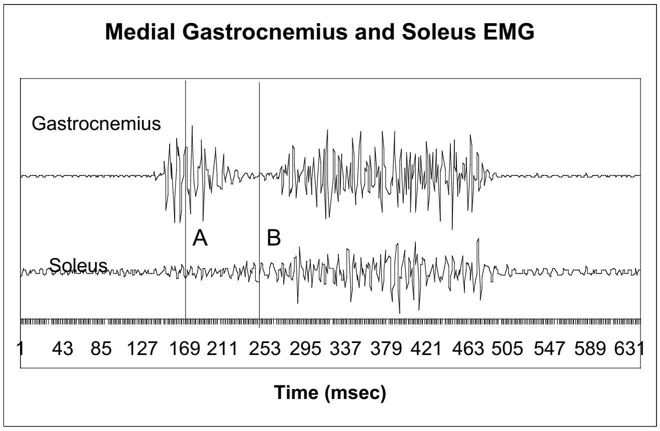


**Fig. 9** Medial gastrocnemius and soleus EMG activity from a typical trial

## Appendix 1

In designing this study, the differences in movement patterns between the copers and non-copers was expected to be small, particularly during walking. We therefore chose to use surface electrodes in to ensure that the discomfort of intramuscular fine wire electrodes would not affect the movement patterns of the subjects. Surface electrodes, however, can record "cross-talk" from adjacent muscles. Cross-talk is EMG activity that is from a muscle other than the one of interest. In our study, the cross-talk between the gastrocnemius and soleus muscles was a possibility, so we carefully inspected our data to ensure that no cross-talk was present. The figure below demonstrates the medial gastrocnemius (MG) and soleus (SOL) EMG activity from a typical subject. The reader can see that the large burst of EMG from the MG, signified by the letter "A" was not picked up by the SOL electrode. Similarly, the quiescent period of MG EMG, signified by the letter "B" occurs at a time when there is increasing activity seen on the SOL electrode so cross-talk was absent in the gastrocnemius recording as well (Fig. 9).

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