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Differences in the movement pattern of a forward lunge in two types of anterior cruciate ligament deficient patients: copers and non-copers

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Abstract

Objective. To determine whether differences in the knee joint movement pattern of a forward lunge could be quantified in healthy subjects and in anterior cruciate ligament deficient subjects who were able to return to the same activity level as before their injury (*copers*) and in those who were not (*non-copers*).

Design. The movement patterns of the injured leg of the coper and non-coper anterior cruciate ligament deficient subjects and the right leg of the control subjects were compared statistically.

Background. The forward lunge seems to be a less stressful test than the commonly used one-legged hop test, which makes it a possible tool for evaluating and comparing the functional performance of non-copers and copers.

Methods. The movement pattern of a forward lunge was analysed by using a two-dimensional inverse dynamics method. The electromyographic activity of the quadriceps and hamstring muscles were recorded.

Results. The non-copers moved more slowly and loaded the knee joint less than the copers and controls. The copers moved more slowly during the knee flexion phase but as fast as the controls during the knee extension. The EMG results suggest that the copers stabilized their knee joint by increasing the co-contraction of the hamstrings during the extension phase.

Conclusions. Differences between the three groups' movement patterns could be quantified. The forward lunge test seems appropriate to discriminate between the knee function in coper and non-coper anterior cruciate ligament deficient subjects.

Relevance

Information about the performance of movements, which significantly load the knee joint in coper and non-coper anterior cruciate ligament deficient patients may contribute to a better understanding of dynamic knee joint stabilization, which is relevant in relation to the development of rehabilitation strategies.

Keywords: Anterior cruciate ligament deficiency; Inverse dynamics; Functional testing; Movement pattern; Electromyography

1. Introduction

Biomechanical and electromyographic (EMG) studies have shown altered kinematics, kinetics and muscular activation during different types of movements such as walking, side-cutting and the one-legged hop for distance in anterior cruciate ligament (ACL) deficient patients [1–5]. The changed kinematics, kinetics and EMG patterns have been interpreted as having a possible protective effect on the knee joint, thereby enabling the ACL deficient (ACLD) subjects to participate in different types of activities [2,3,6,7]. There are different ways to compensate for the ACL deficiency by altering the movement pattern [4]. Previous studies have demonstrated that some ACLD patients are able to return to the same activity level as before their injury (*copers*)

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while others are not (*non-copers*) [7–9]. It is possible that these two types of ACLD patients move according to different patterns [7,10,11].

The one-legged hop test is a commonly used clinical assessment tool to evaluate the functional ability in ACLD patients [4]. Rudolph et al. [12] observed differences in the movement pattern of a one-legged hop performed by non-copers, copers and healthy subjects. However, only four of the non-copers (i.e. 40%) participating in their study were willing to perform the one-legged hop test. Likewise, Barber et al. [13] reported that 60% of the ACLD patients in their study had problems with jumping and landing on their injured limb. It is therefore questionable if the one-legged hop test is a useful assessment tool.

Dynamic instability of the knee joint experienced by ACLD individuals seems to be an essential factor in determining whether they become copers or non-copers [12]. Quantification of the movement and muscle activation patterns of more knee-provoking activities would contribute to a better understanding of the dynamic stabilization of the knee joint in copers and non-copers, which, moreover, could be used in the development of rehabilitation strategies of non-coper ACLD subjects.

The purpose of the present study was to determine whether differences in the movement pattern of the knee joint could be quantified in healthy subjects and in coper and non-coper ACLD subjects during a forward lunge.

2. Methods

2.1. Subjects

Seventeen male patients with complete unilateral ACL deficiency participated in the study. All the patients had an activity level of minimum two hours per week and all of them had been through rehabilitation programs for at least six months after their injury. The Tegner and Lysholm [14] and Tegner et al. [15] scores were carefully used to interview the patients about their knee function after the ACL rupture. The patients were divided into two groups, i.e. copers and non-copers, according to their activity level. The copers were defined as those who were able to return to their normal activity level despite their injury (copers), while the non-copers were defined as those who were unable to return to the same activity level as before their injury. The copers consisted of eight subjects (weight: 76.6 kg (SD, 14.8), height: 1.81 m (SD, 0.06), age: 26.0 years (SD, 4.0)). The mean Lysholm and Tegner scores of the copers were 85.5 (SD, 5.3) and 6.25 (SD, 0.5), respectively and the mean time lapse between injury and testing was 34.0 months (SD, 39.2) (range 6.0–120.0). The non-copers comprised nine subjects (weight: 80.6 kg (SD, 7.1), height: 1.79 m (SD, 0.06), age: 31.2 years (SD, 6.0)) with a mean Lysholm and Tegner scores of 74.0 (SD, 7.1) and 3.8 (SD, 0.6), respectively. The mean time lapse between injury and testing of the non-copers was 51.8 months (SD, 44.0) (range 6.0–144.0).

Nine healthy male subjects (weight: 75.6 kg (SD, 7.0), height: 1.83 m (SD, 0.04), age: 31.0 years (SD, 5.7)) were selected as controls for the biomechanical analysis. Electromyography was recorded in six of the control subjects (weight: 73.8 kg (SD, 7.9), height: 1.81 m (SD, 0.05), age: 31.0 years (SD, 1)). There were no statistically significant differences between the groups with regard to weight, height and age. All subjects gave their informed consent to participate in the experiments, which were approved by the local ethics committee.

2.2. Procedure

When the subjects arrived at the laboratory the investigator instructed them in how to perform the forward lunge. Five small reflecting spherical markers (12-mm diameter) were placed on the head of the fifth metatarsal, the lateral malleolus, the lateral femoral epicondyle, the greater trochanter and the anterior superior iliac spine on the injured (ACLD) leg of the patients and the right leg of the control subjects. All the subjects were dressed in a tight black suit and wore lightweight flexible shoes.

The subjects stood in an upright position in front of a force plate (model OR6-5-1, AMTI, Watertown, MA, USA). They were instructed to perform a forward lunge by taking one step forward, placing the foot on the force plate, flexing the knee to approximately 90° and subsequently extending the knee to push themselves backwards into the starting position. The subjects were asked to keep their upper body perpendicular to the ground and leave the contralateral foot in contact with the ground during the whole movement. The subjects were allowed to practice the movement for as long as they wanted. All the ACLD subjects were asked to report if they felt uncomfortable (e.g. pain, giving-way symptoms) when performing the task.

After being accustomed to the testing procedure, the subjects performed three consecutive forward lunges, which were recorded and used in the biomechanical analysis of the movement.

Bipolar surface electrodes (Medicotest N-00-S, 2 cm inter-electrode distance) were then placed on quadriceps (m. vastus lateralis (VL), and m. vastus medialis (VM)) and the hamstring muscles (m. semitendinosus (ST), and m. biceps femoris (BF)) on the right leg of the controls and the injured leg of the ACL deficient subjects. The EMG signals were recorded throughout 15 consecutive forward lunges. The subjects were allowed to rest between the trials for as long a time as they wanted to avoid fatigue [16].

2.3. Biomechanical analysis

Five video cameras operating at 50 frames per second were used to record the movements. The video signals and the force plate signals were synchronized electronically with a custom-built device. The device put a visual marker on one video field from all cameras and at the same time triggered the analogue-to-digital converter that sampled the force plate signals at 1000 Hz. The subjects initiated the data sampling and synchronization when they passed a photocell, which was placed in front of the force plate.

The video sequences were digitised and stored on a PC. Three-dimensional co-ordinates were reconstructed by direct linear transformation using the ariel performance analysis system (APAS, Ariel Dynamics Inc., San Diego, CA, USA). Prior to the calculations, the position data were digitally low-pass filtered by a fourth order zero-lag Butterworth filter with a cut-off frequency of 6 Hz, and the 1000 Hz force plate signals were resampled at 50 sample points per second.

A two-dimensional (2D) biomechanical approach was used to calculate internal flexor and extensor net joint moments about the ankle, knee and hip joint. The 2D inverse dynamics model was based on the free-body segment method [17].

The anthropometric data from Chandler et al. [18] and anthropometric measurements from each subject were used to calculate segment masses, moments of inertia and centres of mass. 2D joint moments were computed in MATLAB by modification of code presented by Van den Bogert and de Koning [19]. Ankle dorsi flexor, knee extensor and hip flexor moments were considered positive, while ankle plantar flexor, knee flexor and hip extensor moments were considered negative.

The angular position of the knee joint was calculated in order to describe the movements in the sagittal plane. Zero degrees defined the anatomical position of full extension and positive values reflected hyperextension of the knee joint. The joint angular velocity was calculated by differentiation of the angular position.

The knee joint power was calculated by multiplying the knee joint moment and the joint angular velocity.

2.4. Electromyography

The EMG electrodes were connected to small custom-built pre-amplifiers (input impedance 80 MΩ, gain = 50). The EMG signals were then led through long shielded wires to custom-built amplifiers with a frequency response between 20 Hz and 10 kHz and sampled at 1000 Hz. Before each test the EMG signals were controlled visually for cross-talk and movement artifacts. The EMG signals were sampled for 2 s with a pre-trigger time of 200 ms. The recordings were triggered when the subject hit the force plate.

The EMG signals were digitally high- and low-pass filtered (Butterworth fourth order zero-lag digital filter, cut-off frequencies 20 Hz and 500 Hz, respectively), full-wave rectified and low-pass filtered at 15 Hz to create linear envelopes. Linear envelopes from 15 trials were used to calculate the average EMG of each muscle for each subject. Ensemble averages were then calculated for the non-copers ($n = 9$), copers ($n = 8$) and control subjects ($n = 6$) using the individual subject means. The mean and peak amplitudes of the linear envelopes were calculated for each muscle over the whole the movement phase. All signals were expressed in microvolts.

2.5. Normalization and data reduction

The biomechanical data were normalized and averaged for each subject. Normalization was performed in MATLAB by interpolating data points to form 100 samples for each trial. Ensemble averages were then calculated for the non-copers ($n = 9$), copers ($n = 8$) and the control group ($n = 9$) using the mean curve for each individual subject. The ground reaction forces were normalized to body mass (N/kg). Furthermore, as the first and last 25% of the movement phase represented the most critical range of motion (RoM) (from approximately 15° to 45° of knee flexion) in which the quadriceps was able to cause anterior tibial translation [20], the knee joint angle and knee extensor moment were averaged within the intervals from 0% to 25% and 75% to 100% of the movement phase.

2.6. Statistics

Peak and mean values of the EMG amplitudes, vertical and anteroposterior ground reaction force components, the knee joint angle, angular velocity, moment and power were compared statistically between the ACLD leg of the non-copers and copers and the right leg of the control subjects by using a one-way analysis of variance test (ANOVA). In cases with significant group effects, the Student–Newman–Keuls method was used to locate the differences. The level of significance was set at 5%.

3. Results

All the subjects were able to perform the forward lunge without any discomfort.

The absolute duration of the movement was significantly longer for the non-copers than the control subjects, while there was no significant difference in the duration of the movement between the copers and the control subjects (Table 1). It took the non-copers approximately 27% longer to complete the forward lunge compared to the controls.

Table 1
Absolute time of movement

Variable	Non-copers ($n = 9$)	Copers ($n = 8$)	Control ($n = 9$)	P-value
Absolute time (s)	1.26 (0.14)	1.08 (0.24)	0.99 (0.22)[a]	0.027

Values are means (SD).
[a] Significant difference between non-copers and control.

3.1. Ground reaction forces

The vertical and anteroposterior ground reaction force components normalized to body mass are displayed in Fig. 1 (top panel). The vertical ground reaction force (F_y) did not differ significantly between the three groups (Fig. 1, Table 2). The second F_x peak (F_{x2}) of the controls and the copers was significantly larger than that of the non-copers (Fig. 1, Table 2).

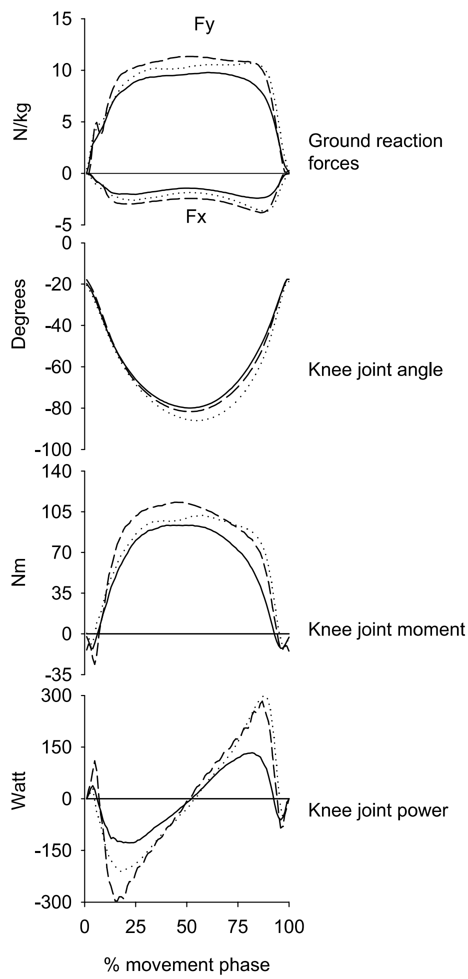

Fig. 1. Average vertical (F_y), anteroposterior (F_x) ground reaction force components, knee joint angle, moment and power curves of non-copers ($n = 9$, solid lines), copers ($n = 8$, dotted lines) and control subjects ($n = 9$, dashed lines). A negative F_x indicates that the direction of the force vector is posterior. A positive knee joint moment indicates that the joint is dominated by the extensors. 0% on the x-axis is heel strike and 100% is heel-off.

3.2. Kinematics

There were no significant differences in the knee joint angle between the copers, non-copers and control subjects (Fig. 1, Table 3). The knee joint angle of the copers peaked at 54.9% (SD, 4.4%) of the movement phase, while that of the non-copers and controls peaked at 50.1% (SD, 7.5%) and 51.1% (SD, 2.1%), respectively (Fig. 1). However, no significant difference was observed in this variable ($P = 0.164$).

The peak joint angular velocity of the knee flexion was significantly higher in the control subjects than in the non-copers and the copers (Table 3). The peak joint angular velocity of the knee extension was significantly higher in the controls and the copers than in the non-copers (Table 3).

3.3. Joint moments and power

The average knee extensor moment measured within the last 25% of the movement phase was significantly smaller in the non-copers than in the copers and the controls (Fig. 1, Table 3). No differences between the groups were observed in the peak knee extensor moment and the average knee extensor moment measured within the first 25% of the movement (Table 3).

The negative peak power of the knee extensors was significantly smaller in the non-copers than in the controls (Fig. 1, Table 4). The positive peak power of the knee extensors was significantly larger in the copers and the control subjects than in the non-copers (Fig. 1, Table 4).

3.4. Electromyography

The mean and peak amplitude of the VM was significantly higher in the copers (mean: 182.4 μV (SD, 87.4), peak: 483.3 μV (SD, 352.2)) than in the non-copers (mean: 98.3 μV (SD, 48.9), peak: 189.3 μV (SD, 89.8)) (mean: $P = 0.05$, peak: $P = 0.041$) (Fig. 2). The mean amplitude of the ST was significantly higher in the copers (27.3 μV (SD, 12.8)) than in the non-copers (17.4 μV (SD, 5.0)) and the controls (14.6 μV (SD, 4.4)) ($P = 0.023$) (Fig. 2). No significant differences in the mean and peak amplitude of the BF and VL were observed between groups (Fig. 2).

Table 2
Vertical (F_y) and anteroposterior (F_x) ground reaction force components

Variable (N/kg)	Non-copers ($n = 9$)	Copers ($n = 8$)	Control ($n = 9$)	P-value
Peak F_y	10.3 (1.0)	11.9 (2.1)	11.7 (0.9)	0.063
Peak F_{x1}	−2.3 (0.5)	−3.0 (0.8)	−3.1 (0.9)	0.062
Peak F_{x2}	−2.6 (0.8)	−4.1 (0.8)[a]	−4.0 (1.1)[b]	0.003

Values are means (SD).
[a] Significant difference between non-copers and copers.
[b] Significant difference between non-copers and controls.

Table 3
Knee joint angle and extensor moment averaged within the first and last 25% of the movement phase (peak values are representative for the entire movement phase)

Variable	Non-copers ($n = 9$)	Copers ($n = 8$)	Control ($n = 9$)	P-value
Knee joint angle (deg)				
1–25%	−43.9 (7.4)	−45.4 (10.3)	−43.3 (5.3)	0.864
75–100%	−44.6 (8.8)	−51.5 (9.3)	−46.7 (4.4)	0.192
Peak flexion	−80.6 (11.9)	−86.4 (15.0)	−81.8 (4.6)	0.539
Knee joint angular velocity (deg s^{-1})				
Knee flexion	−221.5 (26.3)	−256.1 (63.6)	−311.3 (58.8)[a]	0.004
Knee extension	250.3 (84.8)	371.2 (66.6)[b]	360.9 (94.4)[c]	0.01
Knee joint moment (Nm)				
1–25%	33.4 (18.3)	40.6 (14.4)	43.6 (16.5)	0.419
75–100%	29.8 (12.4)	54.7 (14.1)[b]	46.5 (16.1)[c]	0.005
Peak moment	100.3 (23.8)	110.1 (26.1)	116.1 (22.3)	0.345

Values are means (SD).
[a] Significant difference between controls and copers and controls and non-copers.
[b] Significant difference between non-copers and copers.
[c] Significant difference between non-copers and controls.

Table 4
Negative and positive peak power of the knee extensor muscles

Variable (W)	Non-copers ($n = 9$)	Copers ($n = 8$)	Control ($n = 9$)	P-value
Negative power	−180.7 (48.0)	−245.7 (107.9)	−318.6 (123.7)[a]	0.024
Positive power	155.8 (72.7)	361.1 (115.4)[b]	306.8 (141.2)[a]	0.003

Values are means (SD).
[a] significant difference between non-copers and controls.
[b] significant difference between non-copers and copers.

4. Discussion

The purpose of the present study was to determine whether the forward lunge could be used to quantify differences in the knee joint movement pattern in coper and non-coper ACLD subjects. The main findings were that non-copers performed the movement significantly more slowly and with a reduced knee extensor moment than the controls and copers. The copers performed the knee flexion part of the forward lunge more slowly than the controls. However, the copers completed the knee extension as fast as the controls, probably because they were able to stabilize the knee joint by increasing the co-contraction of the hamstring muscles.

Previous studies have reported that non-coper ACLD subjects may be unable to perform the one-legged hopping test [12,13]. Therefore, a forward lunge was chosen for the movement test because, like the one-leg hop test and many sports activities, it consists of a knee flexion and a subsequent knee extension controlled by an eccentric and concentric contraction respectively of the quadriceps. Contraction of the quadriceps may cause anterior translation of the tibia [20–25] which stresses the knee joint—especially in an ACL deficient situation. However, all ACLD patients who participated in the present study were able to perform the forward lunge without any knee discomfort or giving-way symptoms. The lack of any discomfort may be explained by the fact that the exercise primarily took place in the sagittal plane, which minimized knee joint rotations. Furthermore, both feet were in contact with the ground during the movement and there was no landing phase as in the

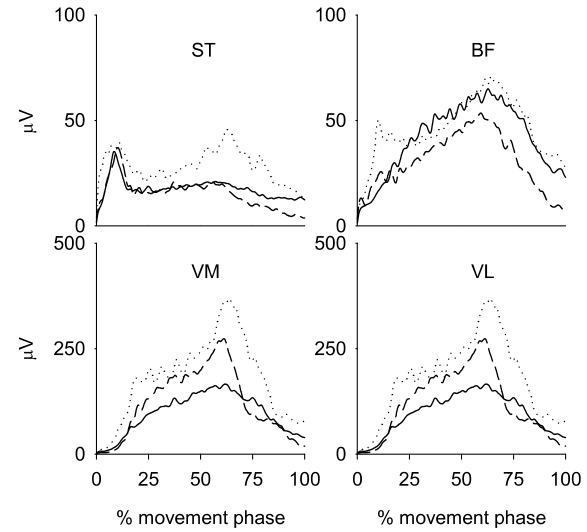

Fig. 2. Average linear envelopes (μV) of the m. semitendinosus (ST), m. biceps femoris (BF), m. vastus lateralis (VL) and m. vastus medialis (VM) of the non-copers ($n = 9$, solid lines), copers ($n = 8$, dotted lines) and the control subjects ($n = 6$, dashed lines). 0–100% is heel strike and heel-off of the forward lunge. Maximal knee flexion occurs at approximately 50% of the movement phase.

one-legged hopping test whereby the eccentric loading on the knee joint was reduced during the knee flexion phase.

The subjects were allowed to perform the forward lunge in their own way. Only few instructions were given about how the subjects should position their body during the movement, such as keeping their upper body perpendicular to the ground and their feet in contact with the floor. Despite these marginally restrictive test conditions the results showed that the three groups moved according to different movement patterns. The copers performed the forward lunge much as the control subjects, while the movement pattern observed in the non-copers was different (see Table 1 and the knee flexion patterns in Fig. 1). This is consistent with the results reported by Rudolph et al. [12], who observed that coper ACLD subjects performed a one-legged hop almost identical to that of healthy subjects while the non-copers differed from these two groups. The forward lunge test therefore seems to be an appropriate way of identifying movement pattern differences between coper and non-coper ACLD subjects.

In the present study, a two-dimensional link-segment model was used to quantify the movement patterns. This is a simple method that can easily be applied in a clinical setting since only one video camera and one force plate are needed. However, some limitations of the method should be addressed here. Firstly, it is important to note that a joint moment only reflects the net effect of the activity at the joint, meaning that if a co-contraction is involved, the analysis will only yield the net effect of both agonist and antagonistic muscles. It is therefore necessary to supplement the biomechanical analysis with EMG recordings of the knee extensor and flexor muscles activity to get an idea of the degree of co-contraction. Secondly, although it has been demonstrated that kinematic and kinetic data of movement analyses are repeatable [26] one should be aware of the fact that even small variations in the marker placement, which defines the joint axes, may affect the results considerably [27].

The time needed to complete the task appeared to be a key factor in the movement pattern of the non-copers. The non-copers spent more time on the forward lunge than the control subjects, which meant that the peak angular velocity of the knee joint of both the flexion and extension was significantly reduced in the non-copers. This "slow motion strategy" adopted by the non-copers could possibly be interpreted as an attempt to reduce the quadriceps force needed to decelerate and accelerate the body segment masses during the knee flexion and extension. A reduced quadriceps force and thus a reduced knee extensor moment would decrease the anterior translation of the tibia [23,28]. The first and last 25% of the movement phase are the most critical RoM (45–0° of knee flexion) since quadriceps contraction can cause considerable anterior displacement of the tibia relative to the femur [20]. There was no difference in the knee extensor moment during knee flexion (i.e. within the first 25% of the movement phase) whereas the knee extensor moment during knee extension (i.e. within the last 25% of the movement phase) was significantly lower in the non-copers compared to the copers and the controls. The results of the ground reaction forces showed that the peak of the horizontal ground reaction force during knee extension (F_{x2}) was significantly lower in the non-copers compared to the copers and control subjects. This indicates that the non-copers attempted to slow down the backward acceleration of the body during the extension phase, possibly to reduce the load on the knee joint.

The peak power of the knee extensors during knee flexion and extension was significantly lower in the non-copers than in the copers and the controls. Presumably, the difference in the power for the non-copers was mainly due to the slower angular velocity in the flexion part and to both the reduced moment and slower angular velocity during the extension part.

The copers performed the forward lunge very similarly to the controls. However, differences in the peak angular velocity of the knee flexion and the amplitude of the medial hamstring muscle were observed between the two groups indicating that the copers to some extent used a different strategy to accomplish the movement. The peak angular velocity of the knee flexion was lower in the copers than in the controls.

A slower angular velocity during knee flexion has also been observed in other studies of ACLD subjects [4,12] and may be explained by the fact that the hamstring muscles have a limited capacity for dynamic knee joint stabilization during fast knee flexion movements controlled by eccentric quadriceps muscle contractions [29–31]. The reduced angular velocity did not result in a significantly lower knee extensor moment or peak power during the knee flexion of the forward lunge in the copers. Perhaps the slower angular velocity reflects a safety motor program in the copers, which allows them to react if they suddenly experience knee joint instability or pain during the knee flexion.

The copers performed the knee extension part of the forward lunge in the same manner as the control subjects. The peak angular velocity indicated that the copers were able to perform the knee extension as fast as the control subjects. This may be explained by the fact that the copers stabilized their knee joint by increasing the co-contraction of the hamstring muscles since the mean amplitude of the medial hamstring muscle was significantly higher in the copers than in the non-copers and control subjects. It has been demonstrated that the capacity of hamstring muscles to counteract the action of the quadriceps is high, even during fast concentric knee extension movements [30].

An increased activity of the hamstring muscles would theoretically decrease the calculated net knee extensor moment unless the quadriceps activity is increased proportionally. The mean and peak amplitudes of the m. vastus medialis were significantly higher in the copers than in the non-copers but not in the controls. The lack of a significant difference between the copers and the controls may be due to the small sample size of the control subjects ($n = 6$), which renders any conclusions based on EMG results tentative. The described investigations must be performed on a larger sample size before the differences in muscle activation pattern between the three groups can be clarified, especially with regard to coactivation patterns of the quadriceps and hamstring muscles.

5. Conclusion

The results of this study showed that differences in the movement pattern of a forward lunge could be quantified in non-copers, copers and healthy subjects.

The non-copers performed both the knee flexion and the extension of the forward lunge more slowly than the control subjects. Moreover, they performed the knee extension with a smaller knee extensor moment probably in order to reduce the anterior translation of the tibia relative to the femur. The copers also performed the knee flexion more slowly but, their knee extension was as fast as the controls. The latter may be explained by the fact that the copers stabilized the knee joint by increasing the co-contraction of the hamstring muscles. The reduced peak angular velocity of the knee flexion observed in both the copers and the non-copers may reflect a safety motor program, which enables the ACLD subjects to react if they suddenly experience knee joint pain or giving-way symptoms since the ability of the hamstring muscles to stabilize the knee joint in this phase is marginal. However, the exact mechanisms responsible for the dynamic knee joint stability are still unclear. Further investigations of e.g. co-contraction of the quadriceps/hamstring muscles in coper and non-coper ACLD subjects are necessary to explain the different movement patterns observed in these subjects. Moreover, the forward lunge test may be an appropriate functional test that can be used to evaluate the effect of different kinds of interventions, such as training and ACL reconstruction in coper as well as non-coper ACLD subjects.

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References

[1] Kaalund S, Lass P, Le Fevre S, Sinkjær T, Arendt-Nielsen L, Simonsen O. Clinical significance of altered gastrocnemius muscle coordination in anterior cruciate ligament-deficient knees. Scand J Med Sci Sports 1992;2:79–83.
[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;72:871–7.
[3] Branch TP, Hunter R, Donath M. Dynamic EMG analysis of anterior cruciate deficient legs with and without bracing during cutting. Am J Sports Med 1989;17:35–41.
[4] Gauffin H, Tropp H. Altered movement and muscular-activation patterns during the one-legged jump in patients with an old anterior cruciate ligament rupture. Am J Sports Med 1992;20:182–92.
[5] Beard DJ, Soundarapandian R, O'Connor J, Dodd C. Gait and electromyographic analysis of anterior cruciate ligament deficient subjects. Gait Posture 1996;4:83–8.
[6] Devita P, Hortobagyi T, Barrier J, Torry M, Glover KL, Speroni DL, et al. Gait adaptations before and after anterior cruciate ligament reconstruction surgery. Med Sci Sports Exerc 1997; 29:853–9.
[7] Rudolph KS, Eastlack ME, Axe MJ, Snyder-Mackler L. 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;8:349–62.
[8] Noyes FR, Matthews DS, Mooar PA, Grood ES. The symptomatic anterior cruciate-deficient knee. Part II: the results of rehabilitation, activity modification, and counseling on functional disability. J Bone Joint Surg [Am] 1983;65:163–74.
[9] Eastlack ME, Axe MJ, Snyder-Mackler L. Laxity, instability, and functional outcome after ACL injury: copers versus noncopers. Med Sci Sports Exerc 1999;31:210–5.
[10] Sinkjaer T, Arendt-Nielsen L. Knee stability and muscle coordination in patients with anterior cruciate ligament injuries: an electromyographic approach. J Electromyogr Kinesiol 1991;1: 209–17.

[11] Ciccotti MG, Kerlan RK, Perry J, Pink M. An electromyographic analysis of the knee during functional activities. II. The anterior cruciate ligament-deficient and -reconstructed profiles. Am J Sports Med 1994;22:651–8.

[12] Rudolph KS, Axe MJ, Snyder-Mackler L. Dynamic stability after ACL injury: who can hop? Knee Surg Sports Traumatol Arthrosc 2000;8:262–9.

[13] Barber SD, Noyes FR, Mangine RE, McCloskey JW, Hartman W. Quantitative assessment of functional limitations in normal and anterior cruciate ligament-deficient knees. Clin Orthop 1990;255:204–14.

[14] Tegner Y, Lysholm J. Rating systems in the evaluation of knee ligament injuries. Clin Orthop 1985;198:43–9.

[15] Tegner Y, Lysholm J, Odensten M, Gillquist J. Evaluation of cruciate ligament injuries: a review. Acta Orthop Scand 1988;59: 336–41.

[16] Pincivero DM, Aldworth C, Dickerson T, Petry C, Shultz T. Quadriceps-hamstring EMG activity during functional, closed kinetic chain exercise to fatigue. Eur J Appl Physiol 2000;81: 504–9.

[17] Winter DA. Biomechanics and motor control of human movement. New York: John Wiley & Sons; 1990.

[18] Chandler RF, Clauser CE, McConville JT, Reynolds HM, Young JW. Investigation of inertial properties of the human body. Aerospace Medical Research Laboratory 1975; Tech. Rep. No. 74-137.

[19] Van den Bogert AJ, de Koning JJ. An optimal filtering for inverse dynamics analysis. IXth Biennial Conference, Canadian Society for Biomechanics, 1996.

[20] Renström P, Arms SW, Stanwyck TS, Johnson RJ, Pope MH. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med 1986;14:83–7.

[21] Arms SW, Pope MH, Johnson RJ, Fischer RA, Arvidsson I, Eriksson E. The biomechanics of anterior cruciate ligament rehabilitation and reconstruction. Am J Sports Med 1984;12:8–18.

[22] Draganich LF, Vahey JW. An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. J Orthop Res 1990;8:57–63.

[23] Hirokawa S, Solomonow M, Lu Y, Lou ZP, D'ambrosia R. Anterior–posterior and rotational displacement of the tibia elicited by quadriceps contraction. Am J Sports Med 1992;20:299–306.

[24] More RC, Karras BT, Neiman R, Fritschy D, Woo SL, Daniel DM. Hamstrings—an anterior cruciate ligament protagonist. An in vitro study. Am J Sports Med 1993;21:231–7.

[25] MacWilliams BA, Wilson DR, DesJardins JD, Romero J, Chao EY. Hamstrings co-contraction reduces internal rotation, anterior translation, and anterior cruciate ligament load in weight-bearing flexion. J Orthop Res 1999;17:817–22.

[26] Kadaba MP, Ramakrishnan HK, Wootten ME, Gainey J, Gorton G, Cochran GV. Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. J Orthop Res 1989; 7:849–60.

[27] Cappozzo A. Three-dimensional analysis of human walking: experimental methods and associated artifacts. Human Movement Sci 1991;10:589–602.

[28] Nisell R, Ericson MO, Nemeth G, Ekholm J. Tibiofemoral joint forces during isokinetic knee extension. Am J Sports Med 1989;17:49–54.

[29] Aagaard P, Simonsen EB, Trolle M, Bangsbo J, Klausen K. Isokinetic hamstring/quadriceps strength ratio: influence from joint angular velocity, gravity correction and contraction mode? Acta Physiol Scand 1995;154:421–7.

[30] Aagaard P, Simonsen EB, Magnusson SP, Larsson B, Dyhre-Poulsen P. A new concept for isokinetic hamstring: quadriceps muscle strength ratio. Am J Sports Med 1998;26:231–7.

[31] Simonsen EB, Magnusson SP, Bencke J, Naesborg H, Havkrog M, Ebstrup JF, et al. Can the hamstring muscles protect the anterior cruciate ligament during a side-cutting maneuver. Scand J Med Sci Sports 2000;10:78–84.