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Hamstrings co-activation in ACL-deficient subjects during isometric whole-leg extensions

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Abstract It has been reported that anterior cruciate ligament (ACL)-deficient subjects increase the level of hamstrings activation and this has been interpreted as a means to cope with increased anterior tibial laxity in the knee. This study aimed to establish to what extent co-activation strategies in ACL-deficient subjects are load level and knee angle dependent. Eleven chronic ACL-deficient and 15 control subjects were positioned in a range of postures and asked to exert a feedback controlled vertical ground reaction force (GRF; 30, 60% and maximum), while horizontal forces were not constrained. Surface electromyography of the leg muscles and GRF were measured. In postures with the knee over and in front of the ankle, ACL-deficient subjects generated, respectively, 2.4 and 5.1% MVC more hamstrings activation than control subjects. Enhanced hamstrings co-activation in ACL-deficient subjects was more apparent in extended than in flexed knee angles. For both ACL-deficient and control subjects, hamstrings co-activation was larger in males than in females. It is concluded that ACL-deficient subjects show a task dependent increase in hamstrings co-activation, but its clinical significance remains to be shown.

Keywords Anterior cruciate ligament · Biomechanics · Electromyography · Co-activation

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Introduction

Following the rupture of the anterior cruciate ligament (ACL), the laxity of the knee joint increases and patients often experience dynamic instability during daily activities. Although reconstructive surgery is frequently applied, some recent studies suggest that it neither improves rotational knee stability [29] nor it reduces the risk of developing osteoarthritis [21]. Therefore, conservative treatment may remain an important treatment option, at least in the near future. Understanding adaptation in muscle activation patterns in ACL-deficient subjects may help to improve conservative treatment strategies.

Besides providing a knee extension moment, quadriceps activation causes an anterior shear force on the tibia relative to the femur and, therefore, strains the ACL at knee angles between 0° and 50° [6, 10, 14]. Electrical [11] as well as mechanical [27] stimulation of the intact ACL is known to elicit hamstring activation, which suggests that the hamstrings might aid the ACL or compensate for its absence [28]. Indeed, many authors reported enhanced hamstrings co-activation in ACL-deficient subjects during functional activities such as gait [8, 18, 25, 31].

Larger knee extension moments will induce larger shear challenges to the knee joint and are thus likely to cause enhanced hamstrings co-activation in ACL-deficient subjects. Furthermore, co-activation may change with knee flexion because, with increasing knee flexion, the backward angle of the hamstrings tendons increases, whereas the forward angle of the patellar tendon decreases relative to the tibial shaft [3, 24]. It should be noted that, unless reduced knee extension moments are accepted, enhanced co-activation requires not just increased activation of the hamstrings, but also of the quadriceps. It has been predicted that pure co-activation of the hamstrings and quadriceps cannot directly reduce anterior tibial shear forces in the knee at small (up to 22°) angles [7, 24], since, at these joint angles, the angle of the patellar tendon is larger than the angle of the hamstring tendons [3]. Therefore, close to full extension, ACL-deficient subjects may need to reduce quadriceps activation [5] in order to reduce shear forces in the knee.

In gait, kinematics, knee moments and muscle activation may all differ between ACL-deficient and healthy subjects [8, 9, 26]. This renders it difficult to determine to what extent the adaptation of muscle activation depends on knee angle and on knee moment level.

Studying muscle activation strategies under isokinetic or isometric conditions in single joint exertions decreases the complexity and may, therefore, be useful to examine control strategies in more detail [1, 2, 13]. The disadvantage of single joint exertions is, however, that subjects are unable to compensate a moment reduction in one joint by an increase in another joint. To allow compensation over joints, while increasing the level of experimental control relative to functional activities, the current study was designed to investigate knee and hip extension moments, and hamstrings and quadriceps activation in an isometric task involving whole-leg extensions in a functional range of leg positions. We hypothesized that ACL-deficient subjects would show a posture and load magnitude dependent increase in hamstrings co-activation.

Materials and methods

Subjects

Eleven chronic ACL-deficient subjects (four males, seven females) participated in this study. Ten ACL-deficient subjects had an ACL injury of the right knee and one had an ACL injury of the left knee. The interval between injury and testing ranged from 0.5 to 20 years (median 9.0 year). The rupture of the ACL was diagnosed by MRI scan, arthroscopy or clinical examination by independent orthopedic surgeons. The ACL-deficient subjects were asked to fill out the Lysholm score [30] and the International Knee Documentation Committee (IKDC) subjective questionnaire [17] to rate knee symptoms. The ACL-deficient subjects had a median score of 83 (range 63–92; a score of 100–95 means excellent and <65 means poor) on the Lysholm and 71 (range 52–91; a score of 100 means no limitation or symptoms) on the IKDC. None of the ACL-deficient subjects had knee pain or swelling at the time of testing. The control group consisted of 15 healthy subjects (10 males, 5 females) without a history of knee problems. In the control subjects, the right leg was tested. Subject characteristics did not differ significantly from the ACL-deficient group except for age (Table 1). Nevertheless, potential confounding factors were taken into account by normalizing variables to body proportions where appropriate and by taking gender into account in the statistical model.

All subjects signed a written informed consent form before the measurements. The Medical Ethics Review Committee of the VU University Medical Center approved the protocol.

Experimental set-up

The subjects were sitting on a custom-built seat (Fig. 1) supporting the back and tuber ischiadicum. Two supports over the shoulders prevented vertical movement of the trunk. The subjects received sensory feedback of the knee position through a rod at the back of the knee. They were not allowed to exert force against this rod during the measurements. When the subject pushed against the rod, the rod bent and the trial was repeated. The bare foot of the injured leg, or for the control subjects the bare foot of the right leg, was placed on a force plate. Isometric whole-leg extensions were performed with the knee in front of, over

Table 1 Relevant anthropometric characteristics of ACL-deficient and control subjects participating in the present study

	ACL-deficient (n = 11) Median	Minimum	Maximum	Control (n = 15) Median	Minimum	Maximum	P value
Age (years)	35.0	20.0	46.0	23.0	18.0	51.0	0.023
Body height (cm)	173.5	167.0	186.5	178.0	164.5	190.0	n.s.
Body weight (kg)	74.6	61.0	120.1	72.3	49.3	93.6	n.s.
Total leg length (cm)	90.2	79.0	94.8	91.0	85.3	97.5	n.s.
Upper leg length (cm)	41.2	33.0	45.5	39.7	37.1	43.4	n.s.
Lower leg length (cm)	50.0	44.7	52.3	49.9	41.2	54.8	n.s.
Foot length (cm)	25.0	23.0	29.2	27.0	23.0	30.0	n.s.

P values indicate significant differences between ACL-deficient and control subjects according to a non-parametric Kolmogorov–Smirnov test

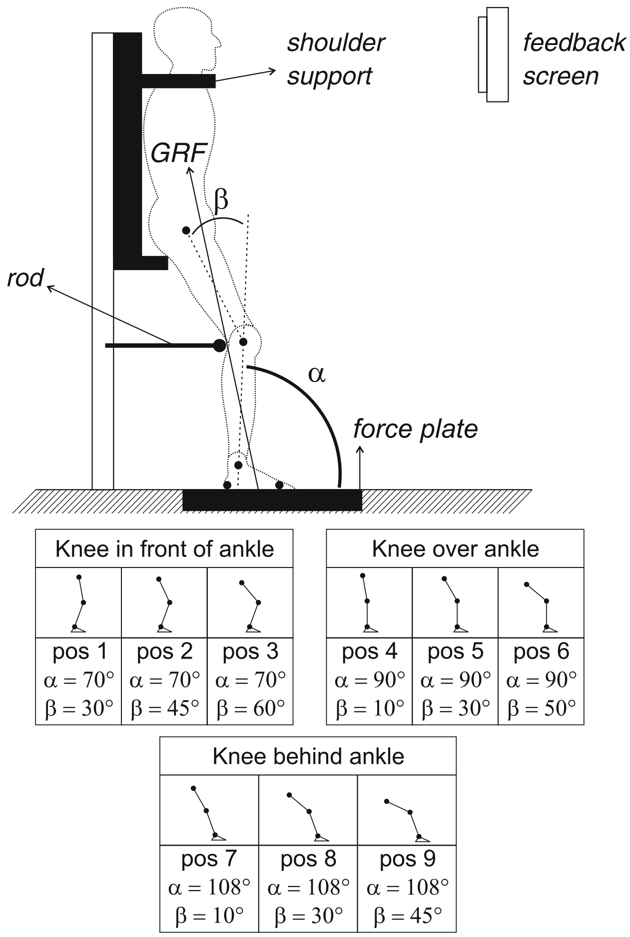

Fig. 1 Schematic overview of the experimental set-up. The subject was positioned in nine different postures. *Pos* posture, α ankle angle, β knee angle, *GRF* ground reaction force. *Dots* indicate the locations of the Optotrak markers

and behind the ankle joint, because the position of the knee relative to the ankle highly affects the required knee and hip moment during the whole-leg extensions. For each of those knee positions (in front of, over, and behind the ankle, with shank angles being 70°, 90° and 108° relative to the forward horizontal, respectively), three knee flexion angles were used, such that whole-leg extensions were performed in nine different postures (Fig. 1), thereby covering a wide and functional range of combinations of knee and ankle joint angle. Physical limitations of the experimental set-up prevented application of the same three knee flexion angles at each knee position. For instance, <30° of knee flexion could not be reached with the knee in front of the ankle. For each experimental position, a 10-s baseline measurement of the vertical force was performed with the subject resting on the chair with the foot on the force plate. Following the baseline measurement, the subject was asked to exert a maximum force by pushing against the force plate with the entire foot for 10 s. The other foot was not on the force plate and only used for balance control. After the maximum force trial, two target levels of 30 and 60% of the difference between maximum and baseline vertical force were calculated and only the vertical force component was displayed on a computer screen in front of the subject. The subject was then asked to go to the target vertical force level in approximately 2 s and maintain the force level for 8 s. During all force exertions, subjects pushed themselves against the shoulder support (which was padded to prevent discomfort) and against the back support. Fatigue was prevented by allowing about 5 min of rest between postures. The order of postures was randomized over subjects.

Measurements

Five LED markers were placed on the following locations: the trochanter major, the lateral femoral condyle, the lateral malleolus, lateral side of the calcaneus and on the base of the fifth metatarsal bone. An opto-electronic movement recording system (Optotrak, Northern Digital Inc., Waterloo, ON, Canada) was used to record the position of the markers during the measurements, at a rate of 100 samples/s. A Kistler force plate (Kistler Instrument Corp., Amherst, NY, USA) was used to measure the ground reaction force (GRF) in the vertical and anterior–posterior direction. The total GRF was calculated as the vector sum of vertical and horizontal components. The sample rate was set at 100 samples/s. A pulse generated by Optotrak started the force plate recordings. The Optotrak and force plate data were averaged over a 5-s isotonic part of each trial. An inverse dynamics approach was used to calculate knee and hip moments in the sagittal plane. Knee extension moments and hip flexion moments are expressed as positive values. Because moments scale with mass and length, moments were normalized to the product of subject's body mass and leg length (yielding units of m s^{-2}). The GRF was normalized to the subject's body mass (yielding units of m s^{-2}).

EMG was recorded from seven muscles (vastus medialis, VM; rectus femoris, RF; vastus lateralis, VL; semimembranosus, SM; semitendinosus, ST; biceps femoris, BF; gastrocnemius medialis, GM). The skin was shaved, abraded and cleaned before electrode placement (Ag/AgCl; square 5 mm × 5 mm pick-up area). The center-to-center electrode distance was 2.5 cm. Surface EMG locations were based on the Seniam guidelines [15]. The electrodes on the SM were placed on the distal part of the muscle, where the muscle is located just below the skin surface. All EMG signals were sampled at 1,000 samples/s (Porti-17™, TMS, Enschede, The Netherlands; 22 bits AD conversion after 20× amplification, input impedance $>10^{12}$ Ω, CMRR >90 dB) and band-pass filtered with 10 and 250 Hz cut-off frequencies. A pulse generated by Optotrak was used to synchronize EMG and Optotrak data. The EMG signals were band-stop filtered between 48.5 and 51.5 Hz, rectified and averaged over the same 5 s as the Optotrak and force plate data.

Prior to the measurements, subjects performed three maximum voluntary isometric contractions (MVIC) for both the quadriceps and the hamstrings at 90° of knee flexion. Another series of three MVIC was performed for the GM muscle at 90° of ankle flexion. The EMG signals measured during MVIC were band-pass filtered with 10 and 250 Hz cut-off frequencies, band-stop filtered between 48.5 and 51.5 Hz, rectified and averaged over 0.5 s sliding windows. For each muscle, the window with the highest value was used to normalize the EMG of that muscle. The normalized EMG signals of the VM, RF and VL were summed proportional to their cross-sectional areas [4], to obtain an indication of the overall quadriceps activation. Similarly, the SM, ST and BF were summed to obtain an indication of the overall hamstrings activation. A co-activation index (CI) was defined according to Kellis et al. [20]:

$$\mathrm{CI} = \frac{2 \cdot \mathrm{ham}}{\mathrm{qua} + \mathrm{ham}}$$

where, qua and ham are the quadriceps and hamstrings activation, respectively. An index of 1 means pure co-activation and 0 means no co-activation.

Statistics

Because of physical limitations, our experimental design did not allow to employ the same three knee angles in each knee joint position relative to the ankle joint. As a result, the experimental design was not balanced. Therefore, a separate repeated measure ANOVA was applied for each of the three knee joint positions. In each ANOVA, gender and ACL status (ACL-deficient vs. control group) were used as between-subject factors. Force level (30, 60% and maximum force) and knee angle (three angles) were used as within-subject factors. Significant interactions with ACL status were further explored with follow-up ANOVAs on subsets of the data. The variables tested were the normalized values (averaged over 5 s) of the magnitudes of backward GRF, the total GRF (vector sum of vertical and horizontal components), knee moment, hip moment, CI and the activation of the quadriceps, hamstrings and GM. For almost all of these variables, the data appeared to be positively skewed, due to the fact that some subjects pushed, in some conditions, in a direction that substantially deviated from the pushing direction generated by most other subjects. Therefore, we applied a log-transformation to all data prior to application of the ANOVAs.

Results

Muscle activation

Consistent with the hypothesis, ACL-deficient subjects showed adaptations in muscle activation that can be considered as a posture dependent, and to some extent load magnitude dependent, increase in co-activation (Figs. 2, 3; Table 2). In the posture requiring the smallest knee extension moments, i.e., with the knee behind the ankle, neither the muscle activation variables nor the moments showed differences between ACL-deficient and control subjects. In contrast, in the postures requiring larger knee extension moments (i.e., with the knee in front of and over the ankle, Fig. 4), a main effect of ACL status was found for hamstrings activation and interactions of ACL status with knee angle were found for hamstrings activation and CI (Table 2). In postures with the knee in front of the ankle, ACL-deficient subjects showed, averaged over three force levels and three knee angles, a median activation level of 6.6% MVC (range 3.2–12.3) hamstrings activation, against 4.2% (range 1.8–17.4) in control subjects (Fig. 2). In postures with the knee over the ankle those numbers were 9.1% MVC (range 2.0–29.0) for ACL-deficient and 4.0% MVC (range 2.2–35.7) for control subjects. The differences between ACL-deficient and control subjects (2.4% MVC for postures with the knee in front of the ankle and 5.1% MVC for postures with the knee over the ankle) were significant.

Furthermore, a main effect (thus independent of ACL status) of gender was found on hamstrings activation in that male subjects showed more hamstrings activation than female subjects in postures with the knee in front of and over the ankle.

The overall effect of ACL status on the CI did not reach significance. However, the ACL status interacted with knee angle for postures with the knee in front of and over the ankle in that the most pronounced difference between ACL-deficient and control subjects was seen in knee angles closer to full extension (Fig. 2). ACL status interacted with force level for postures with the knee in front of the ankle in that the most pronounced difference between ACL-deficient and control subjects was seen at higher force levels (Fig. 3).

The quadriceps activation did not show an overall difference between ACL-deficient and control subjects, but ACL status interacted with gender in postures with the knee in front of and over the ankle, and with force level in postures with the knee in front of or behind the ankle (Fig. 3; Table 2).

No main effect of ACL status or interaction with ACL status was found for the GM activation in any posture.

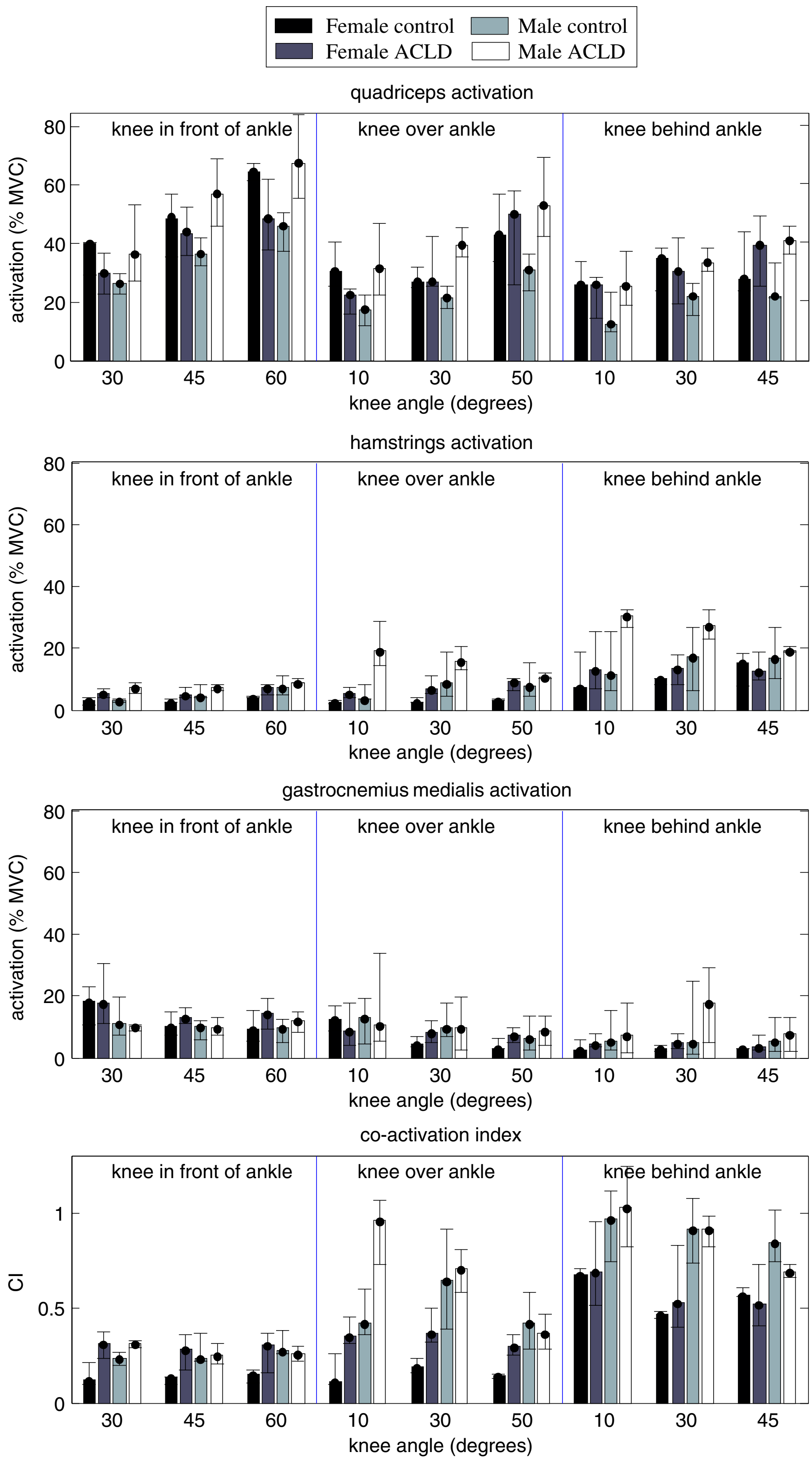

Fig. 2 Median activation levels of quadriceps, hamstrings and gastrocnemius muscles, and median co-activation indices for male (*light gray bars*, $n = 10$) and female (*black bars*, $n = 5$) control subjects and male (*white bars*, $n = 4$) and female (*dark gray bars*, $n = 7$) ACL-deficient subjects (ACLD). Data are presented for isometric whole-leg extension efforts in nine postures, varying in knee angle and knee joint position. Data are averaged over three force levels. *Error bars* represent the 25th and 75th percentiles. *Vertical lines* indicate separations between ANOVAs

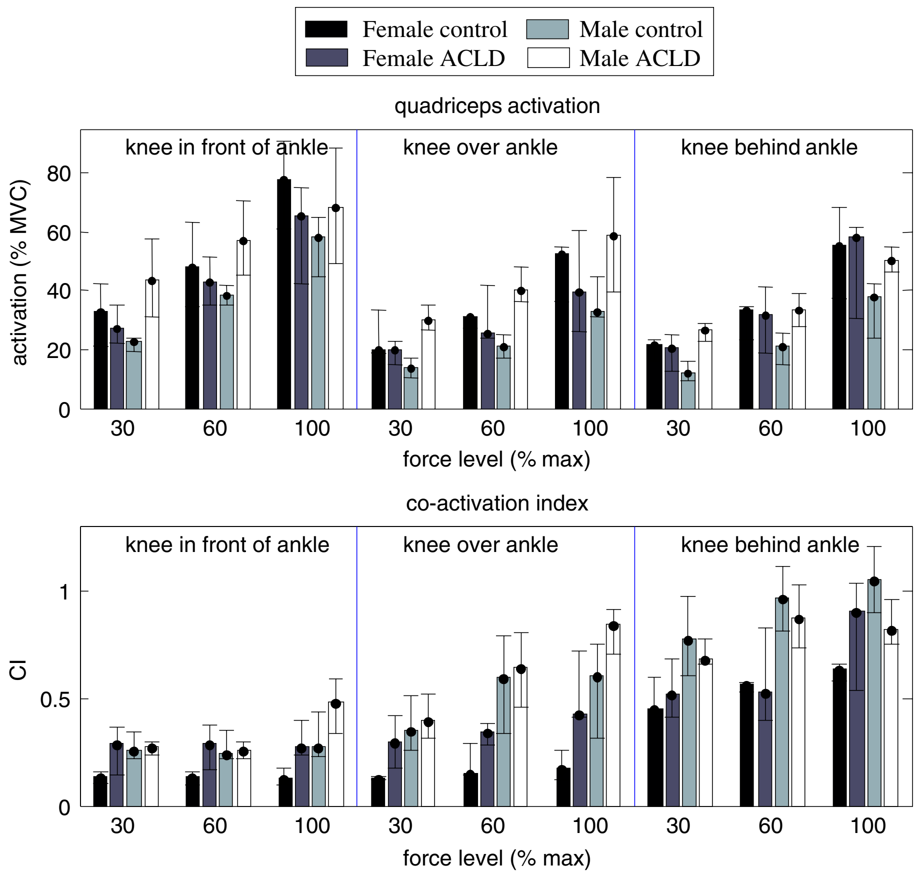

Fig. 3 Median activation levels of the quadriceps and median co-activation indices for male (*light gray bars*, $n = 10$) and female (*black bars*, $n = 5$) control subjects and male (*white bars*, $n = 4$) and female (*dark gray bars*, $n = 7$) ACL-deficient subjects (ACLD). Data are presented for isometric whole-leg extension efforts in nine postures, varying in force level and knee joint position. Data are averaged over three knee joint angles. *Error bars* represent the 25th and 75th percentiles. *Vertical lines* indicate separations between ANOVAs

Kinetics

Both subject groups closely matched the 30 and 60% target force levels. The absolute average difference between target and measured vertical GRF was not significantly different between ACL-deficient subjects (median 3.4 N, range 2.2–12.1) and control subjects (median 3.5 N, range 1.3–8.6).

For the normalized total GRF, main effects of ACL status were seen for postures with the knee in front of and over the ankle (Table 2). For the normalized backward GRF, an interaction of ACL status with force level was seen for postures with the knee over or behind the ankle, with more pronounced differences between groups at higher force levels (Fig. 5; Table 2). Furthermore, for both the normalized backward and normalized total GRF, ACL status interacted with gender for all postures in that reduced forces were seen in ACL-deficient males and to a lesser extent (backward GRF) or absent (total GRF) in ACL-deficient females.

With the knee in front of the ankle, ACL-deficient subjects generated smaller normalized knee moments (Table 2). Furthermore, ACL status interacted with gender and with force level for these postures. The interaction of ACL status with force level showed that the reduced knee moment in ACL-deficient subjects was most pronounced at the highest force level (Fig. 5). The main effect of ACL status and all interactions with ACL status were not significant for the moment at the hip joint (Fig. 4; Table 2).

Not surprisingly, the normalized knee moments, hip moments, total and backward GRF were, except for the total GRF with the knee over or behind the ankle, significantly affected by knee angle and force level in all knee joint positions (Figs. 4, 5).

Discussion

In line with the hypothesis, we found that ACL-deficient subjects showed a task dependent increase in co-activation. In postures requiring large knee extension moments, i.e., with the knee over or in front of the ankle, a consistent increase in hamstrings activation was found in ACL-deficient compared with control subjects. However, the difference between ACL-deficient and control subjects was only 2.4% MVC for postures with the knee in front of and 5.1% MVC for postures with the knee over the ankle. It can, therefore, be questioned whether this result, although statistically significant, is also clinically relevant. In this perspective, it is also important to recall that hamstrings co-activation was knee angle dependent for postures with the knee over or in front of the ankle.

Table 2 *P* values of the most relevant main effects and interactions of three separate repeated measures ANOVAs on the normalized magnitude of the total (nGRFtot) and backward (nGRFx) ground reaction force, on the extension moment at the knee joint (nMomK), on the net extension moment at the hip joint (nMomH), on the quadriceps activation (QUAD), on the hamstrings activation (HAM), on the gastrocnemius activation (GM) and on the co-activation index (CI)

	nGRFtot	nGRFx	nMomK	nMomH	QUAD	HAM	GM	CI
ANOVA knee in front of ankle (ankle angle: 70°; knee angles: 30°, 45° and 60°)								
ACL	0.003	n.s.	0.002	n.s.	n.s.	0.004	n.s.	n.s.
Gender	n.s.	0.033	n.s.	n.s.	n.s.	0.038	n.s.	n.s.
Gender × ACL	0.002	<0.001	0.006	n.s.	0.022	n.s.	n.s.	n.s.
Knee × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	0.030	n.s.	0.002
Force × ACL	<0.001	n.s.	0.001	n.s.	0.004	n.s.	n.s.	0.022
Knee × force × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Knee × gender × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Force × gender × ACL	0.005	n.s.	0.007	n.s.	n.s.	n.s.	n.s.	n.s.
ANOVA knee over ankle (ankle angle: 90°; knee angles: 10°, 30° and 50°)								
ACL	n.s.	n.s.	n.s.	n.s.	n.s.	0.002	n.s.	n.s.
Gender	n.s.	0.034	0.044	0.010	n.s.	0.003	n.s.	0.011
Gender × ACL	0.008	<0.001	n.s.	n.s.	0.016	n.s.	n.s.	n.s.
Knee × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	0.044	n.s.	0.003
Force × ACL	0.002	0.044	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Knee × force × ACL	n.s.	0.047	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Knee × gender × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	0.011	n.s.	0.024
Force × gender × ACL	0.007	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
ANOVA knee behind ankle (ankle angle: 108°; knee angles: 10°, 30° and 45°)								
ACL	0.017	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Gender	n.s.	0.002	0.027	n.s.	n.s.	n.s.	n.s.	0.011
Gender × ACL	0.020	<0.001	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Knee × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Force × ACL	n.s.	0.004	n.s.	n.s.	0.027	n.s.	n.s.	n.s.
Knee × force × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Knee × gender × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.
Force × gender × ACL	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.	n.s.

Non-significant *P* values ($\alpha > 0.05$) are omitted

Knee knee joint angle, *Force* force level, *ACL* ACL status

Enhanced hamstrings co-activation was most pronounced in small knee angles, whereas it has been shown that pure co-activation of the hamstrings and quadriceps cannot directly reduce anterior tibial shear forces in the knee at small (up to 22°) angles [7, 24]. Possibly, co-contraction is being generated in these postures to compensate for the reduced backward shear component of the hamstrings by increasing joint compression. In isolated static and slow dynamic knee extension efforts, enhanced hamstrings activation in knee angles close to the extension has been reported as well [1, 2], but differences between ACL-deficient and control subjects were found to be marginal [2] or absent [22]. In the present study, compensating a reduced knee extension moments by increasing hip or ankle moments were made possible while at the same time, posture was fully controlled. Nevertheless, while ACL-deficient subjects did show a force level and gender dependent decreased normalized backward GRF in postures with the knee in front of or behind the ankle, no evidence of compensatory changes in hip moments was found. In addition, the normalized backward GRF showed roughly the same pattern of difference between ACL-deficient and control subjects as the normalized total GRF and knee moments (see Fig. 5).

Our data showed that hamstrings activation was not just task dependent, but also gender dependent. Male subjects showed, for postures with the knee over or in front of the ankle, higher levels of hamstrings activation than females. This finding was consistent over ACL-deficient and control subjects, as no interaction with ACL status was found. Gender differences in co-activation, with males showing higher levels of hamstrings co-activation than females,

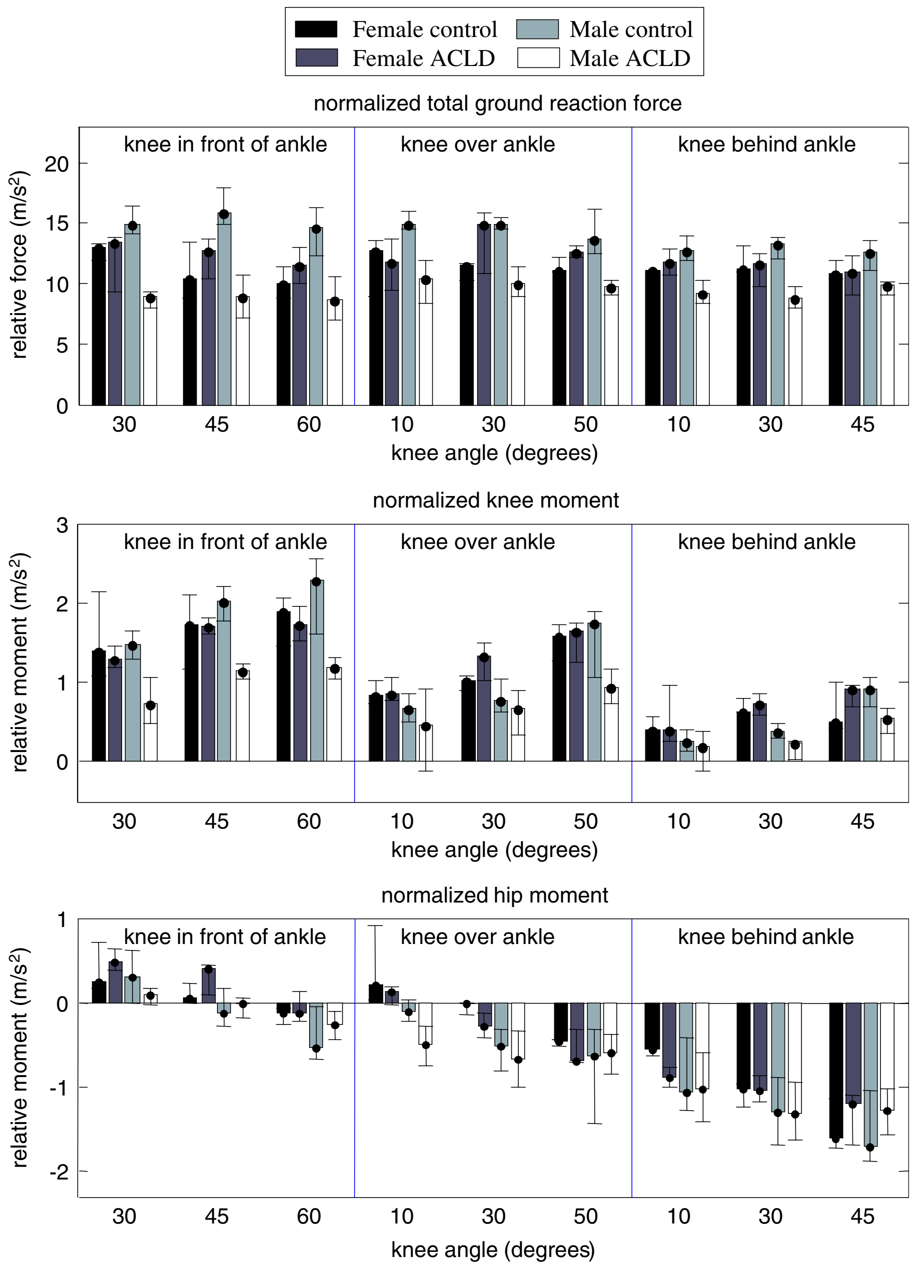

Fig. 4 Median magnitudes of the normalized total ground reaction force, normalized knee moment and normalized hip moment for male (*light gray bars*, $n = 10$) and female (*black bars*, $n = 5$) control subjects and male (*white bars*, $n = 4$) and female (*dark gray bars*, $n = 7$) ACL-deficient subjects (ACLD). Data are presented for isometric whole-leg extension efforts in nine postures, varying in knee angle and knee joint position. Data are averaged over three force levels. *Error bars* represent the 25th and 75th percentiles. *Vertical lines* indicate separations between ANOVAs

have been reported before in fast isokinetic knee extensions [16] and prior to landing in a drop-jump [23].

The quadriceps activation patterns found in the present study neither showed an overall nor a knee angle dependent adaptations in ACL-deficient subjects. Nevertheless, some interaction with force level was seen, indicating that ACL-deficient subjects may limit quadriceps activation at higher load levels. Furthermore, we did find an interaction between gender and ACL status in postures with the knee over or in front of the ankle. However, this interaction suggesting increased quadriceps activation in males, but not in females, should be interpreted with a great care. First, we measured only four male ACL-deficient subjects and some or all of those subjects may have been hesitant to fully activate the quadriceps in the MVC trial, which would have resulted in an overestimation of quadriceps activation. Furthermore, a comparison of quadriceps muscle activation between ACL-deficient and control subjects cannot directly be interpreted in terms of muscle forces as a smaller cross-sectional area of vasti muscles [33] and lower quadriceps strength [19] were found in groups of mainly male ACL-deficient subjects versus control subjects. In line, we found reduced knee moments in male subjects, mainly at higher load levels, which can probably only in part be attributed to enhanced hamstrings activation.

In addition to the hamstrings and quadriceps, the gastrocnemius might affect shear forces in the knee joint [12]. However, in the current study, the activation levels of the

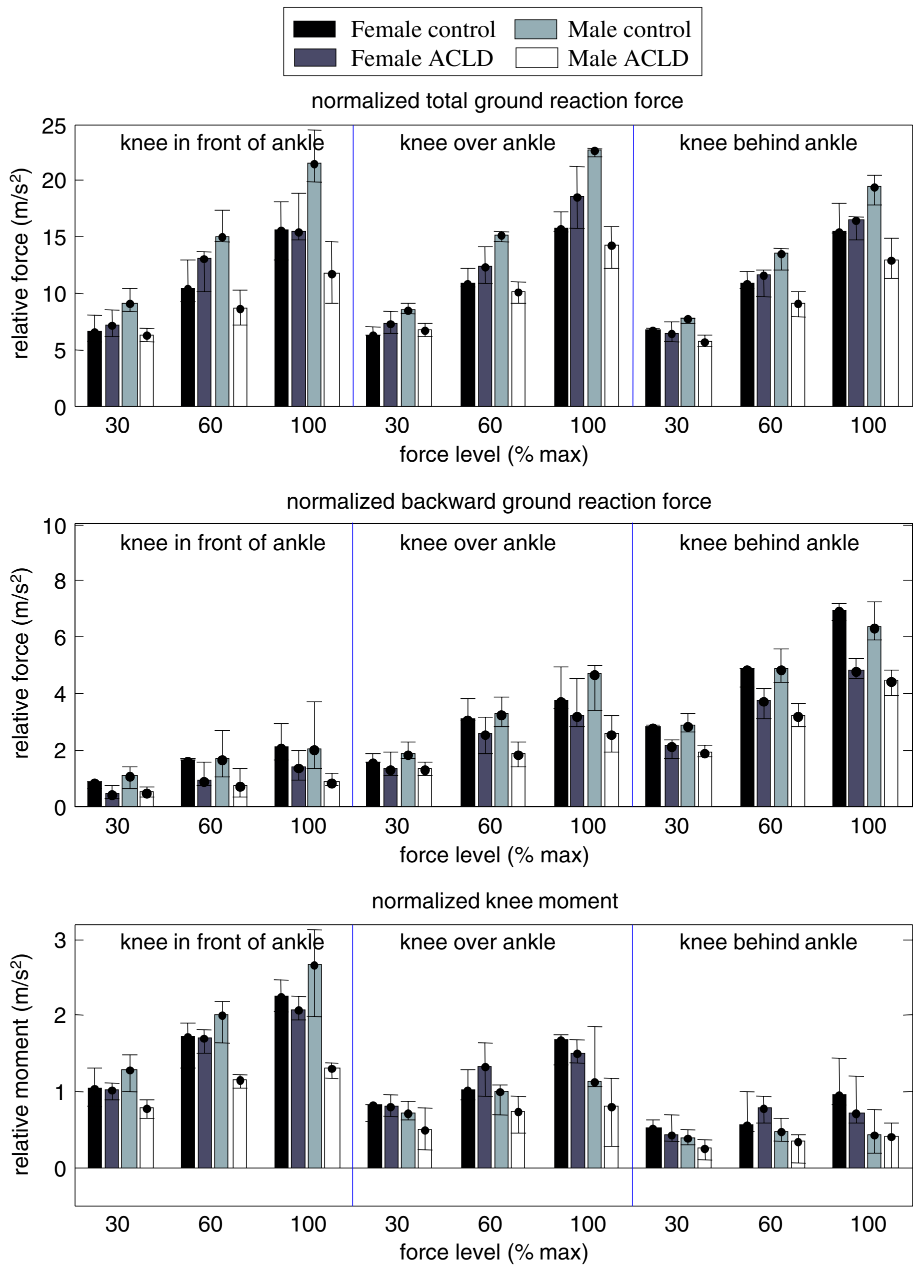

Fig. 5 Median magnitudes of the normalized total ground reaction force, normalized backward ground reaction force and normalized knee moment for male (*light gray bars*, $n = 10$) and female (*black bars*, $n = 5$) control subjects and male (*white bars*, $n = 4$) and female (*dark gray bars*, $n = 7$) ACL-deficient subjects (ACLD). Data are presented for isometric whole-leg extension efforts in nine postures, varying in force level and knee joint position. Data are averaged over three knee joint angles. *Error bars* represent the 25th and 75th percentiles. *Vertical lines* indicate separations between ANOVAs

gastrocnemius were not different between the subject groups.

A limitation of this study is that the level of functioning varied over ACL-deficient subjects. Four ACL-deficient subjects functioned on a high level (IKDC >85%) while four others functioned on a low level (IKDC <65%). Furthermore, the ACL-deficient subjects in the present study showed a large range in time from injury to test. This may have enhanced the variance in our data as Wexler et al. [32] showed that adaptations in ACL-deficient subject develop gradually over time. However, most of the subjects will have had sufficient time to adapt, as for seven of our subjects the injury was more than 7 years ago. Nevertheless, not all subject-related variance could be controlled, resulting in skewed data. By normalizing to subject proportions, by including gender as a between-subject factor and by log-transformation prior to ANOVA application, we minimized the risk of confounding effects. Unfortunately, the ACL-deficient group was too small to perform subgroup analyses. Future studies should focus on whether different strategies are used in groups differing in functional capacity. Finally, the task itself, matching target forces, could have had some effect on co-contraction level, but is unlikely to have affected differences between groups.

Conclusion

Anterior cruciate ligament-deficient subjects were found to use enhanced hamstrings co-activation without concomitant changes in quadriceps activation, and this was more pronounced at higher load levels and near full knee extension. However, the differences were small and it remains uncertain whether this is clinically relevant. Furthermore, female subjects had lower hamstrings activation levels than male subjects. Finally, male ACL-deficient subjects tended to limit quadriceps activation and resulting in knee moments at higher load levels.

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