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Influence of additional load on the moments of the agonist and antagonist muscle groups at the knee joint during closed chain exercise

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Abstract

The present study investigated the influence of additional loads on the knee net joint moment, flexor and extensor muscle group moments, and cocontraction index during a closed chain exercise. Loads of 8, 28, or 48 kg (i.e., respectively, $11.1 \pm 1.5\%$, $38.8 \pm 5.3\%$, and $66.4 \pm 9.0\%$ of body mass) were added to subjects during dynamic half squats. The flexor and extensor muscular moments and the amount of cocontraction were estimated at the knee joint using an EMG-and-optimization model that includes kinematics, ground reaction, and EMG measurements as inputs. In general, our results showed a significant influence of the *Load* factor on the net knee joint moment, the extensor muscular moment, and the flexor muscle group moment (all Anova $p < .05$). Hence we confirmed an increase in muscle moments with increasing load and moreover, we also showed an original *"more than proportional"* evolution of the flexor and extensor muscle group moments relative to the knee net joint moment. An influence of the *Phase* (i.e., descent *vs.* ascent) factor was also seen, revealing different activation strategies from the central nervous system depending on the mode of contraction of the agonist muscle group. The results of the present work could find applications in clinical fields, especially for rehabilitation protocols.

Keywords: Muscular cocontraction; Numerical optimization; Muscle group moments; Load; Closed chain exercises

1. Introduction

Rehabilitation programs employed for restoring the functional capacity of the joints after ligament injuries include either *open chain* exercises when the distal segment is free to move or *closed chain* exercises when the terminal segment is fixed (Escamilla et al., 1998). Relative to open chain efforts, closed chain exercises may be more adapted to rehabilitation programs, especially at the knee joint, because of minimal translation of the tibial plateau and lower forces experienced by the ligaments (Escamilla et al., 1998). For both types of movement, agonist–antagonist muscle cocontractions have been reported (Aagaard et al., 2000; Escamilla, 2001), suggesting contribution of muscular activity to active joint stabilization. Indeed, Basmajian and DeLuca (1985) and Stokes and Gardner-Morse (2003) indicated that co-activation of antagonist muscles about the joints participates in joint stability. Moreover, the cocontraction index (CI) has been reported to be a reliable variable to quantify the co-activation of agonist–antagonist muscle groups during multijoint dynamic exercises (see Kellis et al., 2003, for a comparison of the CI estimation methods).

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Considering open chain exercises and mainly isokinetic movements, the effects of fatigue, injuries, speed or load on co-activation of knee agonist–antagonist muscles have been extensively investigated (Aagaard et al., 2000; Aalbersberg et al., 2005; Kellis, 1998; Kellis and Baltzopoulos, 1998; Kellis and Kellis, 2001; Kingma et al., 2004). First, these studies reported either no evidence of a relationship between the amount of cocontraction and the knee anterior shear force (Aalbersberg et al., 2005; Kingma et al., 2004) or a possible positive correlation (Aagaard et al., 2000). Second, increasing the required agonist moment during open chain exercises (i.e., adding loads) leads to a higher EMG activity of the agonist muscles. The antagonist EMG activity also increases, but this raise is lower than that of the required agonist moment (i.e., a ''less than proportional trend'' for Kingma et al., 2004).

During closed chain exercises, previous works revealed that relative to healthy subjects, non-coper anterior cruciate ligament (ACL) deficient patients may use different activation patterns of the lower limb muscles to counteract the knee antero-posterior laxity (Alkjaer et al., 2002; Kingma et al., 2006; Rudolph et al., 2001). The use of external loads (e.g., additional weights) is frequent during rehabilitation programs, but few studies investigated the influence of load on the activity of the muscles surrounding the knee joint during a closed chain exercise. McCaw and Melrose (1999) reported an increase activity of the three superficial Quadriceps muscles with increasing load and no effect on the Biceps Femoris activity. These results would suggest a decrease in the amount of cocontraction at the knee joint as load increases.

The methods employed in the above studies on agonist–antagonist co-activation focused on EMG data to study cocontraction and, as noticed by Kellis (1998), EMG data alone could lead to misinterpretations because of normalization issues, possible distortions of the signal (Rainoldi et al., 2000) and the influence of joint kinematics in dynamic conditions (Potvin, 1997). Alternatively, the use of an EMG-and-optimization model may overcome the previous limitations and provide a convenient procedure to obtain the cocontraction index from reliable estimates of the contribution of the agonist and antagonist muscle groups to the net joint moment (Amarantini and Martin, 2004; Doorenbosch and Harlaar, 2003; Kellis and Baltzopoulos, 1997; Kellis et al., 2003).

The present work investigated the influence of load on the knee flexor and extensor muscle group moments during dynamic squats, i.e., a closed chain exercise of lower body. Three levels of external load, corresponding to what is usually encountered in rehabilitation protocols, were applied to the subjects. An updated version of the EMG-to-moment optimization process developed by Amarantini and Martin (2004) provided estimates of the knee agonist and antagonist muscle group moments further used to compute CI. Our working hypothesis is that during closed chain exercises, the CI would increase with load to actively stabilize the knee joint. We also hypothesized differences in the CI as well as in flexor and extensor muscle group moments depending on the squat cycle phases because each muscle group may have a different role depending on the mode of contraction of the agonist muscle group (i.e., eccentric or concentric).

2. Methods

2.1. Subjects

Eight male students at the Sport Sciences Faculty of Marseilles, novice in weight lifting and free of knee-injury histories participated in this study. Mean (±s.d.) age, height and mass were 20.1 ± 2.8 years, 177.0 ± 3.2 cm and 75.1 ± 11.8 kg, respectively. The project was approved by the University Review Board and all participants gave informed consent in accordance with the Helsinky convention.

2.2. Instrumentation

A six cameras Vicon624 system (Vicon Motion System, Lake Forest, CA) operating at 120 Hz recorded kinematics data from eight markers attached to the fifth metatarsal head, the lateral malleolus, the lateral femoral condyle, the major trochanter, the head of the clavicle, the jaw, the vertex of the head, and the center of the barbell. Cartesian coordinates were smoothed with a cubic spline smoothing procedure (Matlab Spline Toolbox, version 3.2.2).

Ground reactions were sampled at 240 Hz from a forceplate (AMTI, Model LG6-4-CE, Watertown, USA). Raw dynamic data were low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 10 Hz cutoff frequency.

Electromyographic data were recorded at 1000 Hz (Mega ME 3000 P8, Mega Electronics Ltd.; gain = 412, CMRR = 110 dB) using Ag/Ag-Cl bipolar surface electrodes (Skintact model FS 501, Innsbruck, Austria) placed over the bellies of gastrocnemius medialis (ga), biceps femoris (bf), rectus femoris (rf), and vastus medialis (vm) right leg muscles with a 2 cm center-to-center inter-electrodes distance. These muscles were chosen according to Amarantini and Martin (2004) and Olney and Winter (1985) and included mono-articular (vm) as well as bi-articular muscles spanning both the knee joint as well as the hip (bf and rf) and ankle (ga) joints.

2.3. Instructions

Subjects stood with both feet on the forceplate and performed dynamic half squats, beginning in an upright position, bending the knees until thighs were parallel to the floor and returning to the initial posture. After an active warm-up, the subjects performed seven consecutive cycles at self-selected speed randomly loaded by a barbell of 8 (barbell only), 28, or 48 kg with 4 min rest between each condition. The barbell was positioned across the back of the shoulders. These loads corresponded, respectively, to $11.1 \pm 1.5\%$, $38.8 \pm 5.3\%$, and $66.4 \pm 9.0\%$ of body mass. Numerous studies on closed chain exercises have used either loads corresponding to a percentage of one repetition maximum (Ebben and Jensen, 2002; Escamilla et al., 1998; Escamilla, 2001) or no load (Isear et al., 1997). As the results of the present study may find direct applications for rehabilitation, constant loads were used considering first that it may be hazardous to perform a maximum effort just after surgery and second, that rehabilitation equipments usually offer fixed increasing loads.

2.4. Data processing and modeling

Angular displacements of foot, shank, thigh and trunk segments were computed from smoothed Cartesian coordinates and interpolated to 1000 Hz using third-order splines. Joint angular velocities and accelerations were obtained by analytical differentiation. For each subject and load, each cycle was normalized in time from 0% to 100% of the squat cycle duration. The net knee flexion/extension moment was computed for each time instant t by solving inverse dynamics for a planar four bar-linkage system where ankle, knee, hip, and shoulder joints were considered as frictionless hinges (see Cahouët et al., 2002, for the generalized form of equations). Bilateral symmetry was assumed (Escamilla et al., 1998) and net joint moments were computed using body segment parameters data (Zatsiorsky and Seluyanov, 1983). To estimate the contribution of a single leg, the net joint moment was divided by half.

The first and the last cycles were removed from the analysis and one representative cycle per subject and condition was obtained by averaging the kinematics, net joint moment, and EMG data of the five remaining consecutive cycles. The selected number of cycles was sufficient to obtain reliable EMG data (Arsenault et al., 1986).

The estimates of the knee net joint moment, flexor and extensor moments were obtained from an updated version of the Amarantini and Martin (2004) model. The major update consisted of the incorporation of the isometric EMG-to-moment calibration directly into the dynamic procedure by assuming a linear trend between the muscle group moments and the rectified and filtered EMGs (Amarantini and Martin, 2004). Hence, the coefficients for the isometric EMG-to-moment relationship (α_i in Eq. (2)) were estimated during the dynamic session (Centomo et al., 2007a,b). In the model, the force production capacity of each muscle group is attributed to the selected muscles of the corresponding muscle group.

Thus, the flexor and extensor muscle group moments were estimated at the knee joint by solving the following optimization problem:

Find:

$$\alpha_i = \{\alpha_{\mathrm{ga}}, \alpha_{\mathrm{bf}}, \alpha_{\mathrm{rf}}, \alpha_{\mathrm{vm}}\},\ w_i(t) = \{w_{\mathrm{ga}}(t), w_{\mathrm{bf}}(t), w_{\mathrm{rf}}(t), w_{\mathrm{vm}}(t)\},$$
$$\beta_j = \{\beta_{\mathrm{a}}, \beta_{\mathrm{k}}, \beta_{\mathrm{h}}\} \text{ and } \delta_j = \{\delta_{\mathrm{a}}, \delta_{\mathrm{k}}, \delta_{\mathrm{h}}\}$$

$$\text{that minimize}: C = \frac{1}{2}\cdot\sum_t (M_{\mathrm{K}}(t) - \widehat{M}_{\mathrm{K}}(t))^2 \tag{1}$$

$$\text{with}: \widehat{M}_{\mathrm{K}}(t) = \sum_i (\alpha_i \cdot w_i(t) \cdot S_i(t))^{\mathrm{T}} \cdot [1 + E\cdot(\beta_j \cdot \Delta\theta_j) - E\cdot(\delta_j\cdot\dot{\theta}_j)] \tag{2}$$

$$i = \{\mathrm{ga}, \mathrm{bf}, \mathrm{rf}, \mathrm{vm}\} \text{ and } j = \{\mathrm{a}, \mathrm{k}, \mathrm{h}\}$$

where a refers to the ankle joint, k to the knee, and h to the hip.

$$\text{subject to}: \begin{cases} \alpha_{\mathrm{ga}}, \alpha_{\mathrm{bf}} < 0;\ \alpha_{\mathrm{rf}}, \alpha_{\mathrm{vm}} > 0 \\ \beta_j \text{ and } \delta_j > 0 \\ 0 < w_i(t) < 1 \\ \widehat{M}_{\mathrm{ga}}, \widehat{M}_{\mathrm{bf}} < 0; \widehat{M}_{\mathrm{rf}}, \widehat{M}_{\mathrm{vm}} > 0 \end{cases} \quad \text{as inequality constraints} \tag{3}$$

In (1), $M_{\mathrm{K}}(t)$ represents the knee net joint moment computed by inverse dynamics while $\widehat{M}_{\mathrm{K}}(t)$ is the net joint moment estimated through the optimization procedure using the EMG data as input. In Eq. (2), $M_{\mathrm{Flex}}(t)$ corresponds to the sum at each time t of $\widehat{M}_{\mathrm{ga}}$ and $\widehat{M}_{\mathrm{bf}}$ while $M_{\mathrm{Ext}}(t)$ results from the sum of $\widehat{M}_{\mathrm{rf}}$ and $\widehat{M}_{\mathrm{vm}}$. $S_i(t)$ contains the full-wave rectified and filtered (fourth-order, zero-lag Butterworth, 2.5 Hz cutoff frequency) EMG data of the four selected muscles. α_i is the matrix establishing the isometric EMG-moment relationship and $w_i(t)$ represents the matrix of individual muscle gains. This matrix defines the contribution of each muscle to the corresponding muscle group moment. The mathematical expression of $\widehat{M}_{\mathrm{K}}(t)$ also includes matrices of biarticularity, stiffness, and viscosity (respectively, E, β_j and δ_j) to take into account the force–length and force–velocity relationships (please refer to Amarantini and Martin, 2004 for more details). According to van Dieen and Visser (1999), $\Delta\theta_j(t)$ was computed relative to the mean value of the angular range covered by the ankle, knee, and hip joints during squats.

For each subject, the optimization procedure was completed after concatenating the averaged cycles of the three conditions of load in a single row to obtain constant isometric EMG-moment coefficients (α_i). Indeed, as a result of the design of the experiment, this subject specific physiologic coefficient (i.e., the EMG-to-moment isometric coefficient) should not vary during the experiment. This non-linear constrained optimization problem was solved using a Sequential Quadratic Programming (Boggs and Tolle, 1995) (Matlab Optimization Toolbox, version 3.0.3).

CI was computed at each time instant t using the expression given in Falconer and Winter (1985):

$$\mathrm{CI}(t) = \frac{2\cdot|M_{\mathrm{Antago}}(t)|}{|M_{\mathrm{Ago}}(t)| + |M_{\mathrm{Antago}}(t)|} \tag{4}$$

where $M_{\mathrm{Antago}}(t)$ and $M_{\mathrm{Ago}}(t)$ correspond to $M_{\mathrm{Flex}}(t)$ or $M_{\mathrm{Ext}}(t)$ relative to the sign of the net joint moment ($\widehat{M}_{\mathrm{K}}(t)$).

2.5. Statistics

Based upon the sign of the vertical velocity of the barbell, the minimum, mean, and peak values of knee angular velocity, net joint moment, flexor and extensor moments, and CI were computed during the descent and the ascent phases of the squat cycle. Two factors (*Load* and *Phase*) ANOVAs with repeated measures on both factors were conducted on each dependent variable. Prior to the test, each variable was normalized by the mean absolute value of the barbell vertical velocity to counterbalance the influence of movement velocity because loaded squats were performed at self-selected speed. Moreover, dependent variables were also normalized by subject's body weight.

Follow-up analyses were conducted using two sets of planned comparisons between the levels of *Load* and *Phase* because we were interested in comparisons between specific conditions rather than an overall condition effect. A first planned comparison examined whether each modality of *Load* was affected by a *Phase* effect. A second set of planned comparisons examined the influence of *Load* within each phase. A significance level of .05 was used for all comparisons.

3. Results

As expected from the design of the objective function in the optimization process, the EMG-and-optimization model provided estimates of the knee net joint moment ($\widehat{M}_{\mathrm{K}}(t)$) with a very high correspondence to $M_{\mathrm{K}}(t)$ and a coefficient of determination between both time series close to 1.0.

3.1. Load effect

The three levels of load supported by the subjects corresponded to significant different percentages of their body mass ($F_{2,42} = 1207.48$; $p < .05$). Statistical analyses showed high consistency in subject behaviors with no influence of *Load* neither on minimum, maximum, nor mean values of the knee angular velocity (respectively: $F_{2,14} = 0.37$; $p > .05$; $F_{2,14} = 0.62$; $p > .05$; $F_{2,14} = 0.76$; $p > .05$).

The presence of higher loads significantly increased the knee net joint moment (Table 1, Fig. 1). The mean values of $\widehat{M}_K(t)$ computed over the entire squat cycle were linearly related to *Load*, with r^2 values close to 1.0 (Fig. 2). Indeed, mean $\widehat{M}_K(t)$ values increased by 7.8% between the 8 and 28 kg conditions and by 8.3% between the 28 and 48 kg conditions. Peak values of the knee net joint moment were also significantly affected by the *Load* factor ($F_{2,42} = 3.28$; $p < .05$). Especially during the descent phase, the peaks of $\widehat{M}_K(t)$ for the condition 48 kg were higher than those obtained for the condition 8 kg. This significant influence of *Load* was also shown ($F_{2,42} = 15.63$; $p < .05$) on the minimum values of the knee net joint moment (Table 1). Indeed, whatever the phase, minimum values of $\widehat{M}_K(t)$ for the condition 8 kg were higher than those of the conditions 28 and 48 kg. Moreover, the results for the condition 28 kg were higher than those obtained when carrying 48 kg.

The extensor and flexor muscular moments estimated through the EMG-to-moment optimization procedure were also dependent on the *Load* factor (Table 1, Figs. 1 and 2). The normalized maximum values of the extensor muscular moments for the condition 48 kg were higher ($F_{2,42} = 4.19$; $p < .05$) than those of the condition 8 kg (Table 1 and Fig. 1). Regarding the normalized mean values of the flexor moment, conditions 8 and 28 kg were similar (overall averages: -21.5 ± 10.3 a.u. and -22.6 ± 7.6 a.u., respectively, for 8 and 28 kg) while the condition 48 kg produced lower values (-32.1 ± 9.3 a.u.; $F_{2,42} = 3.96$; $p < .05$). Similar results were found with the minimum peaks of the flexor moment as values of the 48 kg condition (-56.1 ± 19.9 a.u.) were statistically lower than those of the 8 and 28 kg conditions (-36.3 ± 16.9 a.u. and -34.9 ± 10.4 a.u. for 8 and 28 kg, respectively; $F_{2,42} = 7.06$; $p < .05$). Mean values of the extensor moments increased by 6.0% between the 8 and 28kg conditions while these means raised by 16.6% between the 28 and 48 kg conditions. A similar trend was seen for the flexor moment mean values with a low (1.9%) decrease between 8 and 28 kg conditions and a sharp drop (33.2%) between the 28 and 48 kg conditions.

Opposite to these results, no main *Load* effect was seen neither on the minimum, mean, nor peak values of CI. However, a clear trend was seen for the knee angles at the time instants of maximal values of cocontraction index (Fig. 3). Indeed, these angles shifted from $24.46 \pm 9.42°$ for 8 kg, to $29.56 \pm 10.58°$ for 28 kg up to $33.17 \pm 14.13°$ for 48 kg.

3.2. Phase effect

The mode of contraction of the agonist muscle group was considered as eccentric during the descent phase and concentric during the ascent phase. Our results showed a significant influence of the *Phase* factor with higher absolute mean values of the flexor muscle group moments during the ascent phase than during the descent (Table 1). Hence, the amount of antagonist (flexor muscle group) moment depends on the type of muscular action of the agonist muscle group with higher levels of antagonist moment while the agonist group act concentrically. Statistical analyses revealed no influence of *Phase* neither on the minimum, maximum, nor mean values of the knee net joint moment and extensor muscle group moments. Regarding the CI, the influence of *Phase* was also not significant whatever the variable studied. On the contrary, a significant influence ($F_{1,42} = 5.68$; $p < .05$) was seen for the absolute mean values of the flexor moment with descent phase (22.97 ± 9.83 a.u.) lower than the ascent (27.80 ± 8.20 a.u.).

Table 1
Averaged ± s.d. values ($n = 8$) of the parameters investigated in this study (minimal, mean and maximal values)

Load (kg)	Phase					
	Descent			Ascent		
	8	28	48	8	28	48
CI_{mean}	48.5 ± 14.3	44.7 ± 14.2	45.2 ± 9.6	52.5 ± 10.1	53.6 ± 10.8	54.2 ± 11.5
CI_{max}	90.9 ± 14.3	93.7 ± 17.2	98.5 ± 3.8	91.9 ± 9.4	98.4 ± 3.4	99.9 ± 0.1
Net^{L}_{min}	$2.6 \pm 11.9^{\dagger}$	$-9.1 \pm 14.1^{\infty}$	$-30.7 \pm 25.4^{\dagger\infty}$	1.2 ± 11.1	$-12.1 \pm 12.6^{\infty}$	$-27.0 \pm 16.8^{\infty}$
Net_{mean}	52.3 ± 16.9	61.3 ± 20.8	69.6 ± 24.3	46.7 ± 14.5	46.7 ± 18.1	47.3 ± 21.9
Net^{L}_{max}	$112.2 \pm 27.7^{\dagger}$	132.9 ± 31.2	$149.3 \pm 39.7^{\dagger}$	103.1 ± 28.7	120.6 ± 38.6	133.2 ± 49.8
Ext_{mean}	74.6 ± 21.1	81.2 ± 19.5	96.2 ± 23.6	67.5 ± 13.5	72.0 ± 18.8	83.7 ± 17.5
Ext^{L}_{max}	$130.7 \pm 26.7^{\dagger}$	153.7 ± 31.3	$173.7 \pm 37.5^{\dagger}$	122.4 ± 25.6	141.6 ± 38.6	161.6 ± 45.9
$Flex^{L,P}_{mean}$	-22.3 ± 13.5	-19.9 ± 7.5	$-26.7 \pm 8.5^{*}$	$-20.7 \pm 7.0^{\dagger}$	$-25.3 \pm 7.6^{\infty}$	$-37.4 \pm 10.0^{\dagger\infty*}$
$Flex^{L}_{min}$	-37.3 ± 24.7	$-32.6 \pm 8.7^{\infty}$	$-53.4 \pm 21.9^{\infty}$	$-35.2 \pm 9.0^{\dagger}$	$-37.1 \pm 11.9^{\infty}$	$-58.8 \pm 17.8^{\dagger\infty}$

CI corresponds to cocontraction index, Net stands for knee net joint moment, Ext represents the extensor muscular moment, and Flex is the moment of the flexor muscle group. P and L superscripts indicate significant *Phase* and *Load* main effects, respectively. Significant results of planned comparisons are indicated in each cell by *, †, and ∞: * reveals a *Phase* effect for the associated load. Within each phase, † indicates differences between loads 8 and 48 kg, and ∞ indicates differences between loads 28 and 48 kg.

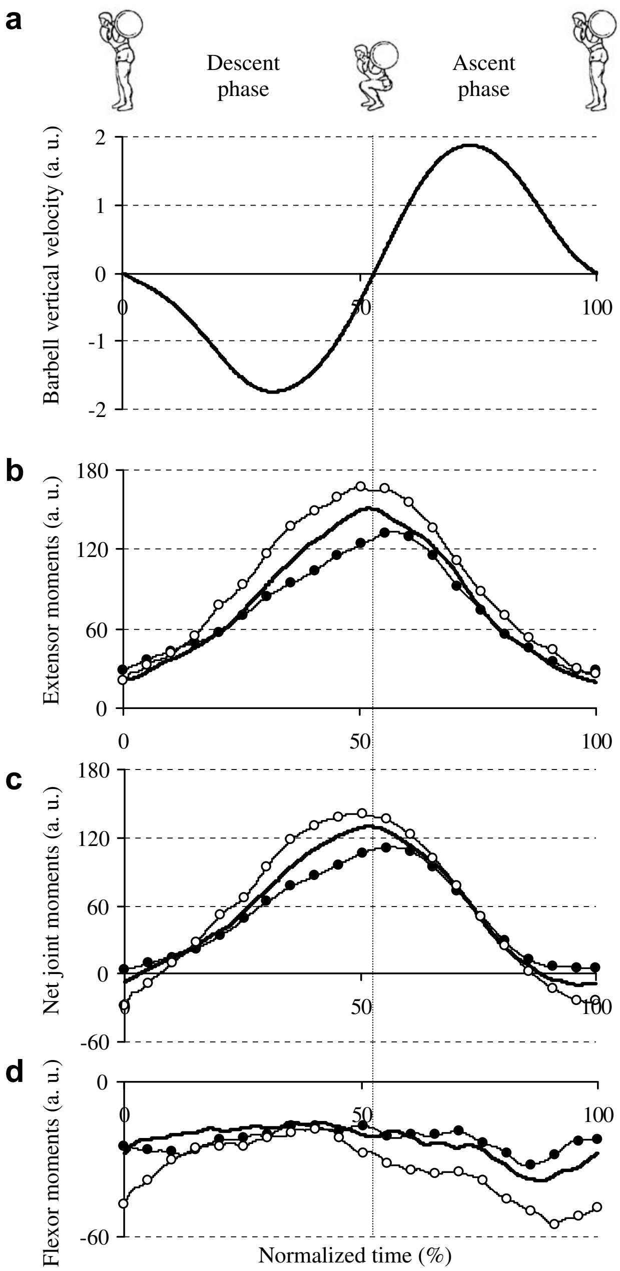

Fig. 1. From top to bottom: (a) Normalized values of the barbell vertical velocity averaged over the three conditions of load, (b) normalized knee extensor muscle group moments, (c) knee net joint moments, and (d) the normalized flexor muscular moments. For each graph, black dots correspond to the 8 kg load, thick black line to 28 kg, and open white dots to 48 kg. All these variables are averaged across subjects ($n = 8$). Please note the influence of *Phase* (descent *vs.* ascent) on the mean values of the flexor muscle group moments.

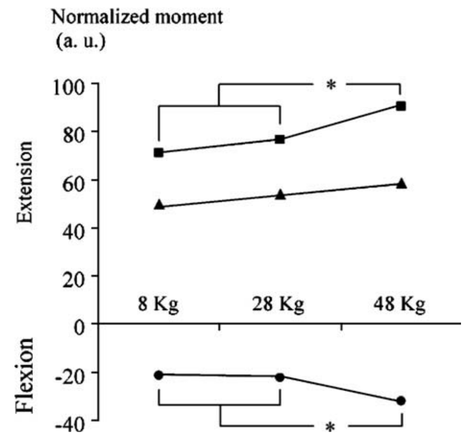

Fig. 2. Means ($n = 8$) of the normalized net (black triangles), extensor (black squares), and flexor (black dots) moments as a function of the barbell weight. Values were averaged over the squat cycle time. * indicates significant differences ($p < .05$).

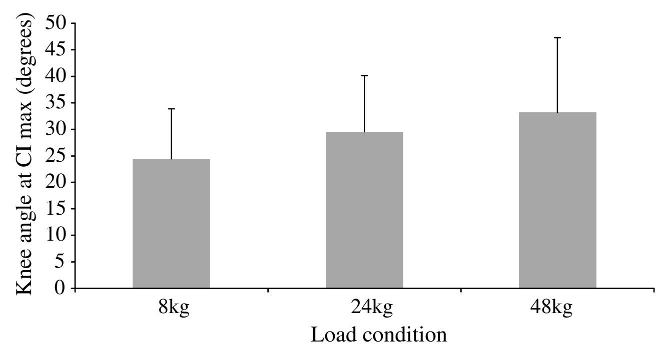

Fig. 3. Means (and s.d., $n = 8$) of the knee angles at the time instant of maximal cocontraction as a function of the barbell weight. Please note the general increase in knee angle as the carried load increases.

4. Discussion

Relative to open chain efforts, closed chain exercises are widely used in rehabilitation protocols following knee ACL injuries because first, the functional restoration is more complete and second, the shear stress on the joint is reduced (Ben Kibler and Livingston, 2001). Numerous studies revealed that cocontraction of the agonist and antagonist muscle groups surrounding the knee may help in stabilizing the joint. The higher stability resulted from an increased axial compression force due to the antagonist coactivity (Stokes and Gardner-Morse, 2003). However, despite the use of different weights during closed chain exercises for rehabilitation purposes, few studies investigated the influence of load either on the activity of the agonist and antagonist muscle groups, or on the cocontraction level. Hence, the present study analyzed the influence of *Load* on the flexor and extensor muscular moments and on the CI at the knee joint during dynamic half-squat movements. Considering the uncertainties associated to the use of the EMG data solely, the flexor and extensor muscle group moments were evaluated using a reliable EMG-to-moment optimization process. Previous studies reported an increased hamstrings activity during the ascent phase (Escamilla et al., 1998; Isear et al., 1997; McCaw and Melrose, 1999). These authors associated this higher activity of the hamstrings bi-articular muscles to an increase in the hip extensor moment to be generated. While this explanation is certainly convincing, the use of the hamstrings muscles solely is not sufficient to estimate the overall flexor activity at the knee joint. Because the gastrocnemius muscle is bi-articular, crossing both the knee and the ankle joints, and shows significant activity during squat movements (Escamilla et al., 1998), its influence on the knee joint flexor activity has to be taken into account. The present EMG-to-moment model included EMG data from the gastrocnemius muscle, thus enhancing the accuracy of the estimation of the knee flexor muscle group moment.

As expected, a constant increase in *Load* (20 kg between each condition) led to a linear evolution of the mean knee net joint moments (+7.8% and +8.3% between the 8 and 28 kg, and the 28 and 48 kg, respectively). This linear trend is confirmed by a coefficient of determination close to 1.0 between the mean knee net joint moment and the *Load*. Our results also showed an increase in the net knee joint moments as well as in the flexor and extensor muscle group moments with *Load* as most of the dependent variables for the 48 kg condition are higher than for the 8 kg situation (Table 1). As McCaw and Melrose (1999), our data showed a significant influence of *Load* on knee joint flexor and extensor muscle groups moments. Indeed, the peaks of the extensor and flexor muscle moments increased with *Load* (see Fig. 1 and Table 1). Mean values of flexor muscle group moments were also affected by *Load,* as absolute values of the conditions 8 kg and 28 kg were inferior to those observed for 48 kg (see Fig. 2). These results reflected the higher demand placed on both muscle groups to achieve the required task, especially when supporting a 48 kg load.

During closed chain exercises, our results provide new insights into the mechanisms underlying changes in muscle group moments. Our findings introduce the new idea that different loads may have different roles during rehabilitation exercises. Opposite to Kingma et al. (2004) who reported a "less than proportional" evolution of the agonist and antagonist EMG activities relative to the net joint moment during an open chain task, our results showed changes in $M_{\mathrm{Ext}}(t)$ and $M_{\mathrm{Flex}}(t)$ "more than proportional" as *Load* increased (Fig. 2). Statistical analysis of mean antagonist muscle moments showed an initial plateau with no difference between weaker loads (1.9% between 8 and 28 kg), followed by a sharp increase for the higher load (33.2% between 28 and 48 kg). This specific trend could reflect a threshold in the moments developed by the antagonist muscle group under different conditions of loading. For the 48 kg load, the significant increase in the moment developed by the antagonist muscle group may increase joint stability (Granata and Marras, 2000) and actively protect the knee ACL (Aagaard et al., 2000). Rehabilitation protocols could take advantage of this feature by proposing closed chain exercises with low loads to mobilize the knee and exercises with higher loads to train the antagonist muscle group to stabilize actively the joints. Considering the differences in the slopes of the antagonist activities between open and closed chain exercises (i.e., respectively, less and more than proportional relative to the net joint moment), our results would also suggest that fundamental findings from studies focusing on open chain exercises may not be fully applied to rehabilitation protocols.

In our study, no difference was seen in the amount of cocontraction despite modifications of both the extensor and flexor muscle group moments. This result may be explained by the equation that governed the estimation of the cocontraction index (Eq. (4)). Indeed, similar increases in the agonist and antagonist muscle group moments would lead to similar amount of cocontraction. This result emphasizes the fact that investigating the agonist and antagonist activities by themselves is complementary to the analysis of the cocontraction index around a joint. However, a clear trend was seen for the values of the knee angles corresponding to the instants of maximal cocontraction depending on the *Load* (Fig. 3). These knee angles where CI was maximal were 24° for 8 kg, 30° for 28 kg and 33° for 48 kg. Kingma et al. (2004) reported that below a 15–22° knee flexion angle, the hamstrings muscles are ineffective to counteract the anterior shear force produced by the quadriceps. Our results tend to show that with higher loads, requiring a higher stability of the knee joint, the peak of cocontraction is shifted to an angular range where the efficiency of the hamstrings muscles to actively stabilize the knee is maximal. These results confirm the advantageous role of cocontraction because the contribution of active stiffness to joint stability depends both on the knee angle and on the external load. These findings also suggest the ability of the neuromuscular system to appropriately activate the opposing muscle groups to assist the passive elements of stability crossing the knee joint.

Our results provide insights on the specific role of antagonist muscle group during closed chain exercises as the amount of antagonist moment depends on the type of muscular action of the agonist group (i.e., concentric/eccentric). It was previously shown that the control of movements involving eccentric contraction requires a single muscle group and conversely that a concentric activation alone may not result in coordinated motions (Enoka, 1996). Our data showed that the mean flexor muscle group moment is lower during the descent phase (while the knee extensor muscles act eccentrically) than during the ascent phase (i.e., when the extensor muscles contract concentrically). Besides an increased joint stability (Granata and Marras, 2000), modifications in the amount of antagonist cocontraction have been reported depending on the level of expertise (Osu et al., 2002), the required precision of a task (Gribble et al., 2003), or the presence of interaction torques (Gribble and Ostry, 1999). From the design of the present experiment that implied novices, low precision task and little amount of interaction torques, it is likely that most of the antagonist coactivity will be related to a demand of a higher joint stability. Hence during the descent phase, the knee extensor muscles would produce force and control movement velocity while the knee flexors act as joint stabilizers (i.e., by increasing the axial compression force). During the ascent phase and considering the conclusion of Enoka (1996), the extensor muscles would generate motion while the flexor muscles would both stabilize knee joint and regulate movement speed.

To conclude, our results emphasize the influence of *Load* and *Phase* on the knee flexor and extensor muscle group moments during closed chain exercises. Despite no modification in the amount of cocontraction with the increasing *Load*, our results showed that changes in the agonist and antagonist muscle group moments were "more than proportional" relative to the net joint moment. Finally, the simultaneous activation of agonist and antagonist muscles also varied depending on the mode of contraction of the agonist muscle group revealing different roles of antagonist cocontraction. Our findings may extend the use of squat exercises for rehabilitation programs (Kellis, 1998). Indeed, working with weak loads following surgery would develop the capacities of force production in the knee extensor muscles with minimal risk of injuries. Latter, adding load would enhance the efficiency of the neuromuscular system in actively assist joint stability. The differences observed in the agonist and antagonist muscle groups moments between the concentric and the eccentric phases of motion may also improve the efficiency of the closed chain rehabilitation protocols, as both modes of contraction are usually encountered in daily functional activities.

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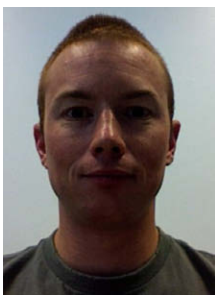

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