Tibiofemoral Joint Kinetics During Squatting With Increasing External Load

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Context: There is limited information about the effects of increasing load while squatting. Objective: To quantify tibiofemoral joint kinetics during squatting with variable loads. Setting: Research laboratory. Participants: 20 male students. Intervention: Tibiofemoral joint kinetics and electromyographic (EMG) activity of four involved muscles were determined by recording the half squat with variable external loads. Main Outcome Measures: Tibiofemoral joint force and external moment components and EMG activity of four involved muscles. Results: Throughout the exercise, a posterior direction for the antero-posterior shear force and a net extension for the external moment were observed. They increased with knee flexion reaching peak force of 29% of the subject body weight (BW) and moment of 88Nm (without external load). All force and moment components and muscle activities increased as the external load increased. Conclusion: These findings suggest that half squat may be safe to use for quadriceps strengthening with very low potential loading on the anterior cruciate ligament (ACL). Our data can help clinicians choose the appropriate external load.

Strengthening exercises that incorporate the use of barbells are considered an integral part of sports training programs and knee rehabilitation. In general, emphasis is put on the muscular aspect, whereas the articular aspect is often neglected even though these exercises may induce important joint mechanical stress which can worsen the syndrome of articular pathology or even cause serious injuries if they do not respect the physiological limits of ligamentous and/or articular structures. The lack of information about the forces carried by the ligaments or ligament grafts during the different strengthening exercises has led to disagreements between specialists as to what constitutes a good rehabilitation program.

The squat exercise is a classic multiple-joint exercise that has become an important component of lower limb strengthening programs in physical therapy and in
a variety of sports. More specifically, the squat has been used as part of treatment of ligament lesions, patellofemoral dysfunctions, total joint replacement, and ankle instability. It is performed in a continuous motion at the 40° (semi squat), 70 to 100° (half squat) and larger than 100° (deep squat). Knowledge of the kinetic parameters occurring during squatting exercise is necessary for clinicians when choosing a suitable rehabilitation program for a particular injury. Although ACL forces are believed to be low during squatting exercises, little experimental data are available to corroborate this claim. Measuring in vivo forces during an activity is a difficult task. In addition, the complex structure and loadings of the internal soft tissues make in vivo measurements even harder. It was stipulated that in vivo tibiofemoral contact forces were difficult to measure because the joint is encapsulated, articulating, and difficult to access. An alternative approach consisted of calculating ligament forces indirectly, starting from noninvasive in vivo experimental measurements and using mathematical models of the joint. Several studies have used this approach to estimate muscle forces and joint shear and compressive forces during various rehabilitation exercises. These studies were limited to 2 dimensional (2D) models and hence 2D force components. Few authors have attempted direct measurement of ligament forces in vitro, under different load conditions. They have used methods that included attaching buckle transducers to the ligaments or attaching load cells to mechanically isolated bone plugs containing the tibial attachments of the ACL and the posterior cruciate ligament (PCL). The two main disadvantages of in vitro methods are that it is very difficult to replicate the loading experienced by the knee joint during in vivo activity, and that these methods are invasive and destructive. Other studies have focused on the EMG patterns of squat exercises to come up with safe squatting strategies or to explain why ACL forces may be relatively low during squatting exercises.

Findings of investigations conducted in sports resistance training programs showed that knee joint stress generated during squatting depends on the lifted load and on the adopted technique. Cadavric studies and biomechanical and EMG analysis have compared the effectiveness of squatting (closed kinetic chain exercises) and open kinetic chain exercises for rehabilitation programs after ACL reconstruction. Their results suggest that tibiofemoral shear forces are posterior during half squat exercise, increase during flexion, and decrease during extension. To our knowledge, there is no experimental data measuring the effects of increasing load on the knee joint kinetics during squatting, except for the study comparing the patellofemoral joint kinetics while squatting with and without an external load.

Therefore, the first purpose of this study was to quantify the tibiofemoral joint force and external moment components, and to measure the EMG activity of four selected lower limb muscles during half-squat exercise using in vitro measurements, a 3D biomechanical model, and inverse dynamic analysis. The second purpose was to determine the effect of a variable external load on the estimated parameters to help assess whether the half squat with different external load can be safely used in knee rehabilitation programs.
Methods

Design
This study involved noninvasive experimental measurements of ground reaction forces and moments, position of markers on the dominant lower limb, and EMG activity during half squat exercise. It used a 3D biomechanical model based on inverse dynamics analysis to calculate tibiofemoral joint force and external moment components. The dependent variables in this study included the tibiofemoral joint force and moment components and the peak root mean square (RMS) values of the EMG activity of four involved muscles. The independent variables consisted of the five different load conditions.

Subjects
Twenty volunteer male physical education students (age = 23 ± 5 years, height = 178 ± 6 cm and body mass = 78 ± 8 kg) participated in the study. Subjects were included in the study if they had previous experience in performing half squat exercises with barbells, and were able to lift a load of 120% BW. They were excluded if they had any lower limb injury during the 2 months preceding the study, or any surgery in the preceding 12 months. All subjects signed informed consent before the experiment.

Instrumentation
A 3D optoelectronic system (SAGA3) which included four cameras was used in this study. We recorded, at a frequency of 50 Hz, the positions of 10 reflective markers attached to the dominant lower limb of each subject at the following bony landmarks as illustrated in Figure 1: the second and fifth metatarsals, the posterior extremity of the calcaneum, the fibular malleolus, the tibial malleolus, the external and internal tibial tuberous, the external and internal femoral condyles, and the greater trochanter. The four cameras were placed in such a position so that each marker is concurrently viewed by at least two cameras to allow 3D reconstruction of the reflective maker coordinates using direct linear transformation. Marker images were digitized and their 3D coordinates relative to a global laboratory reference frame were computed by a 3D stereo-correspondence software.

A force platform (AMTI) was used to measure 3D ground reaction forces and their corresponding moments in orthogonal directions at a sampling frequency of 100 Hz. While performing the exercise, all subjects placed the dominant foot on a preoutlined position on the force platform.

Bipolar pregelled circular surface electrodes with a diameter of 1 cm (model N-00-S, MyoData system) were placed on the following muscles on the dominant lower limb: vastus medialis (VM), vastus lateralis (VL), gastrocnemius (Gas), and gluteus maximus (GM). Two electrodes were placed over each muscle belly with their centers 2 cm apart. The skin surface was shaved, abraded, and cleansed with alcohol to reduce skin impedance before the attachment of the electrodes. The
EMG signals were recorded by a MyoData EMG system at a sampling frequency of 1000 Hz.

The force plateform and EMG data were digitized at 12 bits and connected to the optoelectronic system to synchronize all measurements.

**Experimental Procedures**

All subjects performed the half squat movement with an external weight in the form of a barbell centered across the shoulders. All subjects in our study were classified as right lower limb dominant, as this was the limb with which they kicked a ball. The subjects were instructed to stand with their heels remaining in contact with the ground, feet parallel, shoulder width apart, and with the right foot only on the force platform. From this prescribed position, they were asked to perform the half squat in a continuous consistent motion by bending the knees and lowering the body until their buttocks touched a blocking stretched rope (descending phase) and then returning to the standing position (ascending phase) while keeping the back straight. The rope was individually preadjusted for each subject using a universal goniometer to mark the position where knee flexion reached 90°. This controlled the amplitude of knee flexion and kept it consistent throughout each trial and across all subjects. The descending phase and its subsequent ascending phase occurred smoothly each over a time interval of one second. An audio metronome
and occasional voice commands were used to help each subject maintain the proper rhythm of movement. During the pretest session each subject practiced the half squat exercise to ensure consistent cadence and range of motion. Adequate warm-up time was allowed before beginning the actual testing session.

The testing session included performance of a set of three repetitions of a half squat exercise by each subject. The load to be lifted was individualized based on the subject’s BW. Five load conditions with nominal values of 0%BW, 50%BW, 75%BW, 100%BW, and 120%BW were used.

**Biomechanical Model and Data Processing**

The tibiofemoral joint force and external moment components were calculated with respect to the cardanic axes of the knee joint coordinate system. We refer to the medio-lateral shear component as Fx, the antero–posterior shear component as Fy, and the compression–distraction component as Fz. The external moment components are expressed as flexion–extension (Mx), varus–valgus (My), and internal–external rotation (Mz). All force values are normalized by BW and expressed as a percent BW to ease comparison with previously published data. Our measured time periods were also normalized by the mean value of the movement duration.

Our force and moment calculations were based on the conventional inverse dynamics analysis\(^3\) that uses a 3D biomechanical geometrical lower limb model and assumed that the right and the left sides of the body were symmetric. This model consisted of three rigid body segments (foot, shank, and thigh), linked by two pin-frictionless joints (ankle and knee). The inputs required by the inverse dynamics analysis were the ground reaction force and its moment, the linear accelerations of the centers of mass, the angular accelerations of each segment, and the following inertial parameters for each of the three segments: mass, coordinate of the center of mass, and moments of inertia. The ground reaction force and its moment were provided by the force platform. The accelerations of segments were determined by differentiation from the position of markers. The inertial parameters were extracted from the position of the markers and the anthropometric data tables.\(^3\) We developed a MatLab code to implement the numerous operations involved in inverse dynamics analysis. The force platform data and the 3D coordinates of the reflective markers were first numerically smoothed using cubic spline interpolation.\(^4\)

The EMG signals collected for the four muscles during the movements were rectified, filtered (low-pass at 20Hz using a second-order Butterworth filter), and normalized to the averaged EMG signal recorded for the tested muscle during maximum voluntary isometric contraction (MVIC). The RMS of the EMG amplitude was calculated over the entire range of motion window for each half squat trial, and then normalized by MVIC.

Calculated force, moment, and peak RMS values from the three half squat trials were averaged for each subject. Data for all subjects were then averaged and plotted.

**Statistical Analysis**

ANOVA with repeated measures design was used to quantify the relationship between the load conditions and the following dependent variables: the average values of tibiofemoral force and external moment components and the peak RMS
values of the EMG activity of VL, VM, GM, and Gas muscles. The alpha level was set at $P < .05$. Tukey post hoc testing was performed in the case of a significant interaction.

## Results

### Tibiofemoral Joint Force and External Moment Component Patterns During a Cycle of Half Squat Exercise

Figure 2 shows the average tibiofemoral force components under the condition of zero load. In the first 10\% period of the movement, when knee angle ranged between 160\° and 148\°, all force components had almost constant amplitudes of

![Figure 2](image)

**Figure 2** — Mean ± standard error of the tibiofemoral force components during half squat exercise as a function of knee angle. Solid lines show mean of subjects; dashed lines indicate standard error from the mean. Top to bottom: medio-lateral shear force ($F_x$), antero-posterior shear force ($F_y$), compressive force ($F_z$).
0.1% BW, 10% BW and 60% BW, for Fx, Fy, and Fz, respectively. As the movement progressed, 10% to 40% of the movement cycle, knee flexion increased continuously reaching a final value of 93°, while Fx and Fy increased progressively reaching values of 3% BW and 27% BW, respectively. Subsequently, the knee angle remained almost constant until t = 60% of the movement cycle. Fx increased steadily, and Fy showed only a slight increase. At t = 66% of the movement cycle, Fx and Fy reached their maximal values of 6% BW and 29% BW, respectively. Then they decreased to recover their initial values at t = 90% of the movement cycle. Large magnitudes of the compressive force, Fz, ranging between 45% BW and 60% BW were produced throughout the half squat movement with maximum values observed in the beginning and at the end of the movement.

Figure 3 illustrates the zero load results for the external moment components, Mx, My, and Mz. At the beginning of the movement knee angle and external

![Graph](image_url)

**Figure 3** — Mean ± standard error of the tibiofemoral external moment components during half squat exercise as a function of knee angle. Solid lines show mean of subjects; dashed lines indicate standard error from the mean. Top to bottom: flexion-extension moment (Mx), varus-valgus moment (My), and internal-external rotation moment (Mz).
moment components were relatively constant until \( t = 10\% \) of the movement cycle with values of 23Nm, 29Nm and 6Nm for \( M_x \), \( M_y \) and \( M_z \), respectively. As knee angle decreased to its minimum value of 89°, \( M_x \), \( M_y \), and \( M_z \) increased steadily reaching 83Nm, 44Nm and 20Nm, respectively. \( M_x \) shows the highest peak value of 88Nm, compared with \( M_y \) and \( M_z \) peak values of 48Nm and 21Nm, respectively. At the end of the movement the knee angle increased rapidly from 92° to 160°, while the absolute values of the external moment components decreased rapidly until recovering their initial values.

It is important to mention that the same patterns as those seen in Figures 2 and 3 have been observed for the tibiofemoral force and the external moment components under all load conditions. We choose to display the results corresponding to zero load to facilitate comparison with previously published data.

**Effects of Increasing Load**

As shown in Figure 4, the peak compressive and shear force components increased significantly \((P < .001)\) as the external load increased. The peak compressive force under unloaded condition was equal to 58%BW, while that under 120%BW load was equal to 149%BW. Peak shear forces, \( F_x \) and \( F_y \), increased from 8%BW to 11%BW and from 46%BW to 67%BW, respectively, when the load increased from 50%BW to 100%BW. The largest force increase due to load was observed for the antero-posterior shear force, but this was not statistically verified. For all load conditions, the peak compressive force \((F_z)\) was the largest component compared with the peak shear components. The peak medio-lateral shear force \((F_x)\)

![Figure 4](image)

**Figure 4** — Mean peak tibiofemoral force components during half squat exercise for each external load condition: medio-lateral shear force \((F_x)\), posterior shear force \((F_y)\), and compressive force \((F_z)\). Error bars represent standard deviation from the mean.
was the lowest component at peak values that did not exceed 13%BW even under the largest load condition.

The peak tibiofemoral external moment components increased significantly \((P < .001)\) when the lifted load increased as illustrated in Figure 5. For a load increase from 75%BW to 120%BW, the peak extension moment (Mx), increased from 178Nm to 205Nm, the peak varus moment (My) increased from 76Nm to 101Nm and the peak internal rotation moment (Mz) increased from 38Nm to 48Nm. Irrespective of the magnitude of lifted load, the peak extension moment values (Mx) were always superior to the peak values of varus moment (My) and internal rotation moment (Mz).

Figure 6 reveals that there is a significant \((P < .001)\) effect of load on the peak RMS values for all muscles. The peak RMS values ranging from 50%MVIC to 75%MVIC for a load ranging from 0%BW to 120%BW were registered for the VL muscle under all load conditions. The Gas peak RMS values were significantly lower than the other three muscles, and ranged between 15%MVIC to 30%MVIC. The VM and GM peak RMS values increased from 43%MVIC to 70%MVIC and from 15%MVIC to 50%MVIC, respectively, when the load increased from 0%BW to 120%BW. The GM peak RMS values showed the highest rate of increase with load. The rate of increase of the GM peak RMS values was almost constant across the whole load range (0%BW to 120%BW). On the other hand the rate of increase of the peak RMS values of VL, VM, and Gas were higher in the 0%BW to 50%BW load range than in the 50%BW to 120%BW load range.

![Figure 5](image)

**Figure 5** — Mean peak tibiofemoral external moment components during half squat exercise for each external load condition: extension external moment (Mx), varus external moment (My), and internal rotation external moment (Mz). Error bars represent standard deviation from mean.
The focus of this study was to quantify the tibiofemoral joint force and external moment components during half squat exercise with variable loads. Knowledge of these force and moment components during activity could be useful for aiding practitioners in developing successful cruciate ligament rehabilitation programs.

During the half squat, the cruciate ligaments, ACL and PCL, developed an antero-posterior shear force to resist to the shearing induced by muscle action or external loads. Thus the direction of this shear component can indicate which ligament is loaded and can give an estimation of the magnitude of the associated stress.

Tibiofemoral Joint Force and External Moment Components at Zero Load

In our study the medio-lateral shear force (Fx) was found to be very low with a peak value of 6%BW. This knee shear component was attributed to the stress of the collateral ligaments which do not significantly participate in the half squat movement. It was pointed out that this force component is not injurious to the tibiofemoral joint because the intercondylar eminence on the tibial plateau inserting into the intercondylar notch of the femur during weight-bearing likely supports most of the medio-lateral shear forces rather than the knee cartilage.

We noted large magnitudes of tibiofemoral compressive force (Fz) ranging between 45%BW and 58%BW during the half squat movement. The maximal value of 58%BW (approximately 450N) was registered in the standing position near full extension at a knee angle about 160° with all body segments almost...
aligned, and hence the external forces being essentially vertical as reported by other studies.32,33,35,36 These results agree with the observation that the compressive force remained almost constant during half squat exercise with a peak value about 60%BW (ranging between 500N and 600N) for a load of 223N.35 As the half squat exercise was performed with the subject erect, the weight-bearing force was mainly applied vertically onto the tibiofemoral joint and hence does not induce shearing. The benefit of the tibiofemoral compressive forces in the rehabilitative process is the facilitation of cocontraction between the quadriceps and the hamstring muscles at the knee joint.33

Our study revealed that the highest antero-posterior shear force value (Fy) happened at the lowest knee position. The distance between the center of the knee joint and the line of action of the external resultant tibiofemoral force increased during flexion. This distance decreased with extension resulting in a decrease of the antero-posterior shear force until it reached its initial value at full extension when the center of the knee joint lay approximately on the line of action of the external resultant tibiofemoral force. Our results indicated that Fy remained posterior during the entire half squat movement cycle consistently with previous studies.32,33,35 This suggests that Fy was mainly PCL related. Our peak values of 29%BW were within the range of peak values reported by the aforementioned studies, although they showed a large variation which may be due to differences in experimental conditions or biomechanical methods.

Some studies have decomposed the antero-posterior shear force into muscle forces and passive tibiofemoral shear forces, and then isolated the passive cruciate ligament forces.3,18,24,36 A peak PCL force around 3.5×BW was observed at the beginning of the ascent phase.3 Given that the shear force is equal to the ligament force times the cosine of its angle of inclination with respect to the tibial plateau3 and that this angle increases with increasing flexion,42 and assuming a PCL angle higher than 50° as suggested in41 we concluded that the PCL forces deduced from our posterior shear forces would be of a similar order of magnitude as those reported.3

While there are some reports on the extension moment (Mx) to our knowledge, the varus-valgus (My) and the internal-external rotation (Mz) moments generated during squatting have not been investigated in any studies that use biomechanical models. Our calculations revealed larger magnitudes of Mx compared with My and Mz, with maximal Mx value of 88Nm in agreement with other reports.35 This suggested that the half squat exercise occurred mostly in the sagittal plan, and that the muscles involved were mainly knee extensors which closely replicated the EMG activity of the quadriceps measured in our study. In the frontal plan, a relatively important My with peak value of 48Nm was observed while in the transversal plan an Mz peak value of 21Nm was registered.

All external moment components increased during the descending phase reaching their maximal values at the beginning of the extension and decreased during the ascending phase recovering their initial values at the end of the movement. In the descending phase, the subject bent forward, knee flexion increased, and as a consequence the knee joint center moved away from the line of action of the weight of the subject-load system, thus inducing large values of external moment components. While in the ascending phase, the knee joint center got closer to the line of action of the resultant force, resulting in a decrease of external moment components. The peak values of the external moment components were reached at the beginning of
the extension when the acceleration of the movement was maximal. The behavior that we observed for the extension moment component (Mx) has been verified by other reports.\textsuperscript{33,35} The varus moment (My) and the internal rotation moment (Mz) were produced to prevent the spreading of the external tibial condyle from the vertical body axis.

**Effect of Increasing Load**

Limited information has been published on the effect of external load. Our results agree with a reported peak posterior shear force of 35\%BW calculated from the 3D model of the half squat performed with 223N load.\textsuperscript{35} A Peak tibiofemoral extensor moment close to our results has been reported with a load of 70\%BW.\textsuperscript{43}

Our study showed that the peak tibiofemoral force and external moment components increased with the external load during half squat. In the field of knee rehabilitation, studies of the squat are frequently done under zero load condition. Few researchers concluded that the magnitude of the joint forces was directly related to the amount of resistance applied,\textsuperscript{33} while others corroborate this finding for the patello-femoral joint forces.\textsuperscript{37} A study in the field of sports resistance training verified the force-load relation for the antero-posterior shear component,\textsuperscript{30} while another confirmed this relation for the peak antero-posterior shear and compressive force components.\textsuperscript{28} Published data\textsuperscript{29} agrees with our observation that an increase in the external load was accompanied by an increase in the external moments. The effect of load increase that we observed for the tibiofemoral force and external moment components was consistent with the significant increase of EMG activity that we measured for all muscles. Many authors indicated that the quadriceps muscle force may cause anterior tibial displacement especially near full knee extension. This displacement is counteracted by the quadriceps/hamstring cocontraction during squatting.\textsuperscript{23,25,27,33,35} It is believed that this cocontraction contributes to neutralization of tibiofemoral shear forces imparted by the quadriceps,\textsuperscript{35} thus providing a stabilizing force at the knee during squat.\textsuperscript{25} The extensors of the hip joint (GM) demonstrated the highest rate of increase of EMG activity; whereas the knee extensors (VL and VM) presented the lowest rate of increase. This agrees with reports that the rate of increase of the peak hip external moment as a function of load was greater than the rate of increase of the peak knee external moment.\textsuperscript{29,44} In fact the GM is more sensitive to load because during squat exercises the extensor muscles of the hip joint act in a compensatory way with those of the knee joint.\textsuperscript{29} It is thought that due to its attachment into the iliotibial tract, the GM may play a role in stabilizing the pelvis and the knee in extension such as in squatting, thus enabling a safe and effective movement to occur particularly when the load is heavy.\textsuperscript{45} It has been demonstrated that the main effort to lift the weight with a squat technique is provided by the hip extensors under increasing load.\textsuperscript{44} The higher values of EMG activity for VL and VM that we observed compared with GM and Gas, were in agreement with reports that these muscles are the most active muscles during squat.\textsuperscript{4,24,25,27,33} Most of the studies of EMG activity during squatting tend to pay attention specifically to VM, VL, and hamstring muscles, but few reports have shown that the response of GM and Gas may also be relevant.\textsuperscript{25,46} In our study, the GM has been shown to be active with an important EMG activity level of 49\%MVIC for the heaviest load condition. For unloaded half squat, we recorded a low level of Gas EMG activity (13\%MVIC)
in agreement with. The highest value of Gas EMG activity that we measured is 28% MVIC when squatting with the heaviest external load. This relatively high value suggests that the Gas may possibly limit posterior tibial translation relative to the femur because its origin is on the posterior femoral condyles.

**Limitations of Method**

As mentioned earlier, direct measurement of ligament forces in vivo is very difficult. A method involving noninvasive experimental measurements combined with a biomechanical model of the lower limb has been used in this study. It differs from methods used in previous experimental studies of rehabilitation exercises in that it allows the determination of the 3D tibiofemoral force and external moment component magnitudes and patterns rather than just 2D. It is important to understand the limitations of using a 3D model to study the tibiofemoral joint kinetics. There are certain advantages and disadvantages when presenting the data as total joint kinetics using precise motion analysis and load measurement devices. Total joint kinetics could be calculated with a great degree of confidence and precision; however, as the action and moment arms of the major force-bearing structures crossing the human knee joint of the subject are not known, it is somewhat difficult to extrapolate the magnitude of the stress applied to the different knee joint structures such as bones, muscles, and ligaments. In our model ligaments were assumed to produce a shear force, whereas bones were assumed to produce a compressive force. These approximations are common in biomechanics; nevertheless, it is important to realize that the calculated forces and external moments represent estimations of actual joint dynamics. Other biomechanical models used to separate kinetics into more detailed knee joint structures introduce many assumptions and simplifications leading to overestimation of the tibiofemoral force and external moment components. Because overestimation is built in biomechanical methods, they effectively include a safety factor, and hence are appropriate in a clinical context.

**Clinical Relevance**

The recent trend toward accelerated rehabilitation was initiated to minimize muscle atrophy and maximize knee joint movement without damaging the joint structures or the graft. The kinetic and EMG data that we generated for squatting will eventually help rehabilitation practitioners to adjust their squat-based knee rehabilitation programs. The results presented in this study can be used as a guide to the maximum tibiofemoral force and external moment components under different load conditions. The posterior direction of the antero-posterior shear force revealed that squat may be a safe exercise to use for quadriceps strengthening with very low potential loading on the injured or reconstructed ACL. The PCL was found to be loaded with a posterior shear force that increased with knee angle and external load, reaching up to 70% BW at an external load of 120% BW. This suggested that deep squats and squatting with heavy external loads should be avoided for rehabilitation of the PCL. Half squat exercises with external loads lower than 50% BW led to posterior shear forces that are not likely to be detrimental to injured or reconstructed PCL.

For improved clinical relevance, future investigations should be performed to systematically characterize the effect of other various experimental conditions such us foot placement, load position, knee flexion amplitude, and movement rhythm, on the kinetic patterns associated with squatting.
Conclusions

Our finding that the antero-posterior tibiofemoral shear force has a posterior direction emphasizes that caution must be taken when using half squat with subjects that have PCL disorders, especially at large knee flexion angles and under extreme load conditions. This observed force direction suggests that the half squat with external load may be safe to use for quadriceps strengthening with very low potential loading on the injured or reconstructed ACL. Our study can help clinicians choose appropriate external load conditions for their rehabilitation programs.

References


