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original research

A Biomechanical Investigation of a Single-Limb Squat: Implications for Lower Extremity Rehabilitation Exercise

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Context: Single-limb squats on a decline angle have been suggested as a rehabilitative intervention to target the knee extensors. Investigators, however, have presented very little empirical research in which they have documented the biomechanics of these exercises or have determined the optimum angle of decline used.

Objective: To determine the involvement of the gastrocnemius and rectus femoris muscles and the external ankle and knee joint moments at 60° of knee flexion while performing a single-limb squat at different decline angles.

Design: Participants acted as their own controls in a repeated-measures design.

Patients or Other Participants: We recruited 10 participants who had no pain, injury, or neurologic disorder.

Intervention(s): Participants performed single-limb squats at different decline angles.

Main Outcome Measure(s): Angle-specific knee and ankle moments were calculated at 60° of knee flexion. Angle-specific electromyography (EMG) activity was calculated at 60° of knee flexion. Integrated EMG also was calculated to determine the level of muscle activity over the entire squat.

Results: An increase was seen in the knee moments ($P < .05$) and integrated EMG in the rectus femoris ($P < .001$) as the decline angle increased. A decrease was seen in the ankle moments as the decline angle increased ($P = .001$), but EMG activity in the gastrocnemius increased between 16° and 24° ($P = .018$).

Conclusions: As the decline angle increased, the knee extensor moment and EMG activity increased. As the decline angle increased, the ankle plantar-flexor moments decreased; however, an increase in the EMG activity was seen with the 24° decline angle compared with the 16° decline angle. This indicates that decline squats at an angle greater than 16° may not reduce passive calf tension, as was suggested previously, and may provide no mechanical advantage for the knee.

Key Words: knee moments, electromyography, movement analysis

Key Points

- The 16° decline angle provided the maximum benefit for the knee extensors with the minimum effect for the ankle.
- The 24° decline angle provided a greater challenge to the ankle and targeted the knee extensors.
- The single-limb squat may produce significant cocontractions about the knee and ankle joints that the external net joint moments alone do not reflect.

The use of eccentric activities for rehabilitation has been well documented,[1–7] with researchers showing that these exercises have a significant effect on rate of recovery.[1–9] Khan et al[8] explained that although they appear to have a clinical effect, many of the eccentric exercises and techniques employed have little scientific background to support their use. Purdam et al[10] identified this as an area for further investigation and proposed a nonoperative management technique for patellar tendinopathy. The technique was based on performing a single-limb squat with the eccentrically controlling limb placed on a 25° decline. The basis for using a 25° decline was that, by forcing the ankle into plantar flexion, passive and active calf tension are reduced; therefore, the work done about the ankle is reduced, and the exercise to target the knee extensors is more focused.[10–12] The concept of increased work of the knee extensors is supported by evidence suggesting that performing single-limb squats on a decline board at inclinations greater than 15° can increase the knee-flexion moments by up to 40%.[13] However, little additional evidence is available to suggest what other characteristics might be associated with these exercises, particularly in relation to passive and active tension of the calf muscles. Although empirical data relating to the biomechanics of single-limb squats are limited, researchers recently have documented interesting biomechanical characteristics associated with double-limb squats.[14,15] Kongsgaard et al[15] suggested that knee extensor demand increases while performing decline squats versus a level squat, but gastrocnemius activity can actually increase rather than decrease. Other authors have suggested that decline squats may not reduce passive calf tension as was reported previously.[10,11]

Investigators have documented only the use of 25° decline angles[10–12,14,15] and 30° decline angles.[13] To begin to understand the biomechanics of single-limb squat exercises, researchers must investigate the biomechanical demands associated with several decline angles. As with the suggested angle of inclination, the suggested depth of squat based on the angle of knee flexion[9,11,16] varies between 50° and 90°. Again, very little evidence is available regarding why these angles have been used. Initially, Purdam et al[16] proposed 50° of knee flexion because the force on the patellar tendon is equal to that on the quadriceps tendon when it is in this orientation. However, 90° angles[10] and 70° angles[11] of knee flexion also have been used. Based on these variations in the range of flexion, we see no consensus within contemporary research to support a set limit to the range of motion that is required during a single-limb squat. Clinically, however, the amount of knee flexion that different individuals are able to achieve during eccentric squat activities can be considerably different, so the relevance of controlling the amount of knee flexion is debatable. Despite this, investigators have controlled the knee angle[10,11,16]; however, no empirical research is available to document the angle that participants in studies involving single-limb squats commonly achieve.

Purdam et al[10] identified that further study of eccentric exercises is essential for more validation of these single-limb squat exercises. Although many researchers have highlighted the effectiveness of decline squats as a rehabilitative tool[8–13] and have documented the biomechanical characteristics associated with double-limb squats,[14] very little is known about the biomechanics of these exercises. The purpose of our study was to investigate the biomechanical characteristics of and muscular involvement during single-limb squats performed on a flat surface and at different decline angles. We hypothesized that, based on biomechanical principles, increasing the decline angle would result in decreased ankle-dorsiflexion moments and in decreased activity of the gastrocnemius while increasing the work of the knee extensors.

METHODS

Participants

We recruited 10 participants (age = 21 ± 6.7 years, mass = 70.7 ± 11.5 kg) who had no pain or disorder and were recreationally active university students and staff. We did not document their height. Data were collected from the dominant limb of each participant; the *dominant limb* was defined as the limb with which they would kick a football. Volunteers gave written informed consent before data collection. All data collection conformed to the Declaration of Helsinki. The study was approved by the Faculty of Health Research Ethics Committee, University of Central Lancashire.

Instrumentation

Kinematic data were collected using a 6-camera ProReflex system (Qualisys Medical AB, Gothenburg, Sweden) at 100 Hz. Force data were collected using an AMTI force platform (Advanced Mechanical Technology, Inc, Watertown, MA). We collected electromyographic (EMG) data from the rectus femoris and gastrocnemius muscles using an 8-channel Bagnoli system (Delsys, Inc, Boston, MA). The rectus femoris and gastrocnemius were selected because the muscles share the neuromuscular control mechanisms commonly associated with biarticular muscles during closed kinetic chain exercises.[17] The EMG used Ag/AgCl single differential electrodes with a dimension of 10 mm × 1 mm and a set electrode distance of 10 mm. The electrodes were preamplifed and set to a gain of 1000. The common mode rejection ratio was less than 80 dB. The skin was cleansed with alcohol wipes. Electrodes were positioned over the muscle belly in an attempt to reduce crosstalk, and they were attached using standard DelSys single differential interfaces. All data were collected at 2000 Hz with the use of a 16-bit analog-to-digital converter.

Modeling of the Lower Limbs and Joints

The segments of the lower limbs were modeled based on the calibrated anatomical systems technique (CAST).[18] The CAST marker system models the segments of the body in 6 degrees of freedom by defining an anatomic reference frame based on palpable anatomic landmarks. We used the medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, anterior-superior iliac spines of the pelvis, and posterior-superior iliac spines of the pelvis (Figure 1). Clusters of 4 markers mounted on rigid plastic shells were attached to each segment. After placing all of the markers, we performed a calibration that consisted of data collection for 1 second with the participant standing in the anatomic position. This defined the anatomic coordinate systems that enabled the position and orientation of each segment in space to be identified.[18] Local coordinate systems were defined for all segments of the model, with the y-axis equal to anterior-posterior, x-axis equal to medial-lateral, and z-axis equal to proximal-distal. The centers of the knee and ankle joints were calculated as the median distance between the medial and lateral joint markers. The center of the hip joint was calculated based on pelvic depth and width using the regression equations developed by Bell et al.[19] Joint kinematics were calculated using a Cardan/Euler method with an XYZ order of rotations.

Procedures

We selected 4 decline angles (0°, 8°, 16°, and 24°) to allow for intercomparisons within the range of values previously published.[10–15] Four comparisons were considered to be within the limits of the participants beginning to fatigue. To enable the participants to achieve these angles, we placed an adjustable board on top of the force platform. When a participant stood on the board, his or her ankle was placed into 1 of 4 plantar-flexed angles. The placement of this board did not affect the force data in any way because, at the start of data collection, the force platforms were zeroed. Therefore, the board had an effective mass of 0. The order in which the decline angles were assigned to the participants was randomized by reversing the order and moving the first in the sequence to the last in the sequence. For example, if the first participant performed the squats in the order 0°, 8°, 16°, 24°, the second participant would perform the squats in the order 16°, 8°, 0°, and 24°.

Protocol

Before beginning the tests, each participant was provided with oral instructions followed by 1 practice trial to become

Figure 1. Positions of markers used to define the position and orientation in space of the body segments.

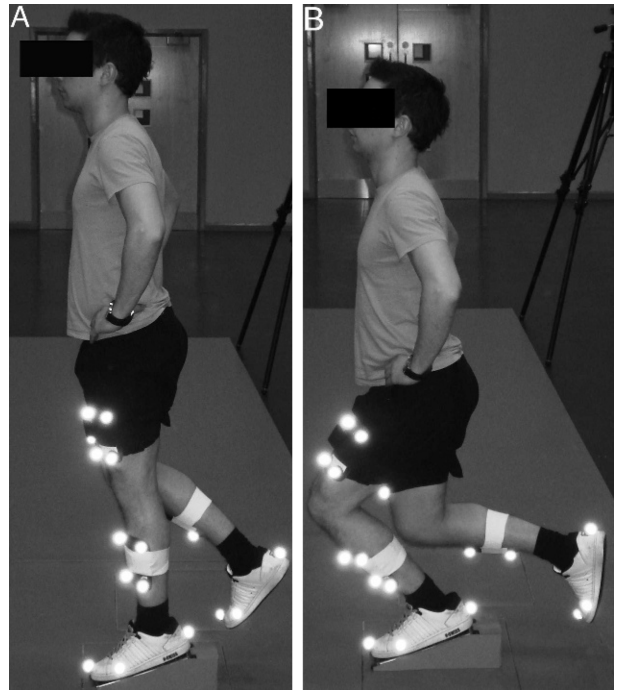

Figure 2. Experimental set-up of the board and participant performing the tests at A, the start position and B, the point of maximum knee flexion.

familiar with the procedure. Next, the participants performed 3 trials at each decline angle from which the mean results were calculated. Three trials were used to avoid fatigue. The test began with the participant positioned away from the force platform and board. On a cue, he or she was instructed to step onto the board with the dominant limb only. The participant was allowed to use the contralateral limb to stabilize himself or herself before the squat movement. We instructed each participant to perform the squat as slowly as possible to approximately 90° (self-assessed) using only the limb on the decline board. When reaching his or her maximum angle, the participant was instructed to slowly return to a straight-leg position (Figure 2). The participants performed the squat with the dominant limb only. Although participants were instructed to squat to 90° of knee flexion, not all participants could achieve this. Therefore, all data were quantified at 60° of knee flexion, because all participants could achieve this position.

Data Processing

The raw force and movement data were exported to Visual3D (C-Motion, Inc, Germantown, MD) for processing. The movement and force data were filtered using a fourth-order, low-pass Butterworth filter with cut-off frequencies of 6 Hz and 25 Hz, respectively. The EMG data were zeroed to remove any offset and band-pass filtered with a high-pass filter of 20 Hz and a low-pass filter of 500 Hz. For the calculation of angle-specific EMG activity, data were full-wave rectified and then enveloped using a fourth-order, low-pass Butterworth filter with a cut-off frequency of 25 Hz. The integrated EMG (iEMG) was calculated based on the rectified data. The knee-joint angles were calculated relative to the tibial coordinate system, and the ankle-joint angles were calculated relative to the foot coordinate system. Movement and force data were used to calculate external joint moments about the knee and ankle joints using inverse dynamics methods, and readings were taken at 60° of knee flexion.

The enveloped EMG magnitudes at 60° of knee flexion and the iEMG values during the squat were found for the gastrocnemius and rectus femoris. Enveloped EMG and iEMG were used because they provide insight into 2 different areas of muscular control and activity. The iEMG provides information about the amount of muscle activity over the entire task, but angle-specific EMG provides only the motor unit activity at 1 specific event during the task. This may be particularly useful when designing patient-specific rehabilitation, because the intensity can be tailored. The EMG and iEMG data were normalized to the maximal dynamic contraction during the movement.[20] For this purpose, we determined that the maximum level of activity during the squat would be 100%. All other data were normalized based on this.

Statistical Analysis

Repeated-measures analysis of variance and post hoc pairwise comparisons were used to identify significant differences for the EMG magnitudes, iEMG values, the knee and ankle moments at 60° of knee flexion, and maximum angle of knee flexion. The Bonferroni adjustment was used to account for multiple comparisons and to reduce the possibility of type I errors. Adjusted *P* values were reported for comparisons of the different decline positions of the foot during the squats. The α level was set at .05.

Table 1. Means (SDs) for Joint Angles, Moments, Normalized Electromyography, and Integrated Electromyography

	Decline Angle			
	0°	8°	16°	24°
Knee joint angle, °	70.9 (8.3)	70.0 (7.9)	72.9 (7.2)	72.6 (9.0)
Joint moments, Nm/kg				
Knee	1.18 (0.34)	1.29 (0.28)	1.44 (0.31)	1.43 (0.29)
Ankle	0.87 (0.16)	0.73 (0.13)	0.59 (0.20)	0.48 (0.14)
Normalized electromyography				
Gastrocnemius	0.44 (0.12)	0.44 (0.11)	0.31 (0.11)	0.51 (0.17)
Rectus femoris	0.90 (0.10)	0.93 (0.07)	0.87 (0.08)	0.90 (0.12)
Normalized integrated electromyography				
Gastrocnemius	0.59 (0.19)	0.60 (0.21)	0.76 (0.20)	0.91 (0.15)
Rectus femoris	0.41 (0.27)	0.78 (0.19)	0.91 (0.08)	0.88 (0.23)

Table 2. Means and Adjusted *P* Values of Post Hoc Pairwise Comparisons of Joint Kinematics, Moments, Normalized Electromyography, and Integrated Electromyography

	Comparisons (°)					
	0 and 8	0 and 16	0 and 24	8 and 16	8 and 24	16 and 24
Maximum knee angle (°)						
Mean difference	−2.2	−5.0	−4.7	−2.9	−2.6	−0.3
Adjusted *P* value	1.000	0.192	0.306	0.533	1.000	1.000
Maximum ankle moment (Nm/kg)						
Mean difference	−0.140	−0.280[a]	−0.390[a]	−0.140	−0.250[a]	−0.110
Adjusted *P* value	0.209	0.001	0.001	0.222	0.002	0.600
Maximum knee moment (Nm/kg)						
Mean difference	−0.110	−0.260[a]	−0.260[a]	−0.150	−0.015	0.003
Adjusted *P* value	0.193	0.012	0.044	0.095	0.143	1.000
Normalized gastrocnemius EMG						
Mean difference	−0.004	0.120	−0.070	0.013	−0.070	0.200[a]
Adjusted *P* value	1.000	0.323	1.000	0.278	1.000	0.018
Normalized rectus femoris EMG						
Mean difference	0.030	0.030	0.000	0.060	0.030	0.030
Adjusted *P* value	1.000	1.000	1.000	0.890	1.000	1.000
Normalized gastrocnemius iEMG						
Mean difference	−0.010	−0.170	−0.320[a]	−0.150	−0.310[a]	−0.140
Adjusted *P* value	1.000	0.622	0.004	1.000	0.018	0.574
Normalized rectus femoris iEMG						
Mean difference	0.370[a]	0.500[a]	0.470[a]	0.130	0.310	0.010
Adjusted *P* value	0.038	0.001	0.011	0.399	0.808	1.000

Abbreviations: EMG, electromyography; iEMG, integrated electromyography.
[a] The mean difference is significant at an α level of .05 with Bonferroni adjustment for multiple comparisons.

RESULTS

Joint Kinematics and Kinetics

A difference was seen for the maximum knee angle (Table 1) that was attained at the different decline angles (P = .009); however, post hoc pairwise comparisons showed no differences among the specific decline angles after the results had been corrected with a Bonferroni adjustment. We found a difference in knee moments at different angles of decline (P = .003). Post hoc comparisons (Table 2) showed increases in knee moments with an increase in decline angle between 0° and 16° (P = .012) and 0° and 24° (P = .044).

Differences were seen among the mean ankle moments (Table 1) at different angles of decline (P = .001). Post hoc pairwise comparisons identified decreases in ankle moments with an increase in decline angle between 0° and 16° (P = .001), 0° and 24° (P = .001), and 8° and 24° (P = .002; Table 2).

Muscle Activity Data

Angle-specific EMG activity of the rectus femoris (Table 1) recorded at 60° of knee flexion showed no differences among decline angles (P = .540; Table 2). However, maximum iEMG of the rectus femoris over the squat showed a difference among decline angles (P = .007; Table 1). Post hoc comparisons (Table 2) identified increases between 0° and 8° (P = .038), 0° and 16° (P = .001), and 0° and 24° (P = .011). The angle-specific EMG data for the gastrocnemius showed a difference among decline angles (P = .208), and post hoc comparisons (Table 2) identified an increase between 16° and 24° (P = .018). The iEMG data for the gastrocnemius also showed an increase (P = .002) in activity with decline angle. Post

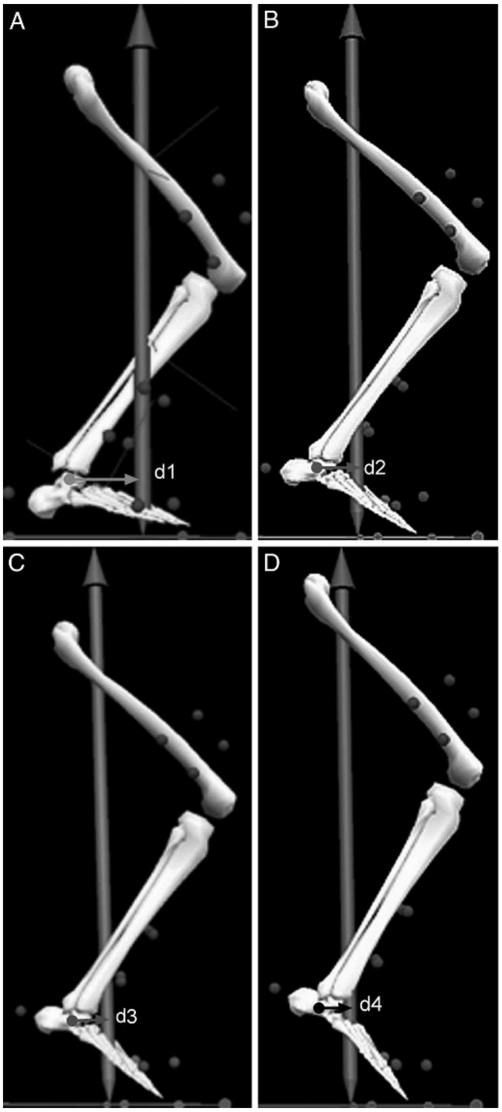

Figure 3. The position of the force vector in relation to the ankle and knee joints for each angle of inclination: A, 0°; B, 8°; C, 16°; and D, 24°. Note that distance d represents the ankle joint moment arm and that as the inclination increases, d decreases and moves in a posterior direction toward the ankle joint center.

hoc comparisons (Table 2) identified these to be between 0° and 24° ($P = .004$) and between 8° and 24° ($P = .018$).

DISCUSSION

Joint Kinematics and Kinetics

The ankle moments during a squat decreased with an increase in the decline angle; however, we only observed statistical significance with specific comparisons. This finding may be explained by the fact that, as the angle of decline increases, the center of pressure moves from an anterior position in a posterior direction toward the ankle joint. The change results from the relationship between foot inclination and the ankle joint; as the foot inclination increases, the distal and proximal ends become relatively closer along the sagittal-plane axis of the body, decreasing the moment arm about the ankle and, therefore, reducing the moment about the ankle (Figure 3).

The knee moments increased as the decline angle increased, but we only observed statistical significance with specific comparisons. Again, this finding can be related to the movement of the center of pressure in a posterior direction, which increases the flexion moment (Figure 3). The finding is interesting to compare with the findings of Zwerver et al,[13] who noted that a 15° change in angle of inclination can result in a 40% increase in knee-flexion moment. Our findings indicate that a 16° change in the decline angle can result in a 30% increase in knee-flexion moment. The mean knee joint angles indicate that, when instructed to perform a squat to 90° of knee flexion, participants on average attain approximately 70° of knee flexion. This degree of flexion may be due to either proprioceptive deficits or the mechanical demands of the exercise. These results support a suggestion that Jonsson and Alfredson[11] originally made. It is interesting to note that Zwerver et al[13] suggested that, to avoid patellofemoral pain syndrome, knee flexion should not exceed 60°.

Muscle Activity

Although the moment about the ankle was reduced with declining angle, this decrease did not correspond to the level of muscle activity seen in the gastrocnemius. Forcing the ankle into plantar flexion changes the nature of the contraction into a stabilizing mechanism, which could explain the increase in gastrocnemius activity between 16° and 24°. The discrepancy between gastrocnemius activity and net external ankle moment occurs because the ankle is placed in an unstable position and requires cocontraction of the tibialis anterior to stabilize it. Based on this explanation, increasing the angle of decline to 24° results in an increase in the activity of the gastrocnemius and does not result in a decrease, as was thought previously.[10,11] This concurs with the results of previous double-limb squat studies[14] in which the authors found an increase in gastrocnemius activity associated with increased inclination.

As the decline squat increased, the knee-flexion moment and the iEMG of the rectus femoris also increased. However, we found no difference between the knee moments and EMG activity of the rectus femoris between 16° and 24°, indicating that a maximum moment exists and that rectus femoris activity may have been attained at 16°. The iEMG of the rectus femoris and knee moments also showed the greatest value at 16°, indicating that the maximum potential benefit as an exercise for knee extensors can be attained at angles as low as 16° and that no additional mechanical advantage is gained when moving to a 24° decline. It is also interesting to note that the percentage increase in the iEMG of the rectus femoris between 0° and 24° was much greater than that in the knee-flexion moment (100% and 30%, respectively). One possible mechanism for this discrepancy is that the hamstrings contribute to this increase in rectus femoris activity through cocontraction. The disproportionate rectus femoris activity also may occur because the function of the gastrocnemius is across both the knee and ankle. Therefore, additional cocontraction of the gastrocnemius may have a restraining effect about the knee by acting as a knee flexor and producing an internal flexion moment. The biarticular action of the gastrocnemius also may control anterior translation of the femur and, thus, maintain the stability of the kinetic chain.

Clinical Relevance

To date, the decline squat has been used as an exercise to target the knee extensors.[8–12,16] The principles behind this rationale were fundamentally correct; however, the optimal method had not been identified. Our study provides a starting point for therapists when beginning the decision-making process for using single-limb squats. The decline squat enables the knee to be flexed, which may be useful if the ankle has a poor range of motion due to immobilization, and, therefore, should be given as an effective knee extensor exercise without moving the ankle into a dorsiflexed position. If the clinical reasoning for the test is to target the knee, then the data presented suggest that the 16° decline angle has the maximum effect on the knee extensors with the minimum effect about the ankle. However, a decline angle of 24° is justified if the aim of the rehabilitation program is to give a greater challenge to the ankle while also targeting the knee extensors.

CONCLUSIONS

Researchers have shown that the use of eccentric single-limb squat exercises has a significant effect on performance[8–12,16]; however, little was known about the exact biomechanics. In this study, we took biomechanical measurements and established that, as the decline angle of the single-limb squat increased, the knee-flexion moment increased, and the ankle-dorsiflexion moment decreased. We found, however, an increase in the EMG activity of the gastrocnemius at the 24° decline angle compared with the 16° decline angle, but we did not find a mechanical advantage about the knee between 16° and 24°. This finding highlights the premise that the single-limb squat may be producing significant cocontractions about each joint that are not reflected by the external net joint moments alone. Such information needs to be considered when using single-limb squats clinically. Although the results from this study are promising, the small sample size may reduce the sensitivity of the study. Through larger-scale studies, investigators may identify further differences among angles of inclination. Additionally, further studies are needed to establish the effectiveness of different angles of decline squat in specific rehabilitation programs.

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