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Bridging Motor Control and Biomechanics

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The Effect of Different Decline Angles on the Biomechanics of Double Limb Squats and the Implications to Clinical and Training Practice

by
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Bilateral decline squatting has been well documented as a rehabilitation exercise, however, little information exists on the optimum angle of decline. The aim of this study was to determine the ankle and knee angle, moments, the patellofemoral joint load, patellar tendon load and associated muscle activity while performing a double limb squat at different decline angles and the implications to rehabilitation. Eighteen healthy subjects performed double limb squats at 6 angles of declination: 0, 5, 10, 15, 20 and 25 degrees. The range of motion of the knee and ankle joints, external moments, the patellofemoral/patellar tendon load and integrated EMG of gastrocnemius, tibialis anterior, rectus femoris and biceps femoris were evaluated. As the decline angle increased up to 20 degrees, the range of motion possible at the ankle and knee increased. The joint moments showed a decrease at the ankle up to 15 degrees and an increase at the knee up to 25 degrees, indicating a progressive reduction in loading around the ankle with a corresponding increase of the load in the patellar tendon and patellofemoral joint. These trends were supported by a decrease in tibialis anterior activity and an increase in the rectus femoris activity up to 15 degrees declination. However, gastrocnemius and biceps femoris activity increased as the decline angle increased above 15 degrees. The action of gastrocnemius and biceps femoris stabilises the knee against an anterior displacement of the femur on the tibia. These findings would suggest that there is little benefit in using a decline angle greater than 15-20 degrees unless the purpose is to offer an additional stability challenge to the knee joint.

Key words: *rehabilitation, biomechanics, electromyography, knee, ankle.*

Introduction

Squats are a popular multi-joint exercise and form an integral part of most rehabilitation programmes (Escamilla, 2001). The use of eccentric squat activities for rehabilitation associated with tendinopathy has been well documented (Cannell et al., 2001; Cook and Khan, 2001; Ohberg et al., 2002; Roos et al., 2004). The exact aetiology of tendinopathy is unknown, however, evidence suggests that a biochemical and biomechanical combination contributes. The biomechanical factors include apparent disorganisation of the collagen fibres (Ohberg et al., 2002) which leads to general thickening of the tendon. In addition, neovascularisation has been identified around the area affected by the tendinopathy (Panni et al., 2000). Eccentric exercises have been shown to have a significant effect on the rate of recovery from tendinopathy (Alfredson et al., 1998; Cannell et al., 2001; Khan et al., 1998; Stanish et al., 1986), but the physiological and biomechanical effect of the exercises is somewhat unknown. Öhberg et al.

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(2002) suggested that the eccentric exercise induced remodelling within the injured tendon that reduced the neovascularisation and realigned the collagen fibres.

Khan et al. (1998) explained that many eccentric exercises and techniques used had little scientific background. Purdam et al. (2003) identified this as an area for further investigation and proposed a conservative management technique for patella 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 in to plantar flexion, passive and active calf tension were reduced, therefore reducing the work done about the ankle, thus producing a more focused exercise to target the knee extensors (Jonsson and Alfredson, 2005; Rainoldi et al., 2001); however, the efficacy of a 25° decline angle is not well established. Zwerver et al. (2008) confirmed increased knee flexion moments at the deepest point of the squat with increasing angles of declination of up to 30°. Richards et al. (2008) recorded statistically significant differences between flat squats and squats performed on a 16° decline and a 24° decline at 60° of knee flexion. Richards et al. (2008) and Zwerver et al. (2007) determined that the joint moments and muscle work done at the ankle may be controlled by altering the orientation of the foot on the squat platform by changing the angle of declination. By increasing the decline angle, a reduction in the loads and muscle activity was seen at the ankle while increasing the knee moments and muscle activity. Therefore, varying the decline angle has the potential to be used in a progressive training regimen aimed at increasing the loads around the knee while minimising the loads at the ankle. However, no information exists about the effect of different decline angles on patellar tendon and patellofemoral loads, although recent work has shown that the patellar tendon load may be estimated during eccentric squats (Frohm et al., 2007).

A number of authors have suggested that a decline squat in comparison to a flat squat produces a significant improvement in the ability of the individual to participate in sports and a reduction in pain (Alfredson et al., 1998; Jonsson and Alfredson, 2005; Khan and Maffulli, 1998; Purdam et al., 2003). There is however no scientific justification given as to why 25° was chosen for the decline. The suggested depth of the squat based on the angle of knee flexion varies between 50° and 90° (Alfredson et al., 1998; Young et al., 2005). Initially Purdam et al. (2003) proposed 50°, with the basis for this being that the force in the patellar tendon is equal to that of the quadriceps tendon when in this particular orientation. However, subsequently Purdam et al. (2003) proposed 90° of flexion, with Jonsson and Alfredson (2005) using 70° of flexion. Based on this variation in the range of flexion, within the literature there is no consensus within contemporary research. However, there can be considerable differences in the amount of knee flexion different individuals are able to achieve during eccentric squat activities and the relevance of controlling the amount of knee flexion is debatable in clinical practice.

Purdam et al. (2003) identified that further study of eccentric exercises was essential for further validation of these single limb squat exercises. In particular, it was identified that biomechanical studies of flat and decline squats were essential. Therefore, the aim of the current study was to investigate the biomechanical effects and muscular involvement when performing squats on different decline angles and to discuss the implications to different rehabilitation protocols; the hypothesis being that there is a biomechanically optimum angle for decline squats for knee rehabilitation.

Material and Methods

Participants

Eighteen pain and pathology free participants were recruited (9 males and 9 females) with an age range between 20 to 46 years, mean body mass of 75.1 kg (a range of 58.3 to 100 kg), all of whom were recreationally active university students and staff. All participants were trained how to perform a squat by an experienced physiotherapist. 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. Research procedures conformed to the Declaration of Helsinki with volunteers giving written informed consent prior to data collection. Prior to any testing ethical approval was gained from the institutional ethics committee (Faculty of Health, University of Central Lancashire).

Decline Squats

Six decline angles (0, 5, 10, 15, 20 and 25 degrees) were selected to perform five double limb squats using the "Rehab Angel" which is an adjustable incline/decline board rehabilitation device (Picture 1). The order in which the decline angles were assigned to the participants was randomised. The participants were required to perform 5 trials at each decline angle from which the mean results were calculated. Prior to beginning the tests, the participants were provided with verbal instructions followed by a practice trial to familiarize themselves with the procedure. The test began with the participant away from the force platform and board, on a cue they were instructed to step on to the board, the participant was then allowed to stabilise prior to the double limb squat. They were instructed to perform the squat as slowly as possible to their maximum comfortable depth. Once they had reached their maximum angle, they were instructed to slowly return to the upright position.

Data Collection

Movement analysis data were collected using a ten camera Oqus system (Qualisys Medical AB, Gothenburg, Sweden), all movement data were collected at 100 Hz. Force data were collected using an AMTI force plate (Advanced Mechanical Technology, Inc. USA). The decline board was placed on the top of the force platform and the force platform was zeroed to allow for the extra weight of the board in accordance with previous work (Richards et al., 2008). Electromyographs (EMG) were collected from biceps femoris, rectus femoris, gastrocnemius and tibialis anterior using a DELSYS Bagnoli system (Delsys. Inc. USA). All analogue data, force and EMG, were collected at 2000 Hz.

Modelling of the lower limbs and joints

The segments of the lower limbs were modelled based on the calibrated anatomical systems technique (CAST) (Cappozzo et al., 1995). The anatomical landmarks used were the medial and lateral malleoli, medial and lateral femoral epicondyles, the greater trochanter, the anterior superior iliac spines of the pelvis and the posterior superior iliac spines of the pelvis. Clusters of 4 markers mounted on rigid plastic shells were attached to each segment. The knee and ankle joint centres were calculated as the median distance between the medial and lateral joint markers. The hip joint centre was calculated based on the regression equations developed by Bell et al. (1990).

Data Processing

The raw data were then exported to Visual 3D (C-Motion Inc. USA) for processing. The movement and force data were filtered using a fourth order low pass Butterworth filter with a cut off frequency of 6 Hz and 25 Hz, respectively. The EMG data were zeroed to remove any offset and then bandpass filtered with a highpass filter of 20 Hz and a lowpass filter of 500 Hz. For the calculation of angle specific EMG data were full wave rectified, then enveloped using a fourth order low pass Butterworth filter with a cut off frequency of 25 Hz. iEMG (integrated EMG) was calculated based on the rectified data. These were then normalized to the maximal observed contraction for each muscle for the different decline angles for each individual (Kellis and Baltzopoulos, 1996; Richards et al., 2008). Joint kinematics were calculated using a Cardan/Euler method with XYZ order of rotations. 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 ankle and knee joints using inverse dynamics methods.

A previously utilized algorithmic model was used to determine patellofemoral contact force and pressure (Ward and Powers, 2004). Patellofemoral joint contact force was estimated as a function of the knee flexion angle (fa) and knee extensor moment (ME) according to the biomechanical model described by Ho et al. (2012). Firstly, an effective moment arm of the quadriceps muscle (mq) was calculated as a function of the knee flexion angle using a non-linear equation, based on cadaveric information presented by van Eijden et al. (1986):

$$mq = 0.00008\ fa^3 - 0.013\ fa^2 + 0.28\ fa + 0.046$$

Quadriceps force (QF) was then calculated using the below formula:

$$QF = ME / mq$$

PFF was estimated using the QF and a constant (K):

$$PFF = QF\ K$$

The constant was described in relation to the fa using a curve fitting technique based on the non-linear equation described by Eijden et al. (1986):

$$K = (0.462 + 0.00147\ fa^2 - 0.0000384\ fa^2) / (1 - 0.0162\ fa + 0.000155\ fa^2 - 0.000000698\ fa^3)$$

PP (MPa) was calculated as a function of the PFF divided by the patellofemoral contact area. The contact area was described in accordance with the Ho et al. (2012) recommendations by fitting a second-order polynomial curve to the data of Powers et al. (1998) who documented patellofemoral contact areas at varying levels of knee flexion.

$$PP = PFF / \text{contact area}$$

To estimate patellar tendon kinetics, an additional predictive mechanism was used (Janssen et al., 2013). Patellar tendon force (PTF) was determined by dividing the ME by the tendon moment arm (PTMA). The moment arm was quantified as a function of the fa by fitting a 2nd order polynomial curve to the data of Herzog and Read (1993).

$$PTF = ME / PTMA$$

Statistical analysis

Repeated measures ANOVA with a post-hoc pairwise comparison with a least significant difference was conducted for the ankle, knee and hip movement and moments and for the iEMG values. *p*-values were reported comparing the different decline squats.

Results

Ankle and knee range of motion in the sagittal plane

There was a trend of increasing ankle range of movement with an increasing angle of declination (Figure 1, Table 1). The repeated measures ANOVA showed a significant difference between decline angles ($p = 0.001$). The pairwise comparison showed a significant difference between all decline angles with the exception of between 20 & 25 degrees (Table 2).

There was a trend of increasing knee range of movement with an increasing angle of declination (Figure 1, Table 1). The repeated measures ANOVA showed a significant difference between incline angles ($p = 0.001$). The pairwise comparison showed a significant difference between all incline angles with the exception of 20 & 25 degrees (Table 2).

Maximum ankle and knee moments in the sagittal plane

There was a trend of reducing the maximum ankle moment as the angle of declination increased (Figure 2). The repeated measures ANOVA showed a significant difference between decline angles ($p < 0.05$). The pairwise comparison showed significant differences between 0 & 10, 0 & 15, 0 & 20, 0 & 25, 5 & 25 and 10 & 25 degrees; however, no difference was seen between the higher angles, i.e. 15 & 20, 15 & 25 and 20 & 25, indicating a smaller effect on ankle moments at the larger decline angles (Table 2).

There was a trend of increasing the maximum knee moment with an increasing angle of declination (Figure 2). The repeated measures ANOVA showed a significant difference between decline angles ($p < 0.05$). The pairwise comparison showed a significant difference between all incline angles (Table 2).

iEMG for rectus femoris and tibialis anterior

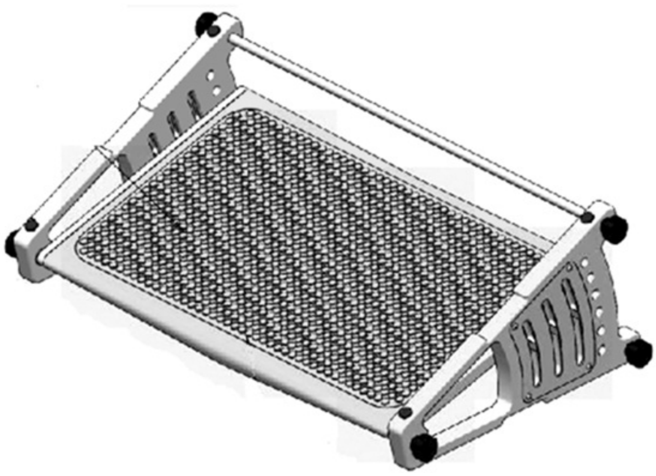
There was a trend of increasing maximum iEMG as the declination angle increased (Figure 3). The repeated measures ANOVA showed significant differences between decline angles ($p = 0.001$). The pairwise comparisons showed differences between 0 & 15, 0 & 20, 0 & 25, 5 & 25, 10 & 15 and 10 & 25 degrees (Table 3). There was a trend of reducing iEMG in tibialis anterior with increasing decline angles (Figure 3). The repeated measures ANOVA showed significant differences ($p = 0.001$). The pairwise comparisons showed significant differences for all pairs except 5 & 10, 10 & 15, 15 & 20, 15 & 25 and 20 & 25 degrees (Table 3).

Maximum iEMG biceps femoris and gastrocnemius

There was a trend showing an increase in the biceps femoris beyond 5 degrees as the angle of declination increased (Figure 4). However, the repeated measures ANOVA did not show a significant difference ($p = 0.318$) (Table 3). The trend seen in iEMG activity in gastrocnemius was that of little or no change up to 15 degrees of declination, after which a rise with the maximum activity occurred at 25 degrees (Figure 4). The repeated measures ANOVA showed no significant differences between decline angles ($p = 0.171$). However, the pairwise comparisons showed a significant increase in gastrocnemius activity between 15 & 25 degrees (Table 3).

Patellofemoral and patellar tendon kinetics

The repeated measures ANOVA for PFF showed a significant difference between decline angles ($p < 0.05$) (Figure 5). The pairwise comparison showed significant differences between 0 & 5, 0 & 20, 0 & 25 and 5 & 25 degrees; no further differences were observed (Table 4). The repeated measures ANOVA for PP showed a significant difference between decline angles ($p < 0.05$). The pairwise comparison showed significant differences between 0 & 10, 0 & 15, 0 & 20, 0 & 25, 5 & 10, 5 & 15, 5 & 20 and 5 & 25 degrees; no further differences were observed (Table 4). The repeated measures ANOVA for PTF showed a significant difference between decline angles ($p < 0.01$). The pairwise comparison showed significant differences between all angles with the exception of between 20 & 25 degrees (Table 4).



Picture 1

The Rehab Angel decline board

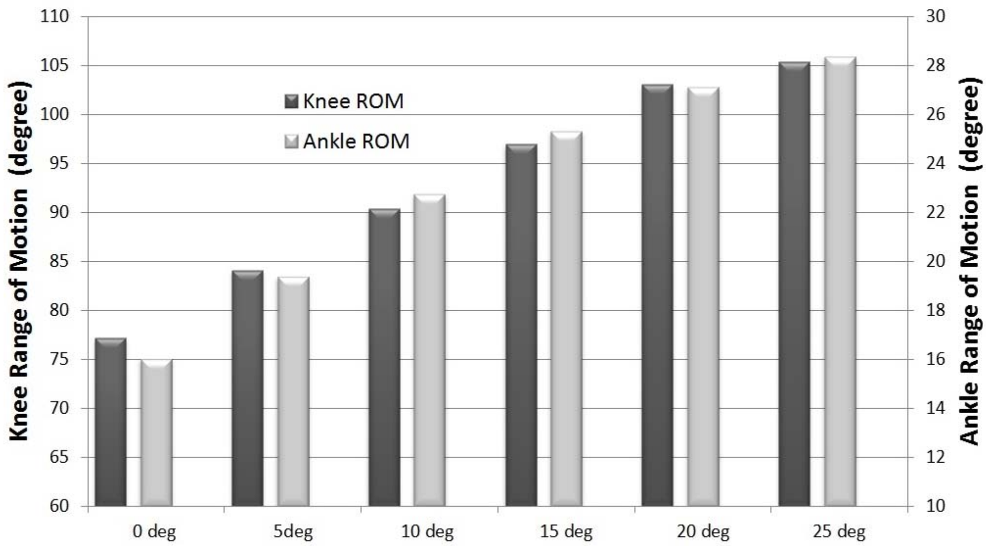


Figure 1

Knee and ankle range of motion during decline squats

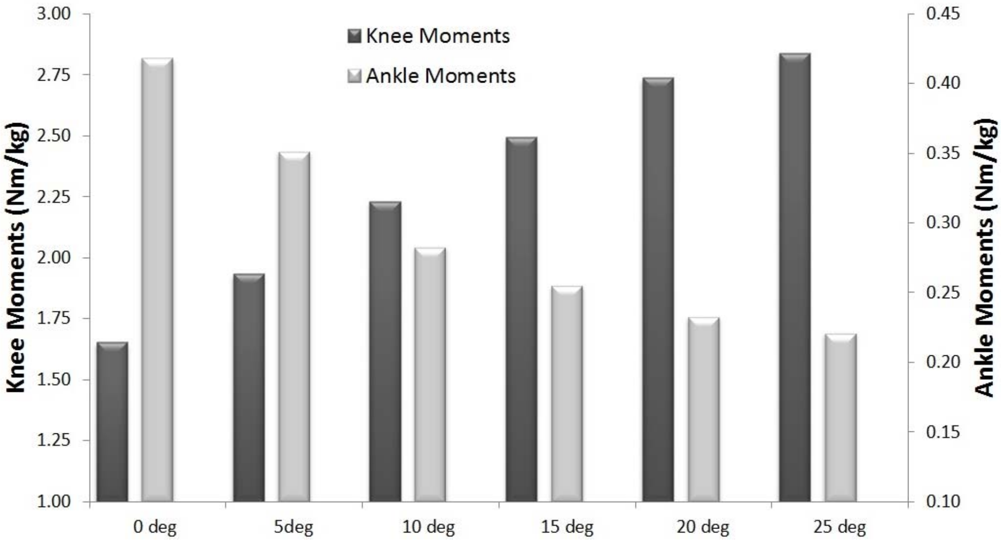


Figure 2

Knee and ankle moments during decline squats

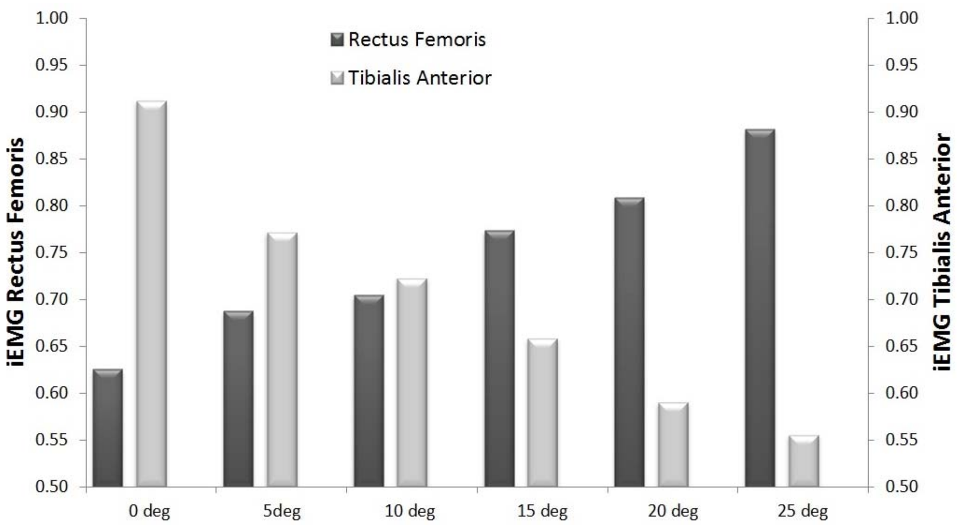


Figure 3

Rectus femoris and tibialis anterior activity during decline squats

Table 1

Means and standard deviations of joint angles, moments, iEMG and patellofemoral forces and pressure at the decline angles

Decline Angle	0	5	10	15	20	25
Ankle ROM Mean (sd)	16.0 (4.2)	19.4 (5.2)	22.8 (6.3)	25.3 (7.4)	27.1 (7.5)	28.4 (8.1)
Knee ROM Mean (sd)	77.2 (13.4)	84.1 (12.9)	90.4 (15.8)	97.0 (16.6)	103.1 (17.1)	105.4 (14.1)
Ankle Moments Mean (sd)	-0.42 (0.32)	-0.35 (0.33)	-0.28 (0.27)	-0.26 (0.31)	-0.23 (0.23)	-0.22 (0.28)
Knee Moments Mean (sd)	1.65 (0.66)	1.93 (0.68)	2.23 (0.76)	2.49 (0.75)	2.74 (0.73)	2.84 (0.70)
Rectus femoris Mean (sd)	0.63 (0.19)	0.69 (0.18)	0.70 (0.14)	0.77 (0.14)	0.81 (0.15)	0.88 (0.12)
Tibialis anterior Mean (sd)	0.91 (0.11)	0.77 (0.15)	0.72 (0.15)	0.66 (0.14)	0.59 (0.21)	0.56 (0.25)
Biceps femoris Mean (sd)	0.69 (0.20)	0.62 (0.15)	0.69 (0.13)	0.72 (0.11)	0.74 (0.17)	0.76 (0.19)
Gastrocnemius Mean (sd)	0.62 (0.29)	0.58 (0.21)	0.59 (0.20)	0.52 (0.19)	0.59 (0.19)	0.72 (0.19)
PFF (N) Mean (sd)	3366 (1516)	3702 (1508)	3705 (1513)	3792 (1427)	5065 (2920)	4505 (2222)
PP (MPa) Mean (sd)	6.46 (2.54)	6.86 (2.69)	7.40 (2.64)	7.73 (2.83)	9.34 (4.67)	8.47 (3.38)
PTF (N) Mean (sd)	3479 (1509)	4049 (1781)	4654 (2104)	5114 (2147)	5614 (2219)	5683 (2094)

Table 2

Mean difference (MD) and pairwise comparisons of joint angles and moments between the decline angles

Decline Angle		0 - 5	0 – 10	0 - 15	0 - 20	0 - 25	5 - 10	5 - 15	5 - 20
Ankle ROM	MD	-3.38*	-6.73*	-9.29*	-11.10*	-12.34*	-3.35*	-5.91*	-7.72*
	p	0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001
Knee ROM	MD	-6.93*	-13.22*	-19.77*	-25.87*	-28.18*	-6.29*	-12.84*	-18.94*
	p	0.001	>0.001	>0.001	>0.001	>0.001	0.005	0.001	>0.001
Ankle Moment	MD	-0.067	-0.135*	-0.163*	-0.185*	-0.197*	-0.068	-0.096	-0.119
	p	0.146	0.022	0.034	0.007	0.01	0.256	0.124	0.069
Knee Moment	MD	-0.281*	-0.580*	-0.841*	-1.086*	-1.186*	-0.299*	-0.560*	-0.805*
	p	0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001

Decline Angle		5-25	10-15	10-20	10-25	15-20	15-25	20-25
Ankle ROM	MD	-8.97*	-2.56*	-4.37*	-5.61*	-1.81*	-3.05*	-1.24
	p	>0.001	0.008	>0.001	>0.001	0.016	0.004	0.092
Knee ROM	MD	-21.25*	-6.55*	-12.65*	-14.96*	-6.10*	-8.41*	-2.31
	p	>0.001	0.001	>0.001	>0.001	0.003	0.004	0.297
Ankle Moment	MD	-0.130*	-0.028	-0.05	-0.062*	-0.022	-0.034	-0.011
	p	0.043	0.436	0.087	0.048	0.593	0.187	0.715
Knee Moment	MD	-0.905*	-0.261*	-0.506*	-0.606*	-0.245*	-0.345*	-0.100*
	p	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001	0.031

Table 3

Mean difference (MD) and pairwise comparisons of EMG activity between the decline angles

Decline Angle		0 - 5	0 - 10	0 – 15	0 - 20	0 - 25	5 - 10	5 - 15	5 - 20
Rectus femoris	MD	-0.062	-0.079	-0.148*	-0.183*	-0.257*	-0.017	-0.086	-0.121
	p	0.072	0.09	0.013	0.019	0.002	0.562	0.081	0.08
Tibialis anterior	MD	0.140*	0.189*	0.253*	0.322*	0.356*	0.048	0.113*	0.181*
	p	0.017	0.012	0.001	0.001	0.002	0.273	0.003	0.002
Biceps femoris	MD	0.068	0.000	-0.03	-0.055	-0.075	-0.067	-0.097	-0.123
	p	0.323	0.992	0.592	0.569	0.479	0.213	0.055	0.141
Gastroc nemius	MD	0.046	0.031	0.108	0.037	-0.099	-0.015	0.062	-0.009
	p	0.187	0.576	0.189	0.691	0.434	0.581	0.258	0.898

Decline Angle		5 - 25	10 - 15	10 - 20	10 - 25	15 - 20	15 - 25	20 - 25
Rectus femoris	MD	-0.195*	-0.069*	-0.104	-0.178*	-0.035	-0.109	-0.074
	p	0.006	0.046	0.084	0.004	0.579	0.059	0.193
Tibialis anterior	MD	0.216*	0.064	0.133*	0.168*	0.068	0.103	0.035
	p	0.006	0.088	0.002	0.013	0.199	0.094	0.519
Biceps femoris	MD	-0.143*	-0.03	-0.055	-0.076	-0.025	-0.046	-0.021
	p	0.044	0.38	0.449	0.375	0.639	0.43	0.694
Gastroc nemius	MD	-0.145	0.077	0.006	-0.13	-0.071	-0.207*	-0.136
	p	0.148	0.068	0.902	0.128	0.147	0.005	0.071

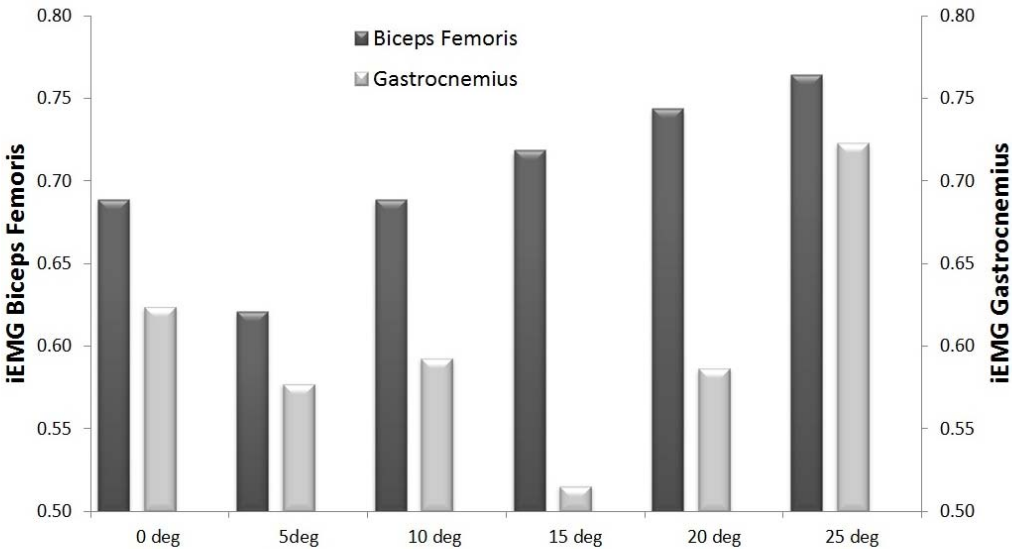


Figure 4

Biceps femoris and rectus femoris activity during decline squats

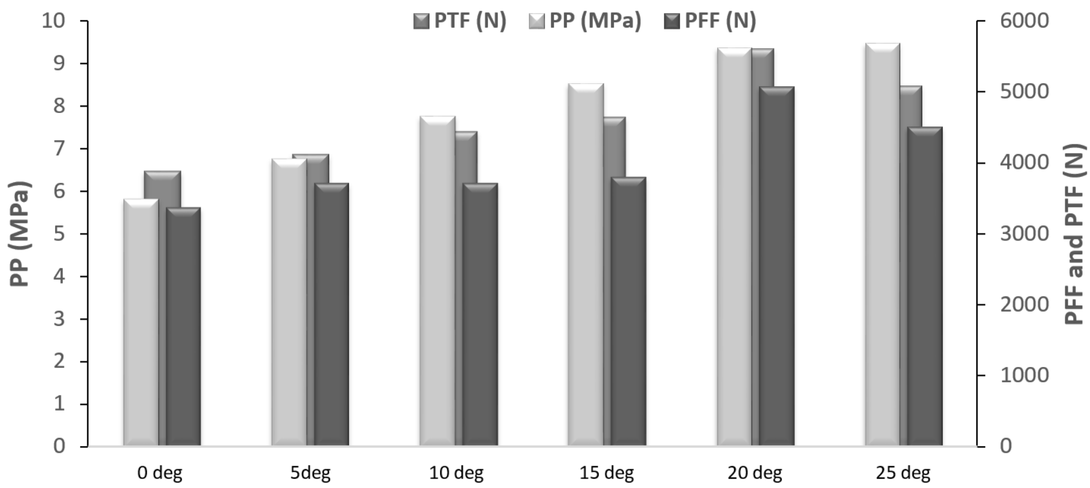


Figure 5

Biceps femoris and rectus femoris activity during decline squats

Table 4

Mean difference (MD) and pairwise comparisons of patellofemoral force, pressure and patellar tendon forces between the decline angles

Decline Angle		0 - 5	0 – 10	0 - 15	0 - 20	0 - 25	5 - 10	5 - 15	5 - 20
PFF (N)	MD	-335.8*	-336.6	-429.6	-1694.8*	-1134.4	-3.48	-91.12	-1367.4
	p	0.005	0.112	0.192	0.029	0.010	0.965	0.721	0.059
PP (MPa)	MD	-0.41	-0.94*	-1.26*	-2.85*	-2.03*	-0.56*	-0.87*	-2.46*
	p	0.281	0.013	0.025	0.028	0.016	0.006	0.007	0.024
PTF (N)	MD	-574.5*	-1173.1*	-1627.3*	-2142.2*	-2210.7*	-609.5*	-1061.3*	-1569.4*
	p	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001	>0.001

Decline Angle		5 - 25	10 – 15	10 - 20	10 - 25	15 - 20	15 - 25	20 - 25
PFF (N)	MD	-806.6*	-87.8	-1363.4	-802.3	-1274.6	422.9	562.1
	p	0.039	0.656	0.062	0.091	0.081	0.192	0.275
PP (MPa)	MD	-1.63*	-0.34	-1.95	-1.06	-1.62	-0.75	0.88
	p	0.016	0.166	0.075	0.084	0.076	0.111	0.179
PTF (N)	MD	-1639.9*	-463.0*	-956.7*	-1024.7*	-504.4*	-566.3*	-67.6
	p	>0.001	0.001	>0.001	>0.001	>0.001	>0.001	0.529

Discussion

As the decline angle increases, the ankle range of motion also increases with greater movement into plantarflexion. The ankle moments during a squat showed a significant decrease with an increase in the decline angle up to 20 degrees. This may be explained by the fact that as the angle of decline increases, the centre of pressure moves posteriorly from the forefoot towards the ankle joint. This has the effect of decreasing the moment arm about the ankle, therefore reducing the moment about the ankle; although there appears to be a limit to how low the ankle moment can go. The tibialis anterior muscle shows a reduction in activity as the decline angle increases which indicates less co-contraction is required as the incline angle increases. Both the reduction in moments and co-contraction would indicate a graduated offloading of the ankle joint. However, gastrocnemius activity increases as the decline angle rises above 15 degrees at the same time as tibialis anterior reduces. One explanation for this activity is that gastrocnemius, which is a two joint muscle, also plays a role in controlling and stabilising the knee as the decline angle exceeds 15 degrees.

The knee range of motion, the knee moments, PFF, PP PTF, all increase as the decline angle increases. It was particularly noteworthy that the PTF showed significant increases up to 20 degrees, but not further increase in the load was seen between 20 and 25 degrees. In addition, rectus femoris activity increases as the moment increases, all which would indicate a graduated increase in the load at the knee. However, biceps femoris activity increases significantly at a decline angle over 15 degrees which parallels gastrocnemius activity, both muscles acting as posterior stabilisers to the knee. This provides an active protective mechanism for anterior translation of the femur on the tibia, and therefore potentially protects the posterior cruciate ligament during the task. This indicates that as the angle of decline increases over 20 degrees, there appears to be an increase in the activity of the posterior muscles and not a decrease (Jonsson and Alfredson, 2005; Purdam et al., 2003). This finding is supported by previous squat studies (Frohm et al., 2007; Richards et al., 2008) which found an increase in gastrocnemius activity associated with increased inclination.

To date the decline squat has been used as an exercise to target the knee extensors (Alfredson et al., 1998; Jonsson and Alfredson, 2005; Purdam et al., 2003; Zwerver et al., 2007). The principles behind this rationale were fundamentally correct; however, the optimal angle had not been identified. This study aimed to provide further information to allow therapists to make more informed decisions when using squat exercises especially during the rehabilitation of PCL deficient knees. Fanelli et al. (2009) advocate early isometric and closed chain quadriceps exercises in the range of 0-60 degrees for grade 1 and 11 PCL injuries; these could be potentially enhanced by the use of a decline squat board at angles of 15-25 degrees. If the clinical reasoning for the test is to target the knee, then the data presented would suggest up to 20 degrees would have the maximum effect on the knee extensors and patellar tendon forces with the minimum effect about the ankle. However, a decline angle of 25 degrees would be justified if the aim of the rehabilitation program was to give a greater challenge to both the knee and the ankle.

In conclusion, this investigation indicates that using a graduated decline squat angle offers knee rehabilitation that allows a graduated increase in the load applied to the knee and graduated reduction in ankle moments and forces as the decline angle increases, with an optimum angle of between 15-20 degrees, which is less than the decline angles previously used in rehabilitation. Using a range of decline squat angles allows clinicians to be able to offer a controlled graduated rehabilitation environment for squatting tasks. This may also be useful in athletic training due to the greater work done by the quadriceps. This paper provides normative data to compare with individuals in recovery programs and has the potential to be used in physical conditioning and possible avoidance of overload.

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