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Double-bundle posterior cruciate ligament reconstruction: a biomechanical analysis of simulated early motion and partial and full weightbearing on common reconstruction grafts

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Abstract
Purpose The purpose of this study was to determine the biomechanical effects of simulated immediate motion and weightbearing during rehabilitation on different double-bundle posterior cruciate ligament reconstruction (DB-PCLR) graft options.
Methods Nine each of commercially prepared (allograft) Achilles tendon allografts, fresh-frozen (autograft) bone-patellar tendon-bone grafts, and fresh-frozen quadriceps tendon grafts were paired with commercially prepared anterior tibialis allografts, fresh-frozen semitendinosus grafts, and fresh-frozen semitendinosus grafts, respectively. Graft pairs were loaded to simulate early range of motion on a stationary bicycle, partial weightbearing (30 %), and full weightbearing.
Results Acquired laxity (displacement, mm) between graft pairs was not significantly different during simulated early range of motion. However, during simulated partial weightbearing, the median acquired laxity of the patellar tendon/semitendinosus pair (1.06 mm) was significantly less than that of the quadriceps tendon/semitendinosus (1.50 mm, $p = 0.01$) and Achilles/anterior tibialis (1.44 mm, $p = 0.003$) graft pairs. During simulated full weightbearing, significantly less acquired laxity was observed for the patellar tendon/semitendinosus graft pair (2.38 mm) compared to the Achilles/anterior tibialis pair (4.85 mm, $p = 0.04$), but a significant difference was not observed compared to the QT/semitendinosus graft pair (3.91 mm, n.s.). There were no significant differences in the ultimate loads between any of the graft pairs.
Conclusions Simulated early range of motion and early partial weightbearing did not result in clinically significant acquired graft laxity in common graft options utilized for DB-PCLR. However, simulated full weightbearing did result in clinically significant acquired graft laxity, and therefore, early rehabilitation protocols should avoid implementing full weightbearing that could contribute to graft failure.

Keywords Double-bundle posterior cruciate ligament reconstruction · Rehabilitation · Allograft · Autograft · Weightbearing · Range of motion

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Introduction

Historically, non-weightbearing and early immobilization in extension with limited prone range of motion post-operative rehabilitation protocols following posterior cruciate ligament (PCL) reconstructions have been advocated to protect the PCL graft from the posteriorly directed forces of gravity and the hamstrings [8, 36, 46]. Although these relatively conservative rehabilitation protocols have been implemented, the outcomes with regard to objective anteroposterior laxity with longer-term follow-up of single-bundle PCL reconstructions (SB-PCLRs) have revealed continued laxity compared to the contralateral knee of 4–6 mm in most studies [22, 26, 47]. Noting that a complete PCL tear is considered present with 8 mm of increased posterior tibial translation compared to the contralateral knee [24], it is concerning that between 50 and 75 % of this diagnostic measurement remains as residual PCL laxity. In an attempt to more accurately recreate native knee anatomy and kinematics and potentially decrease the likelihood of the development of unwanted anteroposterior laxity, anatomic double-bundle PCL reconstructions (DB-PCLRs) have been increasingly advocated [43, 51].

There is still some debate over the benefits of reconstructing both bundles of the PCL due to conflicting biomechanical [6, 11, 28, 38, 44, 50] and clinical evidence [11, 17, 48]. The roles of the two bundles of the PCL have been elucidated, and many authors have demonstrated that both the anterolateral bundle (ALB) and the posteromedial bundle (PMB) are functionally important and likely codominant to both anteroposterior stability and rotation throughout knee motion [19–21, 27, 51]. Additionally, Spiridonov et al. [43] have reported improved subjective outcomes and objective stability utilizing a DB-PCLR and a traditional rehabilitation protocol. Post-operatively, all patients were non-weightbearing for 6 weeks. Their physical therapy regimen emphasized immediate quadriceps activation and prone knee flexion to 90 degrees. All patients were also managed with a dynamic PCL functional brace post-operatively. Unlike traditional single-bundle techniques, concordance was predictably achieved in both subjective outcomes and objective stability. The potential benefits of immediate mobilization have also been demonstrated in the laboratory setting for various ligaments and showed improved mechanical and structural properties of the medial collateral ligament with mobilization [52, 53] and improved clinical outcomes following ACL reconstruction [23]. Moreover, the effects of a more aggressive post-operative rehabilitation protocol on a larger overall DB-PCLR graft volume that is more anatomic are unknown and merit further investigation.

The purpose of this study was twofold: (1) to determine whether a loading protocol designed to simulate an early range of motion and weightbearing rehabilitation protocol leads to acquired graft laxity and (2) to determine whether the laxity in the current, most commonly utilized reconstruction allograft options is comparable to readily available autograft options. It was hypothesized that there would be no clinically significant increase in the acquired laxity of either autograft or high-quality allograft options commonly used in DB-PCLRs due to simulated early knee range of motion and weightbearing.

Materials and methods

Allografts commonly used in PCL reconstructions were obtained from a commercially prepared source (AlloTrue, Allosource, Centennial, Colorado). Nine Achilles tendons (median age 57 years, range 49–61) and 9 anterior tibialis tendons (median age 51 years, range 21–65) were the allografts utilized in this study. Autografts potentially available for PCL reconstructions were harvested from fresh-frozen cadaveric specimens to simulate the use of autograft tissues. These grafts consisted of 9 bone-patellar tendon-bone (BTB) grafts (median age 56 years, range 48–65), 9 quadriceps tendon (QT) grafts (median age 61 years, range 48–65), and 18 semitendinosus grafts (median age 62 years, range 48–65).

Graft preparation

The calcaneal bone blocks of the Achilles grafts were trimmed to create bone plugs that were 11 mm in diameter and 25 mm in length, and the tendon was sized, if necessary, to pass through a tunnel diameter of 11 mm. Similarly, QT grafts and BTB grafts were trimmed to the same dimensions to fit through an 11-mm sizing block. All semitendinosus and anterior tibialis grafts were doubled and sized to pass through a 7-mm tunnel. Grafts sized to 11 mm and those sized to 7 mm were intended to replicate the anterolateral and posteromedial bundles, respectively, of a double-bundle PCL reconstruction construct.

Both the bone plugs and soft tissue ends of the various grafts were then fixed into rigid polyurethane foam blocks (Pacific Research Laboratories, Sawbones, Vashon Island, Washington) in clinically relevant reconstruction pairs (Table 1). Polyurethane foam blocks were chosen as a surrogate for human bone to allow for uniform modelling of the viscoelastic properties, strength, stiffness, and pullout testing of the chosen graft pairs to be conducted independent of potentially varying bone material properties and geometry shown to be present in cadaveric bone [12, 41]. Cancellous bone ranges in volumetric density between 0.09 and 1.26 g/cm^3 (5.6–78.7 pcf) [42], and cortical bone densities have been reported between 2.0 and 2.2 g/cm^3 (125–137 pcf) [7, 14]. Therefore, polyurethane blocks measuring 6 cm × 6 cm with a core of 20 pcf (0.32 g/cm^3) foam laminated with a 1.5-mm layer of 50 pcf (0.8 g/cm^3) were chosen to mimic the cortical and cancellous layers of human bone in a young athletic population with high bone mineral density [1, 2, 32, 33]. Both polyurethane densities chosen were greater than those previously reported in graft fixation research [4, 42] to replicate the population most likely to undergo PCLR [43] as opposed to a "worst-case scenario" of osteopenic bone.

Tunnel and graft sizes were chosen to be consistent with the technique for anatomic DB-PCLR described by Spiridonov et al. [43] and further validated by Wijdicks et al. [51]. Interference screw fixation was chosen for all interfaces to ensure consistency among graft pairs (Table 1).

Table 1 Graft pairs, bone tunnel sizes, and interference screw fixation sizes/types

Femoral notch						Tibial plateau	
Anterolateral bundle			Posteromedial bundle				
Graft	Tunnel size (mm)	Interference screw (mm)	Graft	Tunnel size (mm)	Interference screw (mm)	Tunnel size (mm)	Interference screw (mm)
Achilles allograft	11	7 × 25 titanium	Anterior tibialis allograft	7	7 × 25 biocomposite	12	11 × 28 biocomposite
Quadriceps tendon autograft	11	7 × 25 titanium	Semitendinosus autograft	7	7 × 25 biocomposite	12	11 × 28 biocomposite
Bone-patellar tendon-bone autograft	11	7 × 25 titanium	Semitendinosus autograft	7	7 × 25 biocomposite	12	9 × 25 titanium

The ALB and PMB were first fixed in their "femoral" tunnels with a bone bridge gap distance of 3 mm [3, 43], and then, a second polyurethane testing block ("tibial" side) was clamped at a separation distance of 35 mm to replicate the native length of the PCL [37]. A graft tensioning device (Arthrex, Naples, Florida) was then used to independently tension each graft to 20 N, and the grafts were secured to the "tibial" block. Independent tensioning at a constant distance was performed to ensure that both bundles experienced equivalent fixation tension. Screw sizes were determined through pilot testing to ensure a balance of adequate fixation and to minimize the risk of graft laceration/amputation and are noted in Table 1.

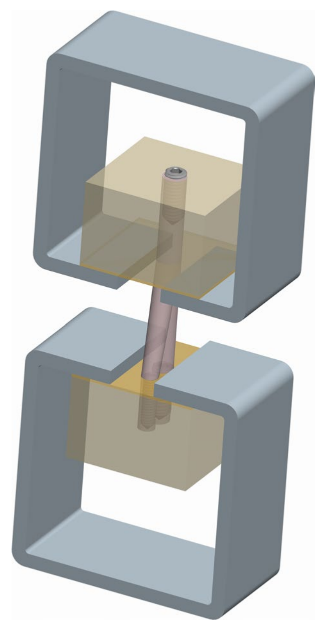

Fig. 1 Dynamic uniaxial tensile testing set-up. A schematic representation of the testing set-up showing grafts attached to polyurethane blocks by interference screw fixation and secured between custom loading fixtures which were attached to the actuator and base of a dynamic tensile testing machine

Testing protocol

After initial graft tensioning and fixation, the polyurethane blocks were secured between custom loading fixtures attached to the actuator of a dynamic tensile testing machine (Instron E10000, Norwood, Massachusetts; Fig. 1). Measurement error of the testing machine was certified by Instron to be less than or equal to ±0.01 mm and ±0.3 % of the indicated force. Eight of nine graft pairs of each group were then subjected to an identical progressive cyclic uniaxial loading protocol. The loading protocol was developed to simulate the maximal forces the PCL could encounter at time zero with both a progressive range of motion and partial (30 %) and full weightbearing. Cyclic load values were based on estimates of the maximal potential posterior tibiofemoral shear forces reported during various activities. The initial cyclic loading phase simulated the forces of immediate range of motion on the PCL as previously demonstrated on a cycle ergometer to simulate the use of a stationary bicycle [9]. Posterior tibiofemoral shear forces were estimated to reach 0.05 times body weight during use of a standardized cycle ergometer [9]; hence, a 50 N force was applied for 600 cycles at 1 Hz (0.05 times body weight force of a 70-kg adult for 10 min of standardized ergometer cycling at 120 W and 60 rpm).

Previous reports have also demonstrated that maximal posterior tibiofemoral shear forces may reach approximately 275 N during normal gait on level ground (0.4 times body weight force [30, 31] of a 70-kg adult). This estimate was used to establish the maximum load for partial (30 %) and full weightbearing, and a cyclic loading protocol with incrementally increasing forces followed

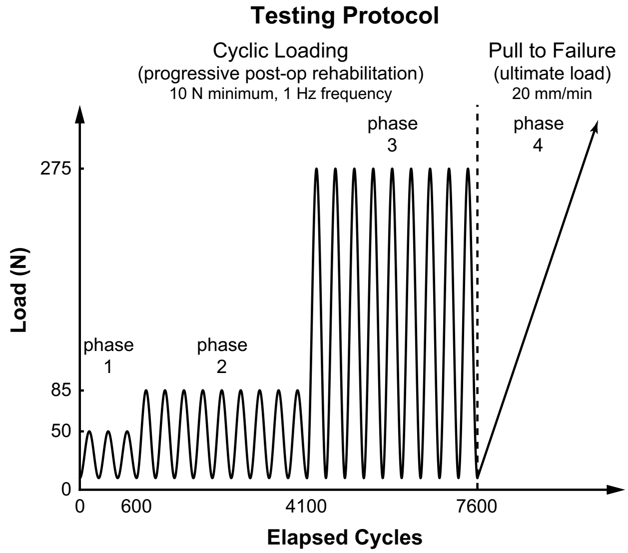

Fig. 2 Graphical representation of the cyclic loading protocol. Phase 1 simulated the forces of immediate range of motion on the PCL during stationary cycling: a 10–50 N force was applied for 600 cycles (i.e. 120 W and 60 rpm). Phase 2 simulated the forces of partial (30 %) weightbearing on the PCL during one post-operative day: a 10–85 N force was applied for 3500 cycles. Phase 3 simulated the forces of full weightbearing on the PCL during one post-operative day: a 10–275 N force was applied for 3500 cycles. Testing concluded with a pull to failure (Phase 4) at 20 mm/min where the elastic limit load (N) and ultimate load (N) of each of the graft pairs were determined

the cycle ergometer simulation. The number of cycles for the simulated partial and full weightbearing loading periods was based on pedometer data for the maximal steps a patient undergoing knee reconstruction might take daily in the first six post-operative weeks—approximately 3500 steps [40]. Therefore, paired graft constructs were subjected to cyclic loading over 3 phases at 1 Hz between 10 N and a progressively increasing maximum load that was incremented from 50 N (simulated range of motion cycling without resistance = phase 1) to 85 N (simulated partial weightbearing = phase 2) and to 275 N (simulated full weightbearing = phase 3) at cycle numbers 600, 4100, and 7600, respectively (Fig. 2). Displacement (mm), stiffness (N/mm), and the elastic limit (N; i.e. the load at which there is change from elastic to permanent deformation of the graft pairs [13]) for each of the constructs were continuously monitored and reported. After the first three loading phases, testing concluded with a pull to failure at 20 mm/min to determine the ultimate load (N) of each graft pair (phase 4). One additional non-cyclically loaded graft pair from each group was also pulled to failure at 20 mm/min to corroborate the load versus displacement curve of each construct.

Statistical analysis

Displacement, stiffness, elastic limit, and ultimate load were summarized for each construct group with medians, minima, and maxima. Each group of graft pairs was compared to one another, and differences were assessed using the Kruskal–Wallis test and Dunn's test for post hoc comparisons. Assuming 8 specimens per group and an $\alpha = 0.05$ for nonparametric, two-tailed, pairwise comparisons, an effect size of $d = 1.55$ is detectable with 80 % power. Adjusted p values <0.05 were considered significant. All statistical analyses were performed using IBM SPSS Statistics, version 20 (Armonk, New York).

Results

Displacement

Acquired laxity (displacement, mm) between graft pairs was not significantly different during simulated early range of motion (phase 1; concluded at cycle 600), and the median and maximal displacement for all of the pairs were less than 0.9 and 1.4 mm, respectively (Table 2). During simulated partial weightbearing (phase 2; concluded at cycle 4100), the median acquired laxity of the BTB/semitendinosus pair (1.06 mm) was significantly less than the QT/semitendinosus (1.50 mm, $p = 0.01$) and Achilles/anterior tibialis (1.44 mm, $p = 0.003$) graft pairs. The maximal acquired laxity during phase 2 for the BTB/semitendinosus, QT/semitendinosus, and Achilles/anterior tibialis pairs was 1.85, 2.50, and 2.37 mm, respectively.

During simulated full weightbearing (phase 3; concluded at cycle 7600), significantly less acquired laxity was observed with the BTB/semitendinosus graft pair (2.38 mm) compared to the Achilles/anterior tibialis pair (4.85 mm, $p = 0.04$), but a significant difference was not observed compared to the QT/semitendinosus graft pair (3.91 mm, n.s.) or between the QT/semitendinosus and the Achilles/anterior tibialis pairs ($p = 1.000$; Table 2). The maximal acquired laxity during phase 3 for the BTB/semitendinosus, QT/semitendinosus, and Achilles/anterior tibialis pairs was 3.44, 15.29, and 20.95 mm, respectively.

Stiffness and pull-to-failure measurements

All stiffness and pull-to-failure data are contained in Table 2. The BTB/semitendinosus construct was stiffer than the Achilles/anterior tibialis pair throughout the testing protocol and stiffer than the QT/semitendinosus pair at the start of phase 3 ($p = 0.03$) and during the pull to

Table 2 Biomechanical properties of common double-bundle posterior cruciate ligament reconstruction grafts

	Loading phase	Cycle	Median (min, max)			Group comparisons (p values)		
			BTB/ST (A)	QT/ST (B)	ACH/AT (C)	A–B	A–C	B–C
Cumulative displacement (mm)	Phase 1	1	0.46 [0.35, 0.62]	0.46 [0.35, 0.74]	0.61 [0.41, 0.78]	n.a.	n.a.	n.a.
		600	0.68 [0.51, 1.11]	0.82 [0.55, 1.33]	0.87 [0.72, 1.27]	n.a.	n.a.	n.a.
	Phase 2	4100	1.06 [0.81, 1.85]	1.50 [1.00, 2.50]	1.44 [1.13, 2.37]	0.006	0.003	n.s.
	Phase 3	7600	2.38 [1.81, 3.44]	3.91 [3.19, 15.29]	4.85 [2.77, 20.95]	n.s.	0.044	n.s.
	PTF	–	6.16 [4.94, 7.77]	11.13 [7.35, 33.41]	10.23 [6.11, 42.20]	0.006	0.024	n.s.
Stiffness (N/mm)	Phase 1	1	56 [46, 59]	50 [38, 58]	40 [37, 52]	n.s.	0.011	n.s.
		600	85 [75, 100]	78 [66, 91]	71 [52, 85]	n.s.	0.040	n.s.
	Phase 2	601	96 [84, 108]	86 [73, 101]	77 [62, 93]	n.s.	0.040	n.s.
		4100	111 [97, 129]	100 [78, 115]	96 [67, 113]	n.a.	n.a.	n.a.
	Phase 3	4101	163 [135, 183]	125 [85, 160]	124 [96, 146]	0.030	0.036	n.s.
		7600	183 [138, 194]	171 [126, 188]	158 [89, 184]	n.a.	n.a.	n.a.
	PTF	–	209 [140, 249]	134 [25, 186]	147 [2, 201]	0.016	0.014	n.s.
Elastic limit (N)	PTF	–	760 [339, 980]	629 [479, 816]	549 [421, 722]	n.a.	n.a.	n.a.
Ultimate load (N)	PTF	–	833 [361, 1133]	723 [499, 1031]	566 [489, 765]	n.a.	n.a.	n.a.

n.a., not applicable; this indicates that there was no significant difference among graft types based on overall Kruskal–Wallis test, and thus, no post hoc tests were performed

n.s., not significant; this indicates a non-significant post hoc group comparison (adjusted p value >0.05)

Group A—BTB/ST = Bone-patellar tendon-bone/semitendinosus "autograft" pair

Group B—QT/ST = Quadriceps tendon/Semitendinosus "autograft" pair

Group C—ACH/AT = Achilles tendon/Anterior tibialis "allograft" pair

PTF pull to failure

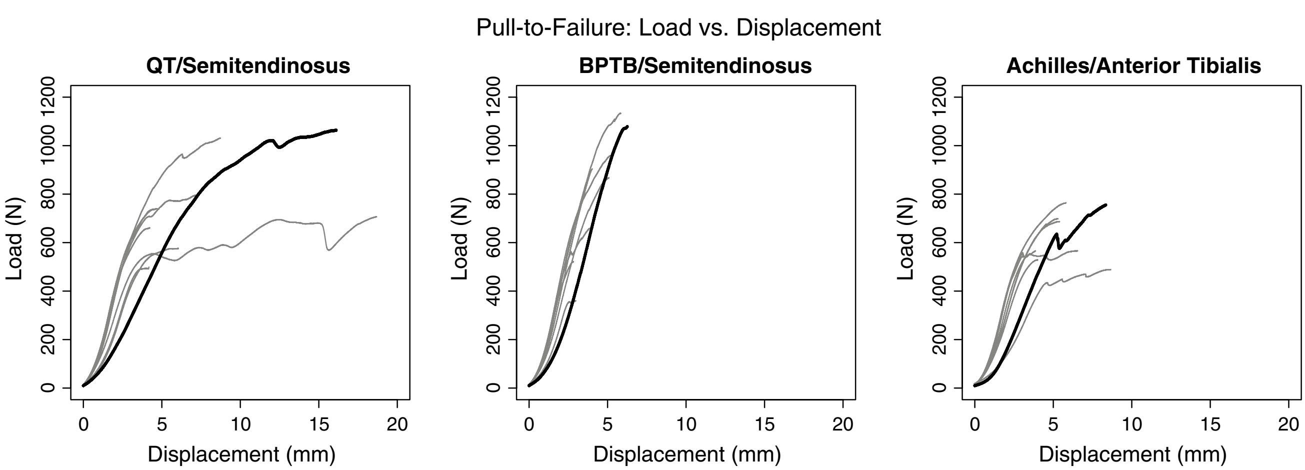

Fig. 3 Pull-to-failure load versus displacement curves of all graft pairs are shown. The *grey curves* (8/plot) represent graft pairs that were cyclically loaded and then pulled to failure. The *black curves* (1/plot) represent non-cyclically loaded graft pairs that were only pulled to failure

failure ($p = 0.02$). There were no significant differences in the elastic limits or ultimate loads between any of the graft pairs. Furthermore, the elastic limits of all the graft pairs were greater than the forces applied to simulate early range of motion activities including cycle ergometry (50 N; phase 1), partial weightbearing (85 N; phase 2), and full weightbearing (275 N) (phase 3; Fig. 3). The non-cyclically loaded graft pairs of each type that were only subjected to pull-to-failure testing also demonstrated similar elastic modulus curves to the cyclically loaded constructs (Fig. 3). The mechanisms of failure and frequencies are given in Table 3.

Table 3 Mechanism of failure for each specimen

Graft pair	Location of graft failure	Mechanism of graft failure
Quadriceps/semitendinosus		
	Tibial interface	
		Semitendinosus 2/9
		Both 4/9
	Femoral interface	
		Semitendinosus 1/9
		Both 1/9
	Midsubstance	
		Quadriceps 1/9
Bone-patellar tendon-bone/semitendinosus		
	Tibial interface	
		BTB 1/9
		Both 1/9
	Femoral interface	
		BTB 5/9
		Semitendinosus 1/9
	Polyurethane block breakage	
		1/9
Achilles/anterior tibialis		
	Tibial interface	
		Achilles 1/9
		Anterior tibialis 1/9
		Both 1/9
	Femoral interface	
		Achilles 4/9
		Anterior tibialis 1/9
	Midsubstance	
		Both 1/9

Discussion

The most important finding of this study was that common DB-PCLR grafts did not acquire clinically significant laxity as a result of a simulated rehabilitation protocol representative of early range of motion on a stationary bicycle and partial weightbearing. In contrast, during simulated full weightbearing, clinically significant differences in the median and maximal acquired laxity were observed between graft options. These findings biomechanically validate the caution implemented with traditional rehabilitation protocols that avoid immediate full weightbearing following DB-PCLR; nonetheless, partial weightbearing may be a safe alternative for early rehabilitation. The BTB/semitendinosus graft pair exhibited the least acquired graft laxity under all loading conditions including simulated full weightbearing. However, during loading phases 1 and 2 (simulated early range of motion on a stationary bicycle and partial weightbearing, respectively), significant differences between the graft options were either not observed or not clinically significant. Additionally, there was no difference in the ultimate loads between the allograft (e.g. Achilles/anterior tibialis) and autograft (e.g. BTB/semitendinosus and QT/semitendinosus) construct options.

The in vitro tensile properties of single-bundle graft options for cruciate ligament reconstruction have been well documented [15, 34, 54, 55], while the tensile properties of common graft construct combinations used for DB-PCL reconstruction have not. In this study, the ultimate failure load of each construct (lowest median = 566 N) was much greater than the forces encountered with weightbearing and range of motion (full weightbearing ~275 N [29, 30]). Additionally, there were several consistent findings with pull-to-failure testing: the graft pairs most commonly failed at the bone–graft interface, the BTB/semitendinosus graft pair demonstrated the greatest stiffness, and the anticipated ultimate loads of the graft constructs (based on previous studies which reported tensile properties of the graft bundles in isolation [15, 34, 54, 55]) were not consistently reached. This likely indicates that interference screw fixation was superior for grafts containing bone plugs versus soft tissue grafts and that partial graft laceration in this model of dense bone may have been a contributing factor to graft failure and slippage, as previously reported [39, 49].

The level of evidence supporting different recommendations for rehabilitation following PCLR is lacking rigorous scientific basis. Rehabilitation protocols have largely been based on level 4 and expert opinion data following SB-PCLRs [8, 10, 25, 36] and on the premise that early range of motion is more likely to be associated with inferior stability outcomes. However, many authors have implemented these conservative immobilization protocols, and suboptimal stability has still been reported [26, 47]. Consequently, others have reported the requirement for manipulation under anaesthesia and/or lysis of adhesions for stiffness in patients utilizing post-operative immobilization [29]. The potential benefits of immediate mobilization have been demonstrated in the laboratory setting for various ligaments and showed improved mechanical and structural properties of the medial collateral ligament with mobilization [52, 53] and improved clinical outcomes following ACL reconstruction [23]. Although the dynamic function of the quadriceps musculature is protective of the healing PCL graft [36, 46], quadriceps inhibition is common following knee trauma and surgery [16]. Rehabilitation exercises that result in earlier reversal of quadriceps inhibition are potentially desirable assuming concomitant graft stretching can be avoided. The results of the current study demonstrate the cyclic forces placed on a PCL graft simulating early range of motion on a stationary bicycle and partial weightbearing do not result in clinically relevant laxity (displacement). Therefore, a more progressive but protected early rehabilitation protocol following DB-PCLR is not likely to result in additional joint laxity [35, 43] and may also lead to improved patient outcomes [23].

This study was not without limitations. A simulated in vitro biomechanical testing model substituting polyurethane foam for bone was utilized. The polyurethane foam was chosen because it was believed to more consistently mimic the cortical and cancellous layers of human bone in a young athletic population with high bone mineral densities [1, 2, 32, 33]. While the synthetic foam was successful in providing homogeneity of material characteristics, the results may not translate perfectly to the time zero in vivo environment. Furthermore, the results of this study are representative of a time zero, in vitro biomechanical model, and the effect of biological healing was correspondingly unable to be studied. Additionally, uniaxial testing is unable to simulate the 6 degrees of freedom of natural knee kinematics; therefore, the results obtained by this study may not yield a complete understanding of the effects of an aggressive rehabilitation protocol on the grafts in patients. Although the constructs were not preconditioned prior to testing, sub-millimetre (~0.5 mm) elongation was observed during the first cycle of loading for all groups and therefore likely did not affect the cyclic elongation behaviour of the constructs. In addition to the inability to load the grafts at different knee flexion angles, tissue–tunnel interactions throughout the flexion arc are also not accounted for, especially the potential for PCL graft abrasion by the tibial tunnel aperture, or the so-called killer turn [5, 18, 45]. Furthermore, a uniform method of PCLR graft fixation (interference screws) was utilized. As such, the use of soft tissue bone staples or fixation screws and washers for the soft tissue grafts could have resulted in different displacement and strength measurements. Nevertheless, this study presents a detailed and consistent biomechanical evaluation of common allograft and autograft options for DB-PCLR.

Conclusions

This study demonstrated that incrementally increased cyclic forces of simulated early range of motion on a stationary bicycle and partial weightbearing were unlikely to lead to acquired laxity in common graft pairs used for DB-PCL reconstruction. Additionally, a clinically relevant difference did not exist between the tensile properties of simulated common autograft and high-quality commercially prepared allograft PCL constructs at cyclic loads up to partial weightbearing. However, simulation of immediate full weightbearing following DB-PCL reconstruction may lead to clinically significant acquired graft laxity; therefore, early rehabilitation protocols should avoid implementing full weightbearing. In order to examine and improve early outcomes following PCL reconstruction, a prospective randomized trial of a traditional rehabilitation protocol versus an early range of motion and progressive weightbearing rehabilitation program is recommended.

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