Knee biomechanics of the dynamic squat exercise

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ABSTRACT

ESCAMILLA, R. F. Knee biomechanics of the dynamic squat exercise. *Med. Sci. Sports Exerc.*, Vol. 33, No. 1, 2001, pp. 127–141. **Purpose:** Because a strong and stable knee is paramount to an athlete's or patient's success, an understanding of knee biomechanics while performing the squat is helpful to therapists, trainers, sports medicine physicians, researchers, coaches, and athletes who are interested in closed kinetic chain exercises, knee rehabilitation, and training for sport. The purpose of this review was to examine knee biomechanics during the dynamic squat exercise. **Methods:** Tibiofemoral shear and compressive forces, patellofemoral compressive force, knee muscle activity, and knee stability were reviewed and discussed relative to athletic performance, injury potential, and rehabilitation. **Results:** Low to moderate posterior shear forces, restrained primarily by the posterior cruciate ligament (PCL), were generated throughout the squat for all knee flexion angles. Low anterior shear forces, restrained primarily by the anterior cruciate ligament (ACL), were generated between 0 and 60° knee flexion. Patellofemoral compressive forces and tibiofemoral compressive and shear forces progressively increased as the knees flexed and decreased as the knees extended, reaching peak values near maximum knee flexion. Hence, training the squat in the functional range between 0 and 50° knee flexion may be appropriate for many knee rehabilitation patients, because knee forces were minimum in the functional range. Quadriceps, hamstrings, and gastrocnemius activity generally increased as knee flexion increased, which supports athletes with healthy knees performing the parallel squat (thighs parallel to ground at maximum knee flexion) between 0 and 100° knee flexion. Furthermore, it was demonstrated that the parallel squat was not injurious to the healthy knee. **Conclusions:** The squat was shown to be an effective exercise to employ during cruciate ligament or patellofemoral rehabilitation. For athletes with healthy knees, performing the parallel squat is recommended over the deep squat, because injury potential to the menisci and cruciate and collateral ligaments may increase with the deep squat. The squat does not compromise knee stability, and can enhance stability if performed correctly. Finally, the squat can be effective in developing hip, knee, and ankle musculature, because moderate to high quadriceps, hamstrings, and gastrocnemius activity were produced during the squat.
Key Words: TIBIOFEMORAL, PATELLOFEMORAL, SHEAR FORCE, COMPRESSIVE FORCE, ANTERIOR CRUCIATE LIGAMENT, POSTERIOR CRUCIATE LIGAMENT, MUSCLE ACTIVITY, QUADRICEPS, HAMSTRINGS, GASTROCNEMIUS, KNEE STABILITY

The dynamic squat exercise is an integral part of strength and conditioning programs for many sports that require high levels of strength and power, such as football, track and field, powerlifting, and Olympic weightlifting. The squat primarily strengthens hip, thigh, and back musculature, which are very important muscles in running, jumping, and lifting. It is commonly believed among athletes and coaches that the squat enhances athletic performance and minimizes injury potential. Because the dynamic squat has been classified as a closed kinetic chain exercise (17,58,65,71), it is also appropriate and commonly used in knee rehabilitation settings. Several studies have demonstrated the favorable use of the squat exercise during knee rehabilitation (21,32,39,47,52,58,71), such as after cruciate ligament reconstructive surgery. Consequently, an understanding of knee biomechanics during the squat is helpful to therapists, trainers, sports medicine physicians, researchers, coaches, and athletes who are interested in closed kinetic chain exercises, knee rehabilitation, and training for sport.

The bodyweight (BW) squat, barbell squat, and machine squat are the most common methods employed while performing the dynamic squat in training and rehabilitation. The squat begins with the individual in the upright position with the knees and hips fully extended. The individual then squats down in a continuous motion until a desired squat depth is obtained and then in a continuous motion ascends back to the upright position. The barbell squat is performed with the barbell across the back (back squat) slightly above (high bar squat) or below (low bar squat) the level of the acromion, or with the barbell held in front of the chest approximately at the level of the clavicles (front squat). The back squat is usually preferred and performed by most athletes in sport, although the front squat is commonly performed by bodybuilders and Olympic weightlifters. The front squat is simulated during the clean and jerk exercise by Olympic weightlifters. The machine squat is also performed in training and rehabilitation. The machine squat is typically performed using a special barbell that moves within a fixed groove (e.g., the Smith Machine squat) or by using padded

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level arms that are positioned on top of the shoulders (2). The squat can be performed with varying degrees of knee flexion, such as the half squat or full squat. The half squat involves squatting down until the thighs are parallel with the ground with approximately 0–100° knee flexion. The deep squat involves squatting down as far as possible until the posterior thighs and legs make contact with each other. The half squat is typically preferred and recommended over the full squat for athletes in training or rehabilitation patients (9,28).

An exhaustive search of MEDLINE (1966 to April 2000) and SPORTDiscus (1949 to April 2000) research databases was conducted to identify scientific literature relative to the biomechanics of the dynamic squat. The search strategy employed initially involved examining all fields containing "squat" and all fields containing "squat" AND "biomechanics." Because the dynamic squat is the focus of this review, all literature relative to the isometric squat was excluded from the search. In addition, only studies published in peer-reviewed scientific journals, or published as conference proceedings/abstracts from scientific conferences, were considered for inclusion in this review. From the research database searches, it was determined that barbell squat, the isotonic machine squat, and BW squat comprised the squat biomechanics literature. Most studies that examined the biomechanics of the dynamic squat have focused on knee biomechanics. Therefore, this review was limited to scientific studies that quantified knee biomechanics during the barbell, machine, and BW dynamic squat exercises. Dynamic squat studies that quantified knee biomechanics have primarily focused on three major areas: 1) knee forces comprising tibiofemoral shear force, tibiofemoral compressive force, and patellofemoral compressive force; 2) knee muscle activity from the quadriceps, hamstrings, and gastrocnemius; and 3) anteroposterior and mediolateral knee stability. An understanding of knee biomechanics during the squat is important because a strong and stable knee is paramount to an athlete's success or a patient's rehabilitation. Therefore, the purpose of this review was to examine knee forces, knee muscle activity, and knee stability during the dynamic squat exercise.

Tibiofemoral Shear and Compressive Forces

Excessive tibiofemoral shear forces can be injurious to the cruciate ligaments, whereas excessive tibiofemoral compressive forces can be deleterious to the menisci and articular cartilage. To date, there are 11 known studies that have quantified tibiofemoral shear or compressive forces during the dynamic squat (2,3,5,13,17,18,20,43,58,60,65). Eight of these studies performed the barbell squat with an external load (2,3,17,18,20,43,58,65), whereas two studies performed the BW squat (13,60). One additional study quantified ACL strain *in vivo* while performing the BW squat along with very light resistance from an elastic cord (5). All squats were performed so at maximum knee flexion (0° knee flexion defined as full knee extension) the thighs were parallel or below parallel with the ground. Comparing tibiofemoral compressive and shear forces among these studies is difficult, because methodologies varied among studies. Eight studies developed mathematical knee models that quantified both external (e.g., gravity, ground reaction) and internal (e.g., muscle, bone, ligament) forces (3,13,17,18,20,43,60,65), whereas the remaining two studies employed knee models that quantified external forces only (2,58). To quantify the actual shear and compressive forces across the articulating surface of the knee, knee muscle forces must be determined. The primary muscles that cross the knee are the quadriceps, hamstrings, and gastrocnemius, which comprise approximately 98% of the total cross-sectional area of all knee musculature (64). When the quadriceps, hamstrings, and gastrocnemius contract, they produce additional compressive and shear force components within the knee. It has been demonstrated that during a maximum voluntary contraction of the quadriceps the force generated ranges from 2000 to 8000 N, depending on knee flexion angle (63). Consequently, knee models that quantified both external and muscle forces should display greater tibiofemoral compressive and shear forces compared with knee models that quantified external forces only. However, developing mathematical models that estimate knee muscle and ligament forces can be difficult, and potentially inaccurate, depending on the variables measured and the methodology employed. These models often estimate muscle force from electromyographic (EMG) data, which frequently do not strongly correlate with muscle forces.

Throughout this manuscript, anterior shear forces will be defined as forces restrained primarily by the ACL, and posterior shear forces will be defined as forces restrained primarily by the PCL. This seems reasonable, because Butler et al. (7) have reported that the ACL provides 86% of the total restraining force to anterior drawer, and the PCL provides 95% of the total restraining force to posterior drawer.

A biomechanical comparison of the studies that quantified shear or compressive forces is shown in Table 1. All eight studies that specified shear force direction reported low to moderate posterior shear forces and PCL loading during the squat, although Nisell and Ekholm (43), Hattin et al. (20), and Toutoungi et al. (60) did show minimum anterior shear forces and ACL loading between 0 and 60° knee flexion. Because the loads lifted varied greatly among studies, mean peak shear and compressive force were normalized and expressed as a percent of the sum of BW and load lifted. Although normalized results vary, values from Dahlkvist et al. (13) and Toutoungi et al. (60) appear inordinately high, especially because the subjects from these two studies performed the BW squat with no external loads. Discounting these values, normalized peak posterior shear forces ranged from 29 to 99%, normalized peak anterior forces ranged from 4 to 14%, and normalized compressive force ranged from 54 to 367%.

Because the ultimate strength of the PCL has been estimated up to 4000 N for young active people (50), the peak posterior shear forces observed during the squat near maximum knee flexion (between 295 and 2704 N from Table 1) are probably not of great enough magnitude to be injurious to the healthy PCL. However, individuals with an injured or reconstructed PCL may want to avoid performing the squat

TABLE 1. Comparison of studies that quantified knee forces during the dynamic squat.

Squat reference	Forces included in knee model	No. of Subjects	Mean Height (cm)	Mean Age (yr)	Mean Body Weight (N)	Mean load lifted (N)	Knee flexion range (°)	Shear force direction acting on tibia	Mean peak tibiofemoral shear force (N) at knee angle (°)	Normalized mean peak tibiofemoral shear force % (BW + load)	Mean peak tibiofemoral compressive force (N) at knee angle (°)	Normalized mean peak tibiofemoral compressive force % (BW + load)	Mean peak patellofemoral compressive force (N) at knee angle (°)	Normalized mean peak patellofemoral compressive force % (BW + load)
Stuart et al. (58)	External	6	181 ± 8	27 ± 5	798 ± 76	223 ± 0	0–90	Posterior	295 ± 32 at 93	29 ± 3	550 ± 50 at 80	54 ± 5	—	—
Andrews et al. (2)	External	3	?	?	?	?	0–110*	?	1000* ± ? at 110*	?	?	?	—	—
Ariel (3)	External and internal	12	182 ± ?	21–25	888 ± ?	1982 ± ?	0–117*	?	1593 ± 106 at 48	56 ± 4	7928 ± 1965 at 106	276 ± 69	—	—
Escamilla et al. (17)	External and internal	12	177 ± 9	29 ± 6	912 ± 145	1437 ± 383	0–95	Posterior	1868 ± 878 at 63	80 ± 37	3134 ± 1040 at 53	133 ± 44	4548 ± 1395 at 85	194 ± 59
Escamilla et al. (18)	External and internal	12	177 ± 8	30 ± 7	917 ± 137	1309 ± 363	0–95	Posterior	2212 ± 801 at 76	99 ± 36	3428 ± 838 at 65	154 ± 38	4674 ± 1195 at 82	210 ± 54
Wilk et al. (65)	External and internal	10	177 ± 9	30 ± 6	912 ± 137	1442 ± 383	0–102	Posterior	1783 ± 634 at 90	76 ± 27	6139 ± 1709 at 91	261 ± 73	—	—
Dahlkvist et al. (13)	External and internal	6	180 ± 5	21 ± 1	732 ± 79	0	0–140*	Posterior	2652 ± 290 at 120*	362 ± 40	4018 ± 1230 at 120*	549 ± 168	5455 ± 260 at 130	745 ± 36
Toutoungi et al. (60)	External and internal	8	?	29 ± ?	765 ± ?	0	0–100	Posterior Anterior	2704 ± 805 at 100 28 ± 36 at 0–50	353 ± 105 4 ± 5	—	—	—	—
Nisell and Ekholm (43)	External and internal	3	173 ± 9	27 ± 7	932 ± 177	2453	0–130*	Posterior Anterior	1800* at 130* 500* at 20*	51* 14*	7000* at 130*	198*	6750* at 130*	191*
Hattin et al. (20)	External and Internal	10	178 ± 5	23 ± 2	790 ± 109	339 ± 64	0–90*	Posterior Anterior	751 ± 585 at 90 150* at 50*	67 ± 52 13*	4147 ± 1375 at 90*	367 ± 122	—	—
Wretenberg et al. (68)	External and internal	8 WL	177 ± 8	19 ± 3	800 ± 108	650 ± 176	0–120	—	—	—	—	—	4700 ± 590	324 ± 41
		6 PL	171 ± 10	31 ± 3	856 ± 200	981 ± 134							3300 ± 1700	180 ± 93
Reilly and Marten (51)	External and internal	3	?	27 ± 3	834	0	0–130*	—	—	—	—	—	6377 ± 130*	765

* value estimated from graphs and text.
? value not stated.
— variable not measured.
WL, weightlifters; PL, powerlifters; BW, bodyweight.

at knee flexion angles higher than 50–60°, because posterior shear forces increase as knee flexion increases. Anterior shear forces were generated during the squat only between 0 and 60° knee flexion, with peak forces between 28 and 500 N (Table 1). Therefore, performing the squat should not be injurious to the healthy ACL, because two independent studies have calculated its ultimate failure load to be between 1725 and 2160 N (45,67). The greater strength of the PCL compared with the ACL is primarily due to a 20–50% greater cross-sectional area (19). Because only minimum anterior shear forces were generated during the squat, the squat may also be a safe and effective rehabilitation exercise to perform for those who wish to minimize tensile loading of the ACL (e.g., after ACL reconstruction), assuming the PCL and other knee structures are healthy. Furthermore, squatting with increased forward trunk tilt and hip flexion has been shown eliminate ACL stress (47), which is largely due to an increase in hamstring activity (73, 74).

Peak compressive forces during the squat ranged from 550 to 7928 N (Table 1). Unfortunately, it is currently unknown at what magnitude compressive force becomes injurious to knee structures, such as menisci and articular cartilage. Excessive loading of the menisci and articular cartilage can lead to degenerative changes. However, compressive forces have been demonstrated to be an important factor in knee stabilization by resisting shear forces and minimizing tibia translation relative to the femur (23, 33, 53, 72).

Studies have shown (35, 36) that expert squatters, such as experienced powerlifters, perform better and have more favorable kinematics (more erect trunk, less horizontal hip and bar displacement, and less vertical bar velocity during the descent) and kinetics (less trunk torques and greater extensor dominant thigh torques) compared with novice squatters. Escamilla et al. (17) used powerlifters and bodybuilders experienced in performing the barbell squat in order to examine knee biomechanics while squatting (Table 1). This population was chosen because they were considered experts in knowing how to correctly perform the squat exercise. Each subject performed three repetitions with their 12 repetition maximum (12 RM) load. Four video cameras collected 60-Hz kinematic data, while two force platforms were utilized to collect 960-Hz kinetic data. EMG, inverse dynamics, mathematical knee modeling, and computer optimization techniques were employed to estimate internal muscle and ligamentous forces (76). Muscle forces (F_{mi}) from the quadriceps, hamstrings, and gastrocnemius were estimated from the equation $F_{mi} = {}_{ci}k_iA_{imi}(EMG_i/MVC_I)$, where c_i was a weight factor adjusted in a computer optimization program to minimize errors in muscle force estimates, k_i was a muscle force-length factor as a function of knee and hip flexion, A_i was the physiological cross sectional area (PCSA) of the ith muscle, $_{mi}$ was the maximum voluntary isometric contraction force (MVIC) per PSCA, and EMG_i and MVC_i were EMG window averages during the squat and MVIC, respectively. Results indicated that compressive and PCL forces progressively increased as the knees flexed and decreased as the knees extended, which is in agreement with several other squat studies (13,18,20,43,58,60,65). Forces were slightly greater during the ascent compared with the descent. Although these forces were estimated and not measured directly *in vivo*, it is interesting that there were no ACL tensile forces (i.e., anterior shear forces) measured throughout the squat exercise. Several studies have also reported no ACL tensile forces during the squat (13,18,58,65). The absence of ACL forces during the squat may in part be due to moderate hamstring activity, because several studies have demonstrated that the hamstrings help unload the ACL by producing a posterior directed force to the leg throughout the knee movement (4,14–16,31,39,46–48,73,74). Quadriceps activity also affects cruciate ligament strain. Quadriceps force, via the patella tendon, exerts an anterior directed force on the leg when the knee is flexed less than approximately 50–60°, and a posterior directed force when the knee is flexed greater than approximately 50–60° (8,22,56). When posterior directed shear forces acting on the leg exceed anterior directed shear forces, the net result will be a posterior shear force, which is restrained primarily by the PCL.

Escamilla et al. (18) compared the effects of technique variations (stance width and foot angle) on shear and compressive forces. Using a 12-RM lifting intensity during the barbell squat, their subjects performed a narrow stance (distance between medial malleoli was equal to distance between anterior superior iliac spines) and wide stance (twice the narrow stance distance) squat with the feet parallel (i.e., both feet pointing straight ahead) and with the feet turned outwards 30°. There were no significant differences in compressive forces and PCL tensile forces between the two foot angle conditions. Comparing the two stance widths, there were no significant differences in PCL tensile forces between the narrow stance and wide stance. However, the wide stance generated 15–16% significantly greater compressive forces than the narrow stance between 19 and 83° knee flexion during the squat descent, and between 59 and 89° knee flexion during the squat ascent. In addition, the squat ascent generated 57–66% greater PCL tensile forces than the squat descent between 27 and 95° knee flexion, the squat descent generated 9–10% greater compressive forces than the squat ascent between 71 and 95° knee flexion, and the squat ascent generated 17% greater compressive forces than the squat descent between 19 and 61° knee flexion. Peak PCL forces were 30–40% greater during the ascent compared with the descent. The authors suggested that the significantly greater compressive forces generated during the wide stance compared with the narrow stance, and between the squat ascent and squat descent, might help protect the knee against excessive shear forces (23,33,53,72).

Nisell and Ekholm (43) did a two-part study. They first examined knee joint loads in world-class powerlifters during the ascent portion of the powerlifting barbell squat (Table 1). Each subject was filmed (4 Hz) with a one-camera motion system in the sagittal plane of movement. By using quasi-static two-dimensional biomechanical knee models by Nisell (41) and Nisell et al. (44), external and muscle forces acting on the leg were estimated. Compressive and shear forces were reported for a 110-kg subject squatting 250 kg. The quadriceps tendon force and tibiofemoral compressive force were approximately the same magnitude from 130° to 60° knee flexion. A maximum value of approximately 8000 N occurred at approximately 130° knee flexion and slowly declined to approximately 5500 N at approximately 60° knee flexion. At 30° knee flexion, compressive force was approximately 3500 N, whereas quadriceps tendon force dropped down to approximately 2000 N. Although these force magnitudes are quite high, most rehabilitation patients and athletes will experience considerably smaller forces, because the loads lifted in this study (approximately 2.5 times BW) are much greater than what most athletes or rehabilitation patients will utilize.

Patellar tendon force was approximately 6000 N at 130° knee flexion and slowly decreased to approximately 2000 N at 30° knee flexion. Although a peak patellar tendon force of 5000 N at 60° knee flexion was calculated by van Eijden et al. (63) during MVIC of the quadriceps femoris, the strength of the patellar tendon in the healthy knee is probably much higher than 5000 N. Cooper et al. (11) have quantified the ultimate strength of the central third of the patellar tendon (a 15-mm bone-patellar tendon-bone composite) to be 4389 ± 708 N. Zernicke et al. (75) reported a ruptured patellar tendon in a 82.2-kg weightlifter while performing a rapid squat descent during a 175-kg clean and jerk. A rapid downward acceleration was followed by tremendous forces being generated in the quadriceps to decelerate the weight in preparation to press the barbell overhead. Large decelerations during the squat generate tremendous forces across the knee structures, which is why the squat descent should be performed in a slow, controlled manner. A 14,500-N tensile force (17.5 times BW) was calculated in the patellar tendon at the time of rupture. Extrapolating these data, the ultimate strength of the patellar ligament in the healthy knee is probably somewhere between 10,000 and 15,000 N, which is approximately 13–19 times BW for an 80-kg individual. The peak quadriceps tendon force observed was the same as the 8,000 N peak force reported by van Eijden et al. (63). How much force the quadriceps tendon can generate before rupture is unclear. However, because Nisell and Ekholm (42) found that the thickness and breadth of the quadriceps tendon to be 35–40% greater than the thickness and breadth of the patellar tendon, it is likely that the ultimate strength of the quadriceps tendon is greater than the 10,000–15,000 N estimate for the ultimate strength of the patellar tendon.

The peak posterior shear force reported by Nisell and Ekholm (43) is similar to the peak PCL tensile force calculated by Escamilla et al. (17,18) and Wilk et al. (65) (Table 1), who also used powerlifters as subjects and quantified muscle forces. At approximately 60° knee flexion, the posterior shear force changed to an anterior shear force. This force increased fairly linearly throughout the remainder of the ascent. It is interesting that even though tremendous muscle forces were generated in the three subjects due to the large loads lifted, tensile forces in the PCL and ACL were only approximately 50% and 25%, respectively, of the estimated ultimate tensile strength in these ligaments. The presence of anterior shear forces during the second half of the ascent is in agreement with data from Beynnon et al. (5), who inserted strain transducers into the anteromedial bundle of the ACL in eight subjects immediately after arthroscopic knee meniscectomies and debridements. After the strain transducers were inserted into the ACL, the experimental procedures began. Under local anesthesia, the subjects were asked to squat down from an upright position to approximately 90° knee flexion, and then ascend back to an upright position. The subjects were then asked to repeat the squat using an elastic resistance cord, which generated 136 N of force at full knee extension and 34 N of force at 90° knee flexion. Minimal ACL strain (<4%) was observed at knee flexion angles less than 70° during both the squat descent and ascent, with no significant differences observed in ACL strain between the two squat conditions. ACL strain was greatest at full extension and progressively decreased as the knee flexed to 90°. There were some limitations to this study that would make it difficult to extrapolate these results to the barbell squat as performed by athletes in training. First, Markolf et al. (33) have reported that lateral and medial knee meniscectomies significantly increase anteroposterior knee laxity in the unloaded knee. Because most of the subjects from Beynnon et al. (5) had lateral and medial meniscectomies, this may have caused higher ACL strain than if the meniscectomies had not been performed. However, because the knee was loaded when ACL strain was recorded, anteroposterior translation due to the meniscectomies may be insignificant, because concomitant compressive force may resist this anteroposterior translation. However, compressive force magnitudes would be low due to a small external load (i.e., BW only), and muscle force production from the quadriceps and hamstrings would also be low due to the minimum resistance used and muscle inhibition due to surgery. Second, how the patients normally would perform the squat was probably affected by the knee surgery they just had a few hours prior. Nevertheless, the results of this study may be applied to patients who just had ACL reconstructive surgery and should be helpful to therapists, trainers, and orthopaedists who work with these patients in early postoperative rehabilitation.

The second part of the study by Nisell and Ekholm (43) involved a force analysis of a bilateral complete rupture of the quadriceps tendon (at its insertion into the superior patella) in a world-class powerlifter injured in competition while lifting a 382.5-kg load. The rupture occurred where the quadriceps tendon inserts into the patella. Although the injury was filmed with a single camera, a biomechanical analysis of the injured subject was not able to be conducted, because the filming was not perpendicular to the lifter's sagittal plane of motion. Because the injury occurred at the deepest portion of the squat just before beginning the ascent, knee joint forces were calculated in this position during a simulated squat using the three healthy subjects and the 382.5-kg load. Estimated quadriceps tendon force for the three subjects was 12–20 times BW, whereas the force in the patellar tendon was 9–14 times BW. Estimated compressive force was 11–17 times BW, whereas estimated posterior shear force was 2–3 times BW. Although these knee forces appear inordinate, very few individuals other than competitive powerlifters are capable or willing to squat with a 382.5-kg external load.

Knee joint biomechanical models and anthropometric data used to estimate knee forces in Nisell and Ekholm (43) were based on 10 subjects from their previous studies (41,44), with a mean height of 180 cm and a mean body mass of 75 kg. This is one limitation in the study by Nisell and Ekholm (43), because their three subjects had a mean body mass 20 kg more than and a mean height 7 cm less (Table 1) than the 10 subjects used in their biomechanical models. Another limitation in their study was that the effect of co-contraction of the hamstring musculature was not considered. Because the calculated knee torque was the resultant torque (i.e., sum of all flexor and extensor torques), hamstring force, which generates a knee flexor torque, would cause greater quadriceps force and extensor torque. Consequently, all calculated knee forces in this study would be underestimated, because these forces are functions of quadriceps tendon and patellar tendon forces. Because the quadriceps tendon thickness is significantly greater than the patellar tendon thickness, it should be able to withstand a higher load before rupture. However, high magnitudes of compressive force and stress (force/area) between the femoral intercondylar notch and the quadriceps tendon may increase the injury potential of the quadriceps tendon. Tendofemoral compressive force between the quadriceps tendon and the femoral intercondylar notch began high at 6000 N at approximately 130° knee flexion, quickly decreased to approximately 1750 N at 90° knee flexion and decreased to approximately 0 N at 60° knee flexion. Assuming a 3.4-cm^2 tendofemoral contact area when the knee is flexed approximately 130° (24), tendofemoral compressive stress would be 17.6 MPa (6000 N$\cdot$0.00034 m^{-2}). This large stress applied repetitively over time may cause degenerative changes in the tendofemoral complex. Hence, performing the squat at knee flexion angles less than 90° will minimize tendofemoral stress and minimize injury potential to the tendofemoral complex.

Ariel (3) used 12 experienced weightlifters to investigate forces acting about the knee joint during a deep knee barbell squat (Table 1). A computer program was written that took inertial, external, and muscle forces into account. The forces were modeled for three of the subjects, and their performance was believed to be representative of the other nine subjects. Subject one bounced at the bottom, subject two lifted the greatest load (295 kg), and subject three exhibited the greatest forward knee movement. Shear forces were generally greatest at knee flexion angles less than 60°. Shear force direction was not stated. Contrary to findings from several studies (2,13,17,18,20,43,58,60), shear forces progressively decreased at knee flexion angles greater than approximately 60°, with minimum shear values occurring at approximately 90–117° knee flexion. Minimum shear values were approximately 600 N for the subject that bounced at the bottom, approximately 120 N for the subject that lifted the most weight, and approximately 1120 N for the subject that had the greatest forward knee movement. Bouncing at the bottom of the squat increased shear force by approximately 33%. Interestingly, the subject that lifted the most had the smallest shear forces, and the lifter that had the greatest forward knee motion had the greatest shear forces. The results of this study indicate that forward knee movement, as well as bouncing at the bottom, both contributed to high shearing forces. Beyond 90° knee flexion the knee was thought to be more vulnerable, and the author suggested that shear forces may adversely affect knee ligaments. For all three subjects, an inverse relationship was observed between compressive and shear forces. This is contrary to the findings of several other studies (13,17,18,20,43,60), which found that both shear and compressive forces increase as knee flexion increases. Compressive forces were generally highest at higher knee flexion angles (Table 1).

Dahlkvist et al. (13) had their subjects perform the deep BW squat during regular, slow, and fast descents and ascents (Table 1). Kinematic data were recorded with one camera system filming at 50 Hz in the sagittal plane of motion. A force platform was used to quantify kinetic data, whereas EMG was used to estimate knee muscle forces. Cadence rates for the ascent and descent phases were not reported. The force in the patellar ligament was higher during the descent compared with the ascent. This was assumed to be due to greater deceleration needed during the descent to slow down the body. Furthermore, in both the slow and fast squat mean shear and compressive forces were 15–35% greater during the descent compared with the ascent. This contrary to the results of several other squat studies (17,18,20,58,60,65), which show similar compressive and shear forces between the descent and ascent. Shear and compressive forces generally were not significantly different between the slow and fast BW squats. The general pattern observed was as knee flexion increased, compressive and shear forces also increased.

Toutoungi et al. (60) examined cruciate ligament forces during the BW squat (Table 1). PCL forces, which were generated throughout the squat descent and ascent, progressively increased as the knees flexed and decreased as the knees extended. PCL forces were approximately 0–500 N between 0 and 50° knee flexion, and approximately 500–2700 N between 50 and 100°. During both the squat descent and ascent, PCL forces were generally less than 100 N up to 40° knee flexion, increased exponentially between 40 and 60° knee flexion, and then increased linearly from 60 to 100°. Therefore, these authors concluded that during PCL rehabilitation the squat should not be performed at these higher knee flexion angles, because PCL loading increased as knee flexion increased. Negligible ACL forces between 0 and 28 N were generated between 0 and 50° knee flexion. Therefore, these authors suggested the squat appears to be a safe exercise to perform during ACL rehabilitation. Interestingly, PCL force magnitudes throughout the squat were very similar to the posterior shear forces reported by Dahlkvist et al. (13), who also examined knee forces during the BW squat. The peak PCL forces reported by Toutoungi et al.

(60) and Dahlkvist et al. (13) appear inordinately high, considering their subjects only performed the BW squat, which requires minimal effort. These peak posterior shear forces reported by these authors are approximately 20–250% greater than the peak posterior shear forces reported by several other authors (3,17,18,20,43), whose subjects performed the barbell squat with an external resistance between 339–2453 N.

Hattin et al. (20) examined the effect of load, cadence, and fatigue on tibiofemoral joint force during the barbell squat (Table 1). Kinematic data were captured at 50 Hz by a three-camera motion system, while a force platform was used to collect kinetic data. Inverse dynamics and external forces were used to calculate knee joint forces. Three load conditions were used comprised of 15%, 22%, and 30% of each subject's 1 RM. Two different cadences were used: a) a slow cadence, where the descent and ascent phases lasted 2 s each; and b) a fast cadence, where the descent and ascent phases lasted 1 s each. To test for fatigue, 50 continuous repetitions were completed for each load and cadence, and were subdivided into initial, middle, and final phases. For the three load conditions, mean peak shear and compressive forces increased 25–85% from the initial phase to the final phase, with shear forces being most affected by fatigue. Hence, fatigue during the squat may increase loading of the cruciate ligaments. Fatigue became most apparent when the subjects were approximately half way through their 50 repetitions. Mean peak shear and compressive forces were 15–30% greater in the fast cadence squat compared with the slow cadence squat, which suggests that the squat should be performed in a slow controlled manner to minimize shear and compressive forces. Knee forces were symmetrical between the descent and ascent, with maximal shear and compressive forces occurring at maximum knee flexion. Mediolateral shear forces were less than 100 N throughout all squat conditions and phases and therefore can be discounted.

Andrews et al. (2) calculated knee shear forces using subjects experienced in both the barbell and machine squat exercises (Table 1). A two-dimensional lifting model was used utilizing external forces and inverse dynamics. Cadence and lifting loads were comparable during both types of squats. Three load conditions (40%, 60%, and 80% of their 4 RM) and fast (1 s) and slow (3 s) ascents were performed by each subject. The descent times were 2 s for all conditions. For both the barbell and machine squat exercises, peak shear force occurred at the lowest position of the squat. Shear forces were similar between the barbell and machine squats but stayed at peak values longer during the machine squat exercise. The authors concluded that shear forces were approximately 30–40% greater during the machine squat compared with the barbell squat. In both the barbell and machine squats, peak shear forces were 10–20% greater in the fast lifting rate compared with the slow lifting rate, which is in agreement with Hattin et al. (20). Hence, injury potential to the cruciate ligaments may be greater during the machine squat and during fast lifting rates.

Stuart et al. (58) used four 60-Hz cameras to collect kinematic data and a force platform to collect kinetic data during the barbell squat (Table 1). External and inertial forces were considered in quantifying compressive and shear forces. Compressive forces were 60–75% BW through the descent and ascent phases of the squat. The considerably smaller compressive and shear forces in this study as compared with other studies (2,3,17,18,20,43) is primarily due to less weight being lifted and the omission of muscle force contributions. Posterior shear forces were observed for all subjects throughout the descent and ascent phases. These shear forces progressively increased as the knees flexed and decreased as the knees extended. The authors concluded that the shear force magnitudes calculated were unlikely to be detrimental to the injured or reconstructed PCL. Furthermore, since no anterior shear forces were observed, performing the squat may be appropriate for ACL patients.

Patellofemoral Compressive Forces

Patellofemoral compressive forces produce stress (compressive force divided by contact area) on the articular cartilage of the patella and patellar surface of the femur. Excessive compressive force and stress, or repetitive occurrences of lower magnitude force and stress, may contribute to patellofemoral degeneration and pathologies, such as patella chondromalacia and osteoarthritis. There are three forces acting on the patella during the squat: 1) quadriceps tendon force, 2) patellar tendon force, and 3) patellofemoral compressive force. During the squat, all these forces are affected by knee flexion angle. Mathematically, compressive force is greatest at higher knee flexion angles, because there are larger force components from the quadriceps tendon and patellar tendon in the compressive direction.

Patellofemoral compressive forces arise from contact between the undersurface of the patella and the femoral condyles. From full extension to full flexion, the patella moves caudally approximately 7 cm, with femoral contact on the patella moving cranially as the knee flexes. Patellofemoral contact has been reported to initially occur between 10 and 20° knee flexion (24,26), which is when the patella begins to glide onto the patellar surface of the femur. The femur makes contact with the medial and lateral inferior facets between approximately 20 and 30° knee flexion, with the medial and lateral middle facets between approximately 30 and 60°, with the medial and lateral superior facets between approximately 60 and 90°, and with the medial vertical "odd" facet and lateral superior facet between approximately 90 and 135° (24,26). At approximately 90° knee flexion, the "odd" facet for the first time makes contact with the lateral margin of the medial condyle (26). Because contact is increased as the knee continues into full flexion, this area is a common site of osteochondritis dissecans.

To date, there are six known studies (Table 1) that quantified patellofemoral compressive forces during the dynamic squat (13,17,18,43,51,68). Four of these studies involved subjects lifting approximately 65–75% of their 1 RM during the barbell squat (17,18,43,68), whereas the remaining two studies involved the BW squat (13,51). Escamilla et al. (17) employed a mathematical model of the patella (42,61,62) to calculate compressive forces as a function of knee angle during the barbell squat (Table 1). Compressive forces increased as the knees flexed, decreased as the knees extended, and were slightly greater during the descent compared with the ascent. During the descent a peak compressive force of 4548 $\pm$ 1395 N occurred at 85° knee flexion, whereas during the ascent a peak compressive force of 4042 $\pm$ 955 N occurred at 95° knee flexion. Because peak compressive forces generally occur near maximum knee flexion, individuals with patellofemoral disorders should avoid performing the squat at high knee flexion angles. However, performing the squat in the functional range between 0 and 50° of knee flexion may be appropriate for patellofemoral patients, because only low to moderate patellofemoral compressive forces were generated in this range.

Escamilla et al. (18) examined the effects of stance width and foot angle on patellofemoral compressive forces during the barbell squat (Table 1). No significant differences in compressive forces were observed between the feet pointing straight ahead and the feet turned outward 30°. Compressive forces progressively increased as the knees flexed and decreased as the knees extended. The authors reported a 15% increase in compressive forces in the wide stance compared with the narrow stance between 21 and 79° knee flexion angle during the squat descent. No significant differences in compressive forces occurred between the narrow and wide stance during the squat ascent. In addition, the squat descent generated 8% greater compressive forces than the squat ascent at higher knee flexion angles between 79 and 95°, whereas the squat ascent generated 16–21% greater compressive forces than the squat descent at lower knee flexion angles between 27 and 63°. It can be inferred from these data that narrow stance may be preferred over the wide stance when the objective is to minimize compressive forces.

Wretenberg et al. (68) employed both weightlifters and powerlifters in quantifying compressive forces (Table 1). The powerlifters employed a low-bar squat position, in which the bar was positioned across the back approximately 3–5 cm below the level of the acromion. The weightlifters employed a high-bar squat position, in which the bar was positioned across the back at approximately the level of the acromion. Compressive forces generally increased as knee flexion increased. The large disparity in normalized values between weightlifters and powerlifters is probably due to technique variations, such as low and high bar positions. Powerlifters typically employ a low-bar squat position and greater forward trunk lean compared with weightlifters. The primary reason for this is to lift more weight, because powerful trunk (e.g., erector spinae) and hip (e.g., gluteus maximus and hamstrings) musculature are more involved in this position compared to the high-bar squat position employed by weightlifters, in which the trunk remains more upright. Consequently, the low-bar squat position produced greater hip extensor torque and less knee extensor torque compared with the high-bar squat. Mean peak knee extensor torques are typically between 100 and 300 N·m during the barbell squat (3,17,30,36,43,58,66,68,69). Because these torques are resultant (i.e., net) torques, they represent the sum of all flexor and extensor knee torques. For example, a 500-N·m extensor torque and a 300-N·m flexor torque yield a 200-N·m resultant torque. Hence, it cannot be deduced that lower knee extensor torque equates to less quadriceps force being produced in the powerlifters, although it is enticing to do so. Less quadriceps force would indeed explain why peak patellofemoral compressive force was less in powerlifters, because quadriceps tendon force and patellar tendon force would also be less. However, EMG data from Wretenberg et al. (68) show greater quadriceps and hamstring activity in the low-bar squat compared to the high-bar squat. Because the hamstrings produce a knee flexor torque, a greater knee extensor torque would be needed. Hence, greater quadriceps force would be needed to generate this force. In addition, gastrocnemius force can also cause the quadriceps to generate more force during the squat, because the gastrocnemius also generates a knee flexor torque by their duel role as knee flexors and ankle plantar flexors. Moderate gastrocnemius activity has been observed during the squat (13,17,18), which is needed during the squat to control ankle dorsiflexion during the descent and cause ankle plantar flexion during the ascent.

Many strength and power athletes train the low-bar squat to develop the hip and trunk extensors primarily. In contrast, many athletes use the high-bar squat to elicit more quadriceps development and less hip and trunk extensor development. Also, because the high-bar squat is more similar to the squat movement performed during the clean and jerk, it is preferred by weightlifters. The low-bar squat technique may be desirable for athletes or rehabilitation patients who want to perform the squat but minimize patellofemoral compressive force. Moreover, the low-bar squat position of greater forward trunk lean has been reported to decrease potential ACL strain, in part due to greater hamstring activity and less quadriceps activity (47). Hence, squatting with greater forward trunk lean may be appropriate for those whose goal is to minimize ACL stress. Furthermore, greater forward trunk lean also minimizes forward knee movement, which has been shown to increase knee shear forces (3). Unfortunately, a greater forward trunk lean may increase the injury risk to back musculature and ligamentous structures.

Nisell and Ekholm (43) employed competitive powerlifters to quantify compressive forces during the barbell squat ascent (Table 1). Like the three previous barbell squat studies (17,18,68), peak compressive force occurred near maximum knee flexion and progressively decreased as the knees extended. Although the mean peak compressive force was 7.2 times bodyweight (Table 1), when normalized by BW and load lifted, the mean peak compressive force was nearly identical to the mean peak compressive forces reported by Escamilla et al. (17,18) and Wretenberg et al. (68).

The final two studies that quantified patellofemoral compressive force were Dahlkvist et al. (13) and Reilly and

Martens (51), whose subjects performed the BW squat (Table 1). Like the four previous studies (17,18,43,68), compressive forces progressive increased as the knees flexed and decreased as the knees extended, peaking near maximum knee flexion. Normalized values from Dahlkvist et al. (13) and Reilly and Martens (51) were approximately 4 times greater than normalized values from Escamilla et al. (17,18), Nisell and Ekholm (43), and Wretenberg et al. (68). Although some of these large disparities in normalized values may be due to methodological differences among studies, it seems unlikely that the BW squat, which requires relatively little effort to perform, can generate the 5500- to 6400-N peak compressive forces reported by Dahlkvist et al. (13) and Reilly and Martens (51).

Compressive forces generated during the barbell squat were 4–7 times BW when squatting with moderate loads (65–75% of 1 RM). Although these loads are higher than most rehabilitation patients will experience, they are typical loads for strength and power athletes while performing the barbell squat. Unfortunately, it is currently unknown how much patellofemoral compressive force and stress is detrimental to the patellofemoral joint. Patellofemoral joint contact has been reported to be 2.6 ± 0.4 cm^2 at 20° knee flexion, 3.1 ± 0.3 cm^2 at 30°, 3.9 ± 0.6 cm^2 at 60°, 4.1 ± 1.2 cm^2 at 90°, and 4.6 ± 0.7 cm^2 at 120° (24). By using these contact areas and squat ascent compressive force data from Escamilla et al.(17), patellofemoral joint stress at 20°, 30°, 60°, and 90° knee flexion would be approximately 1.15 MPa, 2.42 MPa, 7.69 MPa, and 11.6 MPa, respectively. Consequently, patellofemoral compressive force and stress both increase as the knees flex, reaching peak values at approximately 90–100° knee flexion. Beyond 90–100° knee flexion, compressive force has been shown to remain relatively constant (17,18,43). Hence, stress may decrease at larger knee flexion angles, because patellofemoral contact area continues to increase.

The rate of increase in compressive force is maximum between approximately 50–80° knee flexion (17,18), thus generating proportionately greater patellofemoral force compared with lower knee flexion angles. Therefore, performing the squat within the functional range of 0–50° knee flexion will minimize patellofemoral compressive force and stress, and may be effective for athletes or patients with patellofemoral pathologies. For athletes with healthy knees, performing the squat at higher knee flexion angles (approximately 90–110°) should not be problematic, as long as heavy loads are not used excessively. This can be a potential problem for powerlifters and football players, who often train with heavy loads for long periods of time. Periodization techniques should be employed when performing the squat, in which training is divided into light, medium, and heavy intensity cycles throughout the year.

Muscle Activity

To determine which muscles are being developed during the squat and to what degree, it is helpful to quantify muscle activity through the use of EMG. To date, there are 16 known studies that have quantified muscle activity about the knee during the dynamic squat (6,13,17,18,25,27,34,38,40,54,55,58,65,68–70). Eleven of these studies performed the barbell squat with an external load (17,18,34,40,54,55,58,65,68–70), whereas the remaining five studies (6,13,25,27,38) performed the BW squat. The primary knee muscles utilized during the squat are the quadriceps, hamstrings, and gastrocnemius, and co-contractions among these muscles are believed to enhance knee stability (15,16,31,46,47,73,74).

Escamilla et al. (17) and Wilk et al. (65) quantified quadriceps, hamstring, and gastrocnemius activity using a 12-RM load for 10 male subjects experienced in the squat. Quadriceps activity progressively increased as the knee flexed and decreased as the knees extended, with peak activity occurring at approximately 80–90° knee flexion. Similar results were observed in several other studies (18,27,40,55,58,69). Quadriceps activity remained fairly constant beyond 80–90° knee flexion, which has also been observed in other studies (58,68,69). Hence, descending beyond 90° knee flexion, which is near the parallel squat position, may not enhance quadriceps development.

Escamilla et al. (17) reported that the two vasti muscles produce 40–50% more activity than the rectus femoris, which is in agreement with squat data from Escamilla et al. (18), Wretenberg et al. (68,69), and Isear et al. (27). The lower activity observed in the rectus femoris compared with the vasti muscles may be due to its biarticular function as both a hip flexor and knee extensor. Increased activity from the rectus femoris would increase hip flexor torque, with a concomitant increase in the amount of hip extensor torque needed from the hamstrings, gluteus maximus, and adductor magnus (ischial fibers) to extend the hip. The rectus femoris is probably more effective as a knee extensor during the squat when the trunk is more upright, because it is in a lengthened position compared with when the trunk is tilted forward in hip flexion. Compared with each other, the vastus medialis (VM) and vastus lateralis (VL) produced approximately the same amount of activity, which is in agreement with data from several other studies (18,38,55,65).

Hamstring activity from Escamilla et al. (17,18) and Wilk et al. (65) was highest during the squat ascent, with the lateral hamstrings showing greater overall activity than the medial hamstrings. These authors reported peak hamstring activity between approximately 30 and 80% of a MVIC, occurring near 50–70° knee flexion. In contrast, peak hamstring activity from Isear et al. (27), Ninos et al.(40), and Stuart et al. (58) were approximately 12% MVIC, 15% MVIC, and 20% MVIC, respectively, with peak values occurring between 10 and 60° knee flexion. The lower hamstring activity in these studies is probably due to their subjects lifting a lower percentage of their 1 RM. Subjects in Isear et al. (27) used no external lifting loads, subjects in Ninos et al. (40) and Stuart et al. (58) used lifting loads of 25% BW and 28% BW, respectively, whereas subjects in Escamilla et al. (17,18) lifted 140–160% BW. Several studies have reported greater overall hamstring activity during the ascent compared with the descent (17,18,27,34,40,58). Because the hamstrings are biarticular muscles, it is difficult to determine whether these muscles act eccentrically during the descent and concentrically during the ascent, as commonly is believed. In reality, they may actually be working nearly isometrically during both the squat descent and ascent, because they are concurrently shortening at the knee and lengthening at the hip during the descent, and lengthening at the knee and shortening at the hip during the ascent. If they are indeed working eccentrically during the descent and concentrically during the ascent, as is traditionally believed, then data from the above studies would be in accord with data from Komi et al. (29), who reported decreased activity during eccentric work and increased activity during concentric work. In any case, the hamstrings probably do not change length much throughout the squat. Hence, in accordance with the length-force relationship in skeletal muscle, a constant length in the hamstrings will allow them to be more effective in generating force throughout the entire squatting movement.

Four studies have reported gastrocnemius activity and force during the squat (13,17,18,27). Escamilla et al. (17) observed moderate gastrocnemius activity during the squat, which progressively increased as the knees flexed and decreased as the knees extended. Escamilla et al. (17) and Isear et al. (27) reported peak gastrocnemius activity between 60 and 90° knee flexion. Because the ankle dorsiflexes during the descent and plantar flexes during the ascent, it is a common belief that the gastrocnemius contracts eccentrically during the descent to help control the rate of ankle dorsiflexion, and concentrically during the ascent to aid in ankle plantar flexion. However, because the gastrocnemius is a biarticular muscle, its length may not change much throughout the squat, because it shortens at the knee and lengthens at the ankle during the descent, and lengthens at the knee and shortens at the ankle during the ascent.

Four studies have investigated the effects of varying foot angles on quadriceps, hamstrings, and gastrocnemius activity (18,25,40,54). Escamilla et al. (18) had 10 male experienced lifters perform the barbell squat using a 12-RM lifting load. The two foot angles employed were with the feet pointing straight ahead and with the feet turned outward 30°. No significant differences were observed in quadriceps, hamstrings, or gastrocnemius activity between the two foot angles. Signorile et al. (54) had 10 male subjects perform the barbell squat with an 8- to 10-RM load using three different foot angles: a) feet pointing straight ahead, b) toes pointed outward as far as possible (approximately 80° from the straight ahead position), and c) toes pointed inward approximately 30° from the straight ahead position. They found no significant differences in VM, VL, and rectus femoris activity among the three foot positions. Ninos et al. (40) had 25 male and female subjects perform the squat (0–60° knee flexion) with a 25%-BW load using two different foot angles: a) self-selected neutral position and b) foot turned outward 30° from the neutral position. No significant differences were observed in quadriceps (VM and VL) or hamstrings (semimembranosus, semitendinosus, and biceps femoris) activity between the two foot angles. Hung and Gross (25) examined the effects of foot wedges on vastus medialis oblique (VMO) and VL activity. Their study varied foot angle by changing forefoot inversion and eversion rather than changing forefoot adduction and abduction. Sixteen subjects performed a 1-leg squat (0–50° knee flexion) on a level surface with a 10° medial wedge and with a 10° lateral wedge. No significant EMG differences in VMO:VL ratios was observed among the three foot angle positions. Hence, data from these studies show that varying foot angles do not appear to affect quadriceps, hamstrings, or gastrocnemius activity during the squat.

To date, there are four known studies that have investigated the effects of stance width (narrow stance vs wide stance) on knee muscle activity during the squat (1,18,34,59). Escamilla et al. (18) found that gastrocnemius activity was 21% greater in the narrow stance compared with the wide stance barbell squat. In addition, these authors reported no significant differences in quadriceps or hamstrings activity between the narrow and wide stance squats, which is in agreement with EMG data from McCaw and Melrose (34), whose subjects lifted similar relative loads (60–75% 1 RM) and employed similar narrow and wide stances. Tesch (59), using magnetic resonance imaging (MRI) immediately after performing the squat, also showed no differences in quadriceps or hamstrings activity between narrow and wide stance squats. Anderson et al. (1) found no significant differences in VMO:VL ratios between narrow and wide stances during the BW squat but did report significantly greater VMO:VL ratios with increasing knee flexion angles (0–30°, 0–60°, 0–90°). These results imply that increasing knee flexion angles during the BW squat elicited greater activity of the VMO relative to the VL. One additional study (6) investigated the effects of moving the feet forward while performing a machine squat exercise. This author reported that a more forward foot position during a machine squat increased quadriceps and hamstrings activity.

Knee Stability

Knee stability is paramount in enhancing an athlete's performance and training, minimizing injury potential, or improving a patient's knee rehabilitation. Normal tibial internal rotation in knee flexion and tibial external rotation in knee extension (i.e., the screw home mechanism) that occurs when the foot is free to move changes during the squat exercise. Instead, the femur tends to rotate externally during knee flexion and internally during knee extension. As the femur externally rotates during the descent, the tibia also attempts to externally rotate. This is supported Costigan and Reid (12), who demonstrated that during the squat the tibia generated an external rotation torque against the ground during knee flexion and an internal rotation torque during knee extension. The flexion, extension, internal rotation, and external rotation knee movements that occur during the squat are supported in part by cruciate and collateral knee ligaments, which help provide knee stability. The ACL and PCL provide anteroposterior knee stability, while the MCL and LCL provide mediolateral knee stability. Knee ligament arthrometers can be employed to quantify anteroposterior and mediolateral knee stability by measuring anteroposterior and mediolateral tibial translation relative to the femur. This is important because excessive anteroposterior or mediolateral knee translation measurements may indicate knee instability due to damage to the cruciate or collateral ligaments. To date, there are seven known studies that examined how the dynamic squat affects anteroposterior or mediolateral knee stability (5,10,21,28,37,49,57).

Klein (28) examined how the "deep" squat exercise affected anteroposterior and mediolateral knee stability. The deep squat was defined as when the posterior thigh came in contact with the calf, which typically occurs between approximately 130 and 150° knee flexion. This is typically how competitive weightlifters perform the squat, because this is what occurs during the "clean" portion of the clean and jerk exercise. Klein (28) first investigated the strain on the collateral and cruciate ligaments during the deep squat, comparing a deep squat group with a control group. The deep squat group was comprised of 128 competitive weightlifters, all of which practiced the deep squat exercise in training and competition. The control group was comprised of 386 subjects from beginning weight training, basketball, and gymnastic classes from local universities. None of the controls had ever performed the deep squat. Comparing the results within the deep squat group, the LCL was stretched to a greater extent than the MCL. Furthermore, there was 19.4% more right LCL instability compared with right MCL instability and 12% more left LCL instability compared with left MCL instability. Compared with the control group, the deep squat group showed a) 61% greater instability in two or more ligaments of both legs; b) 46% and 58%, respectively, greater MCL instability in the right and left legs; c) 67% and 59%, respectively, greater LCL instability in the right and left legs; and d) 16% and 25%, respectively, greater ACL instability in the right and left legs. Although the PCL was not investigated in this study, Klein (28) surmised that the PCL could be abnormally stressed during the deep squat due to the "jacking apart" action that occurs within the joint due to the posterior thigh musculature coming in contact with the calf muscle at the bottom position of the squat. In effect, this changes the center of rotation from somewhere within the knee joint to the point of contact between the thigh and calf musculature. These *in vivo* results are supported by cadaveric data. Using 64 cadaver knees, Klein (28) found that the MCL stretched a mean of approximately 8% of its original length, whereas the LCL stretched a mean of approximately 13% of its original length during extreme knee flexion. From these cadaveric data, Klein (28) concluded that both the LCL and MCL is susceptible to injury and abnormal stretch during the deep squat. In addition, Klein (28) suggested that as the femur externally rotated during the descent, the menisci are forced to move posteriorly, causing the posterior portions of the menisci to be compressed between the tibial and femoral condyles. As the femur internally rotated at the beginning of the ascent, the posterior medial meniscus is forced toward the center of the joint space. This can place strain on the inner medial meniscus, causing it to tear. The medial meniscus also can tear due to a twisting strain applied to the MCL, especially with severe internal rotation of the femur relative to the tibia. Because the MCL attaches to the medial meniscus, a twisting strain to the MCL may tear or detach the medial meniscus from its adjacent fibrous capsule. Part of this torn meniscus can become displaced toward the center of the joint space and become lodged between the femoral and tibial condyles. This can "lock" the knee is a flexed position, preventing full extension. Based on these *in vivo* and cadaveric data, Klein (28) recommended that the parallel squat be used in place of the deep squat, because the deep squat may produce deleterious effects to the collateral and cruciate ligaments. Many strength and conditioning specialists also recommend the parallel squat over the deep squat due to the deleterious effects the deep squat have on knee stability (9).

Meyers (37) reproduced the deep squat study by Klein (28), using the same mediolateral collateral ligament testing instrument to measure collateral ligament stability. Sixty-nine male subjects were randomly assigned to eight different treatment groups involving variations of the deep and parallel squat consisting of low and high lifting loads and speeds. All subjects trained on alternating days 3 $d \cdot wk^{-1}$ for 8 wk. For each training session each subject performed one set of 10 repetitions, for a total of 240 repetitions for the 8 wk. All subjects were pretested 1 wk before beginning their training program and posttested 1 wk after the end of training. No significant differences were found within any of the eight treatment groups in collateral ligament instability and knee joint flexibility, although the amount of stretch in the LCL decreased from the pretest to the posttest in both the deep squat and parallel squat. Contrary to Klein's results, these data infer that both the deep squat and parallel squat were not detrimental to knee stability.

Henning et al. (21) measured ACL elongation using *in vivo* instrumentation during a one legged half squat between 20–90° knee flexion. Two subjects were used in this study, both with a grade II sprain of their ACL. ACL elongation was expressed relative to fiber elongation using a force of 357 N during a Lachman test. The one legged squat produced 21% as much elongation as the Lachman test. In contrast, normal walking produced 36% as much elongation as the Lachman test, jogging 2.24 $m \cdot s^{-1}$ on a level treadmill produced 63% as much elongation as the Lachman test, and partial knee extension exercises produced over 100% as much elongation as the Lachman test. Because the squat generated lower ACL strain compared with walking or jogging, it was concluded that the squat was a low risk exercise in rehabilitation of the ACL. As previously discussed, Beynnon et al. (5) also reported minimal *in vivo* ACL strain during the squat at knee flexion angles between 20–60°.

Steiner et al. (57) used a commercial knee laxity testing device to measure anteroposterior knee laxity just before and just after performing the squat, playing basketball, and running. Four groups of subjects (37 male and 18 female) with healthy knees were used, comprised of sedentary controls, squat powerlifters, basketball players, and distance runners. Nine sedentary controls were measured before and after a 2-h time interval. Twenty-four athletes performed the powerlifting squat, lifting an average of 1.6 times BW for a mean of 24 repetitions. Ten basketball players were tested before and immediately after a strenuous 1.5-h practice, and 12 distance runners were measured 30 min before and immediately after a sanctioned 10-km run. The percent change between the pretest and posttest for anterior laxity, posterior laxity, and total anteroposterior laxity, respectively, were 1%, 4%, and 3% for the control group, 5%, −3%, and 2% for the powerlifters, 19%, 18%, and 19% for the basketball players, and 20%, 19%, and 19% for the distance runners. Significant increases in anteroposterior laxity were observed in the basketball players and distance runners (average increase of 0.5 mm in anterior laxity and 0.6-mm increase in posterior laxity) but not in the powerlifting and control groups. The powerlifters had the smallest percent change in total anteroposterior laxity, with a decrease in posterior laxity from pretest to posttest. Resting anteroposterior laxities among the groups were also recorded. Distance runners had significantly less anterior laxity than all other groups, whereas powerlifters had significantly less posterior laxity than all other groups. Because the greatest anteroposterior laxity was found in basketball players and distance runners, it was deduced that high compressive loads in the knee during the powerlifting squat might have facilitated the low anteroposterior laxities observed.

Chandler et al. (10) examined how the squat exercise affected anteroposterior knee stability. This study was comprised of two parts: 1) an 8-wk squat training study and 2) a descriptive study of powerlifters and weightlifters. In part 1, 100 male and female subjects volunteered. They were divided into three groups: a) half squat group, b) parallel squat group, and c) control group. All subjects had no previous history of ligament or cartilage injuries. Twenty-seven powerlifters and 28 weightlifters comprised two additional groups for part 2 of this study. The parallel squat and half squat groups were the only groups that performed a periodization weight training program. All subjects were tested for knee stability with a knee ligament arthrometer at 30° and 90° knee flexion. Measurements were taken at pre-, mid-, and post-training intervals. Anterior drawer (69–88 N applied force), posterior drawer (88 N applied force), maximum manual drawer, and the quadriceps active drawer tests were used to quantify knee stability. For part 1 of this study, the only significant difference across trials at 30° knee flexion was in posterior drawer. All groups tested demonstrated significantly greater posterior drawer knee displacement measurements in the posttest compared to the pretest. These results support squat biomechanics data that show low to moderate PCL loading throughout the squat movement but only minimum or no ACL loading (5,13,17,18,20,43,58,60,65). For 90° knee flexion, male subjects showed significantly less displacement in both the anterior and posterior drawer tests. For part 2 of this study comparing the lifting groups with a control group, the powerlifters and weightlifters showed tighter knees for the 90° quadriceps active drawer test. For the anterior drawer at 90°, powerlifters showed tighter knees than the control group, whereas the weightlifting group showed no significant differences compared to powerlifters and controls. When groups were subdivided by skill, low-skilled weightlifters had significantly tighter knees than the controls for the quadriceps active drawer at 90° knee flexion. Because the significant differences seen were all less than 2 mm, perhaps the 8-wk program 2–3 times per week was not long enough to elicit meaningful changes. However, this places more importance on the findings in part 2 of this study. The weightlifters and powerlifters were all successful competitive lifters and had been lifting heavy loads for many years. Interestingly, the control group, who had very little or no squatting experience, consistently had the loosest knees. Furthermore, powerlifters had tighter knees than the controls on seven of the nine measurements, whereas weightlifters were tighter than the controls for four of the nine measurements. The authors concluded that the squat did not have negative effects on knee stability and may be considered safe in terms of not causing permanent stretching of the ligaments.

Panariello et al. (49) examined the effect of the squat exercise on anteroposterior knee translation in professional football players. Thirty-two subjects with normal knees participated in a 21-wk off-season conditioning program involving the parallel squat. Two periodization training cycles were utilized during the 21 wk. The first cycle was 12 wk in duration, immediately followed by a 9-wk cycle. The squat was performed twice a week, with subjects performing an average of 32 repetitions per session and lifting an average of 130–200% BW. A knee ligament arthrometer KT-1000 (Medmetric Corp., San Diego, CA) was used to measure anteroposterior knee stability. Both knees of each subject were tested passively and actively at 30° and 90° knee flexion before the start of the training program, at the end of the 12-wk cycle, and at the end of the 9-wk cycle. Passive anterior drawer tests were conducted with 67 N (15 lbs), 89 N (20 lbs), and 133 N (30 lbs) of force, whereas passive posterior drawer tests were conducted with 67 N and 89 N of force. Active tests were conducted with maximum voluntary isometric contractions of the quadriceps and hamstrings. Mean anterior and posterior drawer measurements were compared between the start of the training program and the end of the 12-wk cycle, between the end of the 12-wk cycle and the end of the 9-wk cycle, and between the start of the training program and the end of the 9-wk cycle. There were no significant differences between any pair of pre- and post-exercise measurements, and no significant differences in anteroposterior knee translations in athletes using the squat as part of their training regimen. Hence, it was deduced that the squat is a safe exercise to include in an athlete's training program.

CONCLUSIONS

This review examined knee biomechanics during the dynamic squat exercise. Low to moderate posterior shear forces were generated throughout the squat, restrained primarily by the PCL. In addition, low anterior shear forces were observed between 0 and 60° knee flexion, restrained primarily by the ACL. Hence, the squat may be an effective exercise to employ after an ACL injury or reconstruction and may also be appropriate with light loads after PCL injury or reconstruction. Low to high tibiofemoral and patellofemoral compressive forces were produced during the squat. Tibiofemoral compressive force helps resists anteroposterior shear forces and translation. Excessive patellofemoral compressive force can lead to patellofemoral pathologies, such as chondromalacia or osteoarthritis. Patellofemoral compressive forces, tibiofemoral compressive forces, and tibiofemoral shear forces all progressively increased as the knees flexed and decreased as the knees extended, reaching peak values near maximum knee flexion. Hence, training the squat in the functional range between 0 and 50° knee flexion may be appropriate for many knee rehabilitation patients. For athletes with healthy knees, performing the parallel squat is recommended over the deep squat, because injury potential to the menisci and cruciate and collateral ligaments may increase with the deep squat. The squat does not compromise knee stability and may enhance stability if performed correctly. Finally, the squat can be effective in developing hip, knee, and ankle musculature, because moderate to high quadriceps, hamstrings, and gastrocnemius activity were produced during the squat. Muscle activity generally progressively increased as the knees flexed and decreased as the knees extended, which supports athletes performing the parallel squat over the half squat.

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