Squatting Kinematics and Kinetics and Their Application to Exercise Performance

Brad J. Schoenfeld
Global Fitness Services, Scarsdale, New York

Abstract

Schoenfeld, BJ. Squatting kinematics and kinetics and their application to exercise performance. *J Strength Cond Res* 24(12): 3497–3506, 2010—The squat is one of the most frequently used exercises in the field of strength and conditioning. Considering the complexity of the exercise and the many variables related to performance, understanding squat biomechanics is of great importance for both achieving optimal muscular development as well as reducing the prospect of a training-related injury. Therefore, the purpose of this article is 2-fold: first, to examine kinematics and kinetics of the dynamic squat with respect to the ankle, knee, hip and spinal joints and, second, to provide recommendations based on these biomechanical factors for optimizing exercise performance.

Key Words squat, biomechanics, kinetics, kinematics

Introduction

The squat is one of the most frequently used exercises in the field of strength and conditioning. It has biomechanical and neuromuscular similarities to a wide range of athletic movements and thus is included as a core exercise in many sports routines designed to enhance athletic performance (20,62). It also is an integral component in the sports of competitive weightlifting and powerlifting and is widely regarded as a supreme test of lower-body strength (17,18).

Benefits associated with squat performance are not limited to the athletic population. Given that most activities of daily living necessitate the simultaneous coordinated interaction of numerous muscle groups, the squat is considered one of the best exercises for improving quality of life because of its ability to recruit multiple muscle groups in a single maneuver (22). The squatting movement has close specificity to many everyday tasks (such as lifting packages and picking up children), as well as having an indirect correlation to countless other chores and hobbies.

Address correspondence to Brad Schoenfeld, brad@lookgreatnaked.com.

24(12)/3497–3506

Journal of Strength and Conditioning Research

The squat also is becoming increasingly popular in clinical settings as a means to strengthen lower-body muscles and connective tissue after joint-related injury. It has been used extensively for therapeutic treatment of ligament lesions, patellofemoral dysfunctions, total joint replacement, and ankle instability (14,56). Moreover, the closed-chain stance required for performance reduces anterior cruciate ligament (ACL) strain (64), making it superior to the knee extension for rehabilitation of ACL injury (21,65).

Performance of the dynamic squat begins with the lifter in an upright position, knees and hips fully extended. The lifter then squats down by flexing at the hip, knee, and ankle joints. When the desired squat depth is achieved, the lifter reverses direction and ascends back to the upright position. This dynamically recruits most of the lower-body musculature, including the quadriceps femoris, hip extensors, hip adductors, hip abductors, and triceps surae (51). In addition, significant isometric activity is required by a wide range of supporting muscles (including the abdominals, erector spinae, trapezius, rhomboids, and many others) to facilitate postural stabilization of the trunk. In all, it is estimated that over 200 muscles are activated during squat performance (66).

Squats can be performed at a variety of depths, generally measured by the degree of flexion at the knee. Strength coaches often categorize squats into 3 basic groupings: partial squats (40° knee angle), half squats (70 to 100°), and deep squats (greater than 100°). However, no standardized measures of quantification have been universally recognized, and terminology can differ between researchers. Other modifying facts associated with the squat involve varying intensity of load, foot placement, speed of movement, level of fatigue, and position of load.

When performed properly, squat-related injuries are uncommon (75). However, poor technique or inappropriate exercise prescription can lead to a wide range of maladies, especially in combination with the use of heavy weights. Documented injuries from squatting include muscle and ligamentous sprains, ruptured intervertebral discs, spondylolysis, and spondylolisthesis (72).

Considering the complexity of the exercise and the many variables related to performance, understanding squat biomechanics is of great importance both for achieving optimal muscular development as well as reducing the prospect of a training-related injury. Therefore, the purpose of this article is 2-fold: first, to examine kinematics and kinetics of the dynamic squat with respect to the ankle, knee, hip. and spinal joints and. second, to provide recommendations based on these biomechanical factors for optimizing exercise performance.

Joint Kinetics and Kinematics During The Squat

Ankle Complex

The ankle complex is comprised of the talocrural and subtalar joints. Actions taking place at these joints include dorsiflexion/plantar flexion and eversion/inversion, as well as a small amount of abduction and adduction (63,73). During squat performance, the talocrural joint (articulation of the tibia and fibula with the talus) facilitates movement through the actions of dorsiflexion and plantar flexion, whereas the primary action at the subtalar joint is to maintain postural stability and limit eversion/inversion at the foot. Normal talocrural range of motion is 20° dorsiflexion and 50° plantar flexion, whereas range of motion about the subtalar joint is approximately 5° each for eversion and inversion without forefoot movement (13). The gastrocnemius and soleus, collectively referred to as the triceps surae, are the primary musculature responsible for carrying out dynamic ankle joint movement, contracting concentrically during plantar flexion and eccentrically during dorsiflexion (54,63).

The ankle complex contributes significant support and aids in power generation during squat performance (28). However, kinetic data of the ankle during squatting is limited because most studies have focused on the biomechanics of the knee, hip, and spine. Peak ankle moments of 50 to 300 Nm have been reported during the squat, which are far below those seen at the knee and hip (18).

Dionisio et al. (15) evaluated the ankle complex as part of a comprehensive analysis of joint activity while squatting. They found that, in the upright position before squatting, the center of pressure (COP) was projected at approximately the mid-foot, with ankle joint torque directed toward plantar flexion. During the acceleration phase, COP shifted toward the heel, with plantar flexion torque decreasing. Finally, the deceleration phase was characterized by displacement of COP to the toes with a correspondingly large increase in plantar flexion torque about the ankle. Specific quantification of joint torque, however, was not reported in this study. What is more, subjects were instructed to squat as fast as possible, making the relevance of these findings to slower cadences questionable.

The gastrocnemius has been the primary ankle joint muscle studied in squat performance. It is believed that the medial head of the gastrocnemius acts as dynamic knee stabilizer during squatting, helping to offset knee valgus moments as well as limiting posterior tibial translation (7,56). The gastrocnemius shows only moderate levels of activation during squatting, with activity tending to progressively increase as the knees flex and decrease as the knees extend (16). This is consistent with the fact that its force arm peaks at or near maximal knee flexion (18).

Toutoungi et al. (70) reported that the soleus was more active than the gastrocnemius when squatting at high degrees of flexion. Considering the respective anatomic configurations of the 2 muscles, this appears logical. The soleus is a pure plantar flexor with proximal attachments at the tibia and fibula and distal attachments at the calcaneous. The gastrocnemius, on the other hand, is a bi-articular muscle that carries out plantar flexion as well as assisting in knee flexion. Given these dual roles, the gastrocnemius functions mostly isometrically during a squat, with little or no change in fiber length throughout performance.

Dionisio et al. (15) reported a co-activation of the gastrocnemius and tibialis anterior during the mid-eccentric phase of a squat, presumably to provide stability to the ankle. As previously noted, however, these results were achieved at a high speed of movement. Thus, its application would be specific to protocols designed to achieve speed-strength, including lower-body plyometrics.

Weakness of the ankle musculature has been implicated in the genesis of faulty movement patterns during the squat. Bell et al. (7) found that a lack of strength in the medial gastrocnemius, tibialis anterior, or tibialis posterior may decrease an individual's ability to control knee valgus and foot pronation motions, as well as contributing to excessive medial knee displacement (MKD) and dynamic valgus.

A high degree of mobility at the ankle is required to facilitate balance and control in both the ascent and descent of the squat. When ankle joint flexibility is compromised, there is a tendency for the heels to rise off the floor at higher degrees of flexion. This can result in compensatory joint moments at the ankle, knees, hips, and spine, potentially leading to injury when squatting under external load. Toutoungi et al. (70) reported that ACL forces were significantly increased when squatting was performed with feet flat versus heels elevated during both the descent (26 ± 31 N vs. 95 ± 40 N) and the ascent (28 ± 36 N vs. 49 ± 57 N) of the move. Hemmerich et al. (26) found that a dorsiflexion angle of 38.5 ± 5.9° was necessary to keep the heels down during a full squat.

It has been shown that those exhibiting reduced range of motion at the ankle joint have a predisposition to MKD. In a study by Bell et al. (7), those with MKD displayed a clinically meaningful 20% reduced range of motion in dorsiflexion while squatting–a finding that was attributed in part to tightness of the soleus. In addition, tightness of the lateral ankle musculature was shown to contribute to tibial abduction and external rotation facilitating excessive MKD and dynamic knee valgus alignment (7). This has implications for injury because knee valgus is believed to increase stress on the ACL, especially in combination with internal tibial rotation (42).

Knee Complex

The knee joint consists of the tibiofemoral, which carries out sagittal plane movement throughout a range of motion of 0 to approximately 160° of flexion (36,63,73). The tibiofemoral joint can be classified as a modified hinge joint that comprises the articulation of the tibia and femur. A small amount of axial rotation is also present at the joint during dynamic movement, with the femur rotating laterally during flexion and medially during extension with respect to the tibia. This causes the instant center of rotation at the knee to shift slightly throughout performance of the squat.

Assisting the tibiofemoral joint is the patellofemoral joint, a gliding joint in which the patella slides over the trochlear surfaces of the femur during flexion and extension of the knee. This provides additional mechanical leverage in extension because of a greater force arm, as well as reducing wear on the quadriceps and patellar tendons from friction against the intercondylar groove.

The knee is supported by an array of ligaments and cartilage. Of these structures, the ACL is often considered the most important stabilizer of the joint. Its primary role is to prevent anterior tibial translation at the knee, particularly at low flexion angles (33). It also plays a role in limiting internal and external rotation of the knee and inhibiting varus/valgus motion. The posterior cruciate ligament (PCL) can be considered the counterpart to the ACL. Its primary function is to restrain posterior tibial translation (37). The medial and lateral collateral ligaments stabilize the knee in the frontal plane, helping to provide resistance to varus/valgus moments.

Whereas the knee ligaments are the main static stabilizers of the joint, the knee musculature assumes a dominant role in dynamic joint stabilization (60). During the squat, the primary muscles acting about the knee are the quadriceps femoris (vastus lateralis, vastus medialis, vastus intermedius, and rectus femoris), which carry out concentric knee extension, as well as eccentrically resisting knee flexion. The quadriceps tendon and patellar tendon facilitate action of the knee extensors, allowing for optimal pull on the tibia during dynamic movement.

The hamstrings (biceps femoris, semitendinosus, semimembranosus) are technically antagonists of the quadriceps, opposing knee extensor moments. In closed-chain exercise, however, they behave paradoxically and cocontract with the quadriceps. This synergistic action has important implications for enhancing the integrity of the knee joint in squat performance. Specifically, the hamstrings exert a counterregulatory pull on the tibia, helping to neutralize the anterior tibiofemoral shear imparted by the quadriceps and thus alleviating stress on the ACL (17,56).

Forces at the knee during squat performance have been extensively studied, with most focusing on 3 major areas of interest: a) tibiofemoral compression and shear and patellofemoral compression, b) muscle activity of the quadriceps and hamstrings, and c) anteroposterior and mediolateral knee stability (16).

Both tibiofemoral and patellofemoral compression has been shown to increase with increasing knee angle (23,49,68). It is theorized that these forces provide a protective function at the knee by initiating a cocontraction between the quads and hams. Specifically, the hamstrings exert a counterregulatory force on the tibia by pulling it posteriorly, thereby attenuating anterior tibial translation and counteracting shear (26,56,77).

The highest recorded peak tibiofemoral compression forces were obtained in a study of powerlifters lifting 2.5 times bodyweight. Maximum values reached approximately 8,000 N at 130° of knee flexion and were consistent with maximal forces at the quadriceps tendon (49). These forces slowly declined to 5,500 N at 60° of flexion. At 30° of knee flexion, compressive force amounted to roughly 3,500 N, whereas quadriceps tendon force was reduced to approximately 2,000 N. Patellar tendon force was approximately 6,000 N at 130° and slowly decreased to approximately 2,000 N at 30°. It is important to note that the ultimate tensile strength of the patellar tendon (attaching the patella to the tibial tuberosity) approximates 10,000 to 15,000 N and thus is more than capable of handling these forces (16). Given that the quadriceps tendon (attaching the quadriceps to the patella) is significantly thicker than the patellar tendon, its strength is probably even greater, making the likelihood of exceeding the stress threshold even less.

Mean peak shear forces in the squat have been reported to exceed 2,700 N, with the greatest forces directed posteriorly (16). Forward translation of the knees has been implicated in increased patellofemoral and tibiofemoral shear forces because the tibia slides anteriorly on the femur during flexion. The cruciate ligaments are the main structures responsible for counteracting this shear. However, because the direction of pull in the squat is altered by flexion angle and displacement of tibia, peak ligament forces do not always coincide with peak shear force, nor are the 2 necessarily proportional (68).

Maximum anterior shear forces during the squat tend to occur within the first 60° of knee flexion (16,35). The ACL provides approximately 86% of the restraining force against anterior shear, a product of its role in counteracting anterior translation of the tibia as well as resisting internal and external rotation in early knee flexion (4,10). Peak ACL forces generally occur between 15° and 30° of flexion, decreasing significantly at 60° and leveling off thereafter throughout the range of motion of the joint (30,36,38,57). In a study by Toutoungi et al. (70), the highest ACL force when squatting with heels on the ground was approximately 95 N, equating to only approximately 6% of the ultimate strength of a young, healthy person's ACL. Stress on the ACL during flexion is significantly alleviated by contraction of the hamstrings, which exert a posteriorly directed force on the tibia and thus share in the burden of reducing anterior translation (16,38).

Posterior shear begins to manifest at approximately 30° of flexion, reaching a maximum near the lowest point of the squat (47,68). The PCL exerts the primary restraint against these forces. In a study by Li et al. (36), PCL torque rose significantly with every flexion angle beyond 30° to a peak of 73.2 N at 90°. The PCL forces then decreased significantly from 90° to 120°, leveling off thereafter. This is consistent with other research showing the PCL forces are greatest at 90° of flexion and least near full extension (37). Escamilla (17) reported that peak PCL forces are 30–40% percent greater during the ascent of a squat as compared to the descent. It is unlikely, however, that squatting loads used by the vast majority of the population would be great enough to cause injury to a healthy PCL. The highest PCL forces reported were approximately 2,222 N, which represents just over 50% of estimated strength capacity of a young, healthy person's PCL (50,53,68). Moreover, connective tissue adapts to regimented resistance training by increasing its tolerance level, further reducing the prospect of injury under loaded conditions (9).

Some practitioners have cautioned against performing deep squats, citing an increased potential for injury to soft tissue structures in the knee during high flexion (32). These concerns, however, appear largely unwarranted. Although it is true that shear forces tend to increase with increasing knee angles, forces on the ACL and PCL actually decrease at high flexion. According to Li et al. (35), knee structures are highly constrained at angles greater than 120°, resulting in much less anterior and posterior tibial translation and tibial rotation in comparison with lesser flexion angles. This constraint apparently is caused by impingement between the posterior aspect of the upper tibia with the superior posterior femoral condyles or the compression of soft tissue structures including menisci, posterior capsule, hamstrings fat, and skin (34–36). The upshot is better stability and greater tolerance to load.

Because compressive forces peak at high degrees of knee flexion (13), the greatest risk of injury during deep squatting would appear to be to be to the menisci and articular cartilage, which are placed under increased stress at high flexion angles (16,36). Unfortunately, currently, no guidelines exist to determine at what magnitude of force injury occurs. There also may be a susceptibility to patellofemoral degeneration given the high amount of patellofemoral stress that arises from contact of underside of the patella with articulating aspect of femur during high flexion (17). This can lead to disorders such as chondromolacia, osteoarthritis, and osteochondritis. It is therefore essential to consider an individual's pathologic condition in determining optimal squat depth.

Muscular forces at the knee are largely produced by the quadriceps. Quadriceps activity tends to peak at approximately 80° to 90° of flexion (20,74), remaining relatively constant thereafter. This suggests that squatting past 90° might not result in further enhancements in quadriceps development.

A majority of studies indicate little difference between activity of the vastus lateralis and vastus medialis in the squat, with each providing approximately equal contributions in force output during performance (42,59). Caterisano et al. (12) reported greater electromyography (EMG) output for the vastus medialis oblique in the partial squat when compared with the vastus lateralis, but this finding did not reach statistical significance.

Activity of the vasti, however, has shown to be significantly larger than that of the rectus femoris, producing approximately 50% greater muscular force output (20,29,75). This would appear logical given that the rectus femoris is both a hip flexor and knee extensor and thus shortens at one end while lengthening at the other during the squat, with little if any net change in length throughout movement. Although no corroborating studies could be found, the rectus femoris likely would have a greater advantage in knee extension when the trunk is more erect because this would increase its force/length advantage.

Hip Joint

The hip is a ball-and-socket joint, comprising the articulation between the head of the femur and the acetabulum of the os coxae. It is freely mobile in all 3 planes of movement, carrying out flexion and extension in the sagittal plane, abduction and adduction in the frontal plane, and internal/external rotation and horizontal abduction/adduction in the transverse plane (63,73). Hemmerich et al. (26) reported mean hip range of motion during squatting to be 95 ± 27° of flexion. This implies that athletes may need to improve hip flexibility to perform deep squats.

During the squat, hip torques increase in conjunction with increases in hip flexion, with maximal torque occurring near the bottom phase of movement (49). Fry et al. (22) demonstrated that forward lean has a significant impact on forces about the hip when squatting. Seven recreationally trained males performed 3 unrestricted squat lifts and 3 restricted lifts where a wooden board was placed immediately in front of both feet so that the knees were prevented from moving forward past the toes. Hip torque was significantly increased during restricted squatting as compared with unrestricted squatting (302.7 ± 71.2 vs. 28.2 ± 65.0), and this was attributed to an increased moment arm at the hips caused by compensatory forward lean.

The primary hip muscles involved during the squat are the gluteus maximus (GM) and the hamstrings. The GM is a powerful hip extensor, acting eccentrically to control squat descent and concentrically to overcome external resistance on the ascent. Given its attachment at the iliotibial band, the GM is also thought to play a role in stabilizing knee and pelvis during squatting (56). The force arm of the GM has its smallest values at a 90° hip angle, which would suggest that it has a reduced capacity to produce torque in this range. However, hip extensor force has been shown to peak at approximately 90° (18). This paradox is apparently caused by an optimal force/length relationship in the GM, which overcomes its disadvantage in force arm length by maintaining a sarcomere length more conducive to force production.

Activation of the GM is greatly influenced by squat depth. Caterisano et al. (12) reported that although average muscle activity of the GM was not significantly different in both the partial squat (16.92 ± 8.78%) and parallel squat (28.00 ± 10.29%), it increased significantly during the full squat

(35.47 ± 1.45%). This is also true of peak values, which displayed significantly greater activity during performance of the full squat as compared with lesser squat depths.

The hamstrings have been shown to be only moderately active during squat performance, producing approximately half the amount of EMG activity as during a leg curl and stiff legged deadlift (20,42,74,76). This is consistent with the biarticular structure of the muscle complex. Because the hamstrings function both as hip extensors and knee flexors, their length remains fairly constant throughout performance, allowing for a fairly consistent force output. Peak hamstrings activity has been shown to occur anywhere between 10° and 70° of flexion, with the lateral hamstrings producing greater activity than the medial hamstrings (20,62,74). As opposed to the GM, squat depth does not appear to have any effect on hamstring involvement, with little variation of peak and mean torque between partial squats, parallel squats, and full squats.

Spine

The spine is comprised of 24 mobile vertebral segments, each displaying 3° of freedom. Individually and as a unit, the spine is capable of flexion and extension in the sagittal plane, lateral flexion in the frontal plane, and rotation in the transverse plane (63,73). The vertebral segments display a tapered appearance from top to bottom, with the vertebral bodies becoming progressively larger and thicker from cervical to lumbar regions.

The intervertebral joints are specialized symphysis joints comprising thick pads of fibrocartilage called intervertebral discs sandwiched between 2 vertebral bodies (63,73). Each disc consists of an outer fibrous ring called the annulus fibrosus, which surrounds a gel-like inner mass called the nucleus pulposus. These discs serve to hold the vertebrae together as well as allowing for dynamic spinal movement.

The vertebral column is supported by an array of muscles, including the erector spinae, transversus abdominis, quadratus lumborum, and deep posterior spinal group (multifidus, rotatores, interspinales, and intertransversarii). The lumbar erector spinae (e.g., iliocostalis, longissimus) are particularly important during the squat because they help to resist vertebral shear and maintain anteroposterior spinal integrity, providing the greatest contribution to spinal stabilization (70).

Proper squat technique requires a rigid spine that eliminates any planar motion. This ensures that a stable, upright posture is maintained throughout movement. However, given the synergistic lumbar-pelvic relationship, absolute spinal angle will generally increase as an individual flexes at the hips. Therefore, the vertebral column and its supporting muscles are subjected to significant internal forces during performance of the lift, especially in deeper squats.

Cappozzo et al. (11) determined that a half-squat with a barbell load between 0.8 to 1.6 times bodyweight produced compressive forces on the L3-L4 segment equating to 6 to 10 times bodyweight, with forces increasing with increases in external load. Considering that the ultimate compressive strength in individuals 40 years and younger is estimated to be approximately 7,800 N (3), this would imply that many athletes are routinely squatting at or above their threshold for spinal failure. Because failure of the vertebrae does not occur in the vast majority of cases, it can be postulated that the trained athlete's spine adapts to mechanical stress by an increased bone modeling, thereby increasing compressive tolerance.

Spinal flexion and extension have shown to significantly impact joint kinetics during squat performance. Squatting with a flexed lumbar spine decreases the moment arm for the lumbar erector spinae, reduces tolerance to compressive load, and results in a transfer of the load from muscles to passive tissues, heightening the risk of disc herniation (44). Moreover, shear forces during squatting have been found to be significantly higher as lumbar flexion increases from the neutral position (52). This is accompanied by changes in the fiber orientation of the erector spinae that reduce their ability to counteract shear.

Alternatively, studies show that compressive forces increase when the spine is held in excessive lumbar extension. Adams et al. (1) reported that a 2° increase in extension from neutral position heightens compressive forces in the posterior annulus by a clinically meaningful average of 16%. It is therefore advisable to maintain a neutral spine throughout performance of the squat, avoiding any excessive spinal flexion or extension. Moreover, because lumbar forces are increased with increases in forward lean (55), it is beneficial to maintain a posture that is as close to upright as possible at all times.

Increasing intra-abdominal pressure (IAP) may serve to alleviate vertebral forces. According to Vakos et al. (72), an increased IAP creates a "balloon" anterior to the spine that resists compression. In addition, it provides an antiflexion moment in the lumbar region that reduces active contraction of the erector spinae, thereby diminishing spinal compression generated by associated muscle tension. McGill et al. (46) reported that when subjects increased IAP by holding their breath during a squat lift of 72.7 to 90.9 kg, lumbar load was significantly reduced. Miyamoto et al. (48) reported that an increased IAP raises intramuscular pressure of the erector spinae muscles and stiffens the trunk, contributing to greater spinal stabilization during dynamic lifts.

The direction of gaze also has been found to influence spinal kinetics and kinematics. A study by Donnelly et al. (16) showed that a downward gaze increases trunk flexion by 4.5° and hip flexion by approximately 8° compared with a straight ahead or upward gaze (15). Given that excessive trunk and hip flexion can place excessive torque on the vertebral column, this suggests it is beneficial to maintain either a straight ahead or upward gaze during performance of the squat.

Confounding Performance Variables

Intensity of Load

Squats can be performed using just one's bodyweight or with an external load. Sahli et al. (58) explored the effect of squatting with a variety of different external loads on tibiofemoral joint forces and found peak compression and shear increased significantly in conjunction with increasing loads. Peak compressive force was found to be 149% of bodyweight with a load equal to 120% body weight (BW) as compared with a compressive force of 58% BW under unloaded conditions. Peak mediolateral and anteroposterior shear force increased from 8% BW to 11% BW and from 46–67% BW, respectively, when external load increased from 50% BW to 100%. These findings are in accord with Markolf et al. (43), who found that a simulated tendon load increased ACL force at every knee flexion angle. These increased forces, however, should be of little concern to those with healthy knees. Studies show that even in powerlifters using loads more than twice bodyweight, tensile forces in the PCL and ACL only reached approximately 50% and 25%, respectively, of their ultimate estimated strength potential (20,74).

A factor of greater clinical importance during squatting is the effect of load on spinal kinetics and kinematics. Kellis et al. (32) found that the absolute angle of the spine increased a nonsignificant 6° from when subjects lifted a load up to 32% of their 1 repetition maximum (1RM) (31). However, when using a weight between 40–70% 1RM, a significant 16° increase in forward inclination was noted. Hay et al. (25) reported similar findings, with a significant increase in trunk forward lean displayed when subjects lifted a load between 40% and 80% of 4RM. Walsh et al. (74) reported a linear correlation between spinal compression and load, as well as a significant degree of hyperextension when subjects lifted at heavier weights (60% and 80% 1RM) attributed to a compensatory action to stabilize the body from falling forward. These findings reinforce the need for proper spinal alignment during the performance of loaded squats so that excessive forces are not placed on the lumbar region.

Foot Placement

Foot placement has been shown to affect squatting kinetics. Escamilla et al. (18) reported statistically significant 15% and 16% increases in patellofemoral and tibiofemoral compressive forces, respectively, in subjects who squatted with a wide stance (defined as 87 to 118% of shoulder width) as compared with a narrow stance (defined as 158 to 196% of shoulder width). In addition, the squat descent generated significantly greater compressive forces than the ascent at higher angles of knee flexion, whereas the ascent produced significantly greater compressive forces than the descent at lower flexion angles. This suggests that a narrow stance may be preferable over wide stance if the goal is to minimize compressive forces at the knee. On the other hand, a narrow stance squat resulted in approximately 4 to 6 cm greater forward knee translation and thus greater shear as compared with a moderate or wide stance. Therefore, a wider stance might be preferable for those seeking to minimize shear at the knees (18).

Several studies reveal that varying squat stance alters muscular recruitment patterns. Escamilla et al. (18) found activity of the gastrocnemius was 21% greater in a narrow versus a wide stance. McCaw and Melrose (45) reported a wide stance significantly increased activity of the GM and adductor longus, with greatest activity seen at 140% shoulder width. Ninos et al., Paoli et al., and Escamilla et al. (20,50,53) also reported increased muscular torque of the hip extensors and adductors in wider stance squats. Stance width has not been shown to alter muscular activity in the quadriceps and hamstrings, however (20,42).

Studies examining the effect of altering foot position (i.e., tibial rotation) during the squat have shown little if any variation in muscle activity of the quadriceps, GM, hamstrings, or gastrocnemius from 30° inward rotation to 80° outward rotation (27,48,61). Given that extreme rotation of the tibia can change normal patella tracking and potentially cause undesirable varus or valgus moments, it appears prudent to avoid exaggerated foot positions in closed chain movements such as the squat.

Speed of Movement

Studies examining the impact of squat cadence on joint kinetics reveal a positive correlation between faster lifting speeds and joint forces. Hattin et al. (24) had subjects perform repetitions of a half-squat lasting 1 second or 2 seconds using an external load between 15 to 30% of 1RM. The faster cadence significantly increased anteroposterior shear and compressive forces at the knee (50% and 28%, respectively), as well as displaying a trend toward heightened mediolateral shear. These findings concur with those of Dahlkvist et al. (14), who also reported an increase in tibiofemoral joint forces at higher speed of movement. Bouncing at the bottom of squat, which often accompanies fast movement patterns, has been shown to increase knee shear forces by an additional 33% (16). Moreover, peak compressive forces at the spine double if weights are lifted rapidly (72). Thus, although a faster speed of movement can be beneficial for transfer to many sporting activities, a slower cadence would be advisable for those seeking to reduce joint-related shear and compressive forces.

Fatigue

Fatigue can have a significant effect on squat kinetics and kinematics. The onset of fatigue can cause undue alterations in squatting technique, which is likely a contributing factor to both short- and long-term injuries (72). In a study by Lattanzio et al. (34), fatigue caused a significant reduction on knee proprioception, presumably because of a decrease in muscle and joint proprioceptor activity. The authors went on to conclude that exercise at or near exhaustion may reduce ligamental mechanoreceptor function, potentially leading to knee instability.

The spine is particularly vulnerable to the effects of fatigue. Failure of the vertebral body occurs at much lower forces when subjected to fatigue, and their compressive strength is reduced up to 30% after 10 loading cycles (2,8). Trafimow et al. (71) demonstrated that subjects changed their squatting technique from an upright to a flexed lumbar spine after fatigue of the quadriceps, thereby placing increased stress on the lumbar region. Similar results were found by Sasaki et al. (60), who reported that substantial fatigue of the quadriceps leads to an increased low back load and a subsequent elevated risk of lumbar injury. Given these findings, it is essential to be cognizant of spinal mechanics as the lower-body muscles fatigue and cease the lift if form becomes compromised. This is particularly important when performing repetition maximum testing, where repeated lifts to determine exact 1RM can sometimes outweigh the potential risks.

Position of Load

Weighted squats can be performed with the external load placed in a variety of positions. The most studied variations are low bar back squats with the bar slightly below the level of the acromion, high bar back squats with the bar slightly above the level of the acromion, and barbell front squats with the bar held in front of the chest at the clavicle (16).

Because of a greater forward inclination of the trunk, the low bar position typical of powerlifters has been shown to produce greater hip extensor torque and less knee extensor torque compared with high bar squat typical of weight lifters (75). This translates into reduced patellofemoral compression and ACL strain in the low bar squat. However, values do not come close to exceeding the strength threshold of these structures in either bar position. Thus, unless contraindicated by an existing injury, both positions are suitable for the majority of lifters.

Gullett et al. (23) studied differences in kinetics between front squats and back squats. Front squats were found to produce significantly lower maximal joint compressive forces at the knee as well as reduced lumbar stress as compared with back squats, with little difference noted in shear forces. This was accomplished without compromising muscle activity in the quadriceps and hamstrings. This suggests that front squats may be a better alternative than back squats for those with ligament or meniscal injuries. What is more, the front squat may isolate the quadriceps to a greater degree than the back squat, making it a viable choice for those seeking to optimize development of the frontal thighs in comparison with the gluteals.

Practical Applications

On the basis of the kinematic and kinetic data reviewed, the following recommendations can be made to ensure optimal squatting performance and safety:

1. Squat depth should be consistent with the goals and abilities of the individual. Because peak patellofemoral compressive forces occur at or near maximum knee flexion, those with patellofemoral disorders should avoid squatting at high flexion angles (16). For those with existing injury or previous reconstruction of the PCL, it is best to restrict flexion to 50° to 60° so that posterior shear is minimized (16). Quadriceps development is maximized by squatting to parallel, with no additional activity seen at higher flexion angles (75). Hip extensor moments increase with increasing squat depth, so full squats may be beneficial for those seeking to maximize strength of the hip musculature.
2. Speed of movement should be based on goal-oriented specificity to the force-velocity curve (6). However, given that speed of movement has been shown to significantly increase both compression and shear forces, there is a tradeoff between optimal transfer of performance and risk of injury. This is especially true on the eccentric aspect of the move where rapid deceleration generates exceedingly high joint forces at the knee. Failure to control descent can result in the ballistic contact between the hamstrings and calf muscles, which can cause a dislocating effect on the knee ligaments (16). Therefore, unless athletic goals specifically dictate otherwise, squat descent should always be executed in a controlled fashion, with a 2 to 3 second eccentric tempo considered a general guideline.
3. A wider stance squat is preferable for those seeking optimal development of the hip adductors and hip extensors (42), whereas a closer stance is more appropriate for targeting development of the gastrocnemius (18). Stance can also be varied to alter joint-related forces: a narrow stance helps to minimize patellofemoral and tibiofemoral compression while a wider stance results in less forward knee translation and thus reduces shear (18).
4. Low bar back squats tend to produce greater hip extensor torque and less knee extensor torque compared with high bar back squats. However, the magnitude of forces for both movements are well tolerated by the associated joint structures, making either position suitable for the majority of lifters (75). The front squat produces significantly lower knee compression and lumbar stress in comparison with back squats, making it a viable alternative for those suffering from various knee and back ailments (22). Front squats also can be particularly beneficial for those competing in weight lifting events because it is an essential component in performance of the clean.
5. Fatigue can have a deleterious effect on squatting technique, potentially leading to knee instability (32) and increased lumbar shear (69). If the lifter opts to squat to momentary muscular failure, it is advisable to have a spotter to ensure safety.

In addition to the aforementioned joint-specific recommendations, some joint-specific recommendations can be made as to squat-related performance variables.

Ankle Joint. Significant strength and mobility is required at the ankle for proper squat performance. Feet should be positioned in a comfortable stance that allows the knees to move in line with the toes (17). Because the feet are outwardly rotated approximately 7° in anatomic position, this can be considered a good starting point to ensure proper patellar tracking. If the lifter's heels rise off the floor during the eccentric phase of movement, efforts should be made to improve flexibility around the talocrural and subtalar joints. Orthotics can be worn to help correct joint imbalances and misalignment. If necessary, a barbell plate or other flat object can be placed underneath the heels to aid in stability.

Knee Joint. Given the fact that shear forces are increased as the knees move past the toes during the downward phase of the squat, attempts should be made to avoid significant forward knee translation on descent. However, this should not be done at the expense of compromising form at the hips and spine, which can place the lumbar region in a biomechanically disadvantageous position and significantly increase spinal shear (21). To reduce tibiofemoral and patellofemoral moments, the lifter should sit back into the squat during descent and resist pushing the knees forward. There should be no varus or valgus motion throughout exercise performance.

Hip Joint. Given the close relationship between movement at the hips, pelvis, and lumbar spine during dynamic squatting, hip mobility is extremely important for proper squat performance, especially at higher flexion angles. Poor joint mobility can lead to greater forward lean and thus increased spinal shear. Although some lifters attempt to increase hip flexion by using posterior pelvic movement during squat descent, this can heighten lumbar stress and is thus not advisable. Flexibility training specific to the hip musculature can help to increase hip mobility and facilitate better squat performance.

Spine. The spine is the most vulnerable of the joints during squatting. Because the lumbar spine is better able to handle compressive force than shear, a normal lordotic curve should be maintained in this region, with the spinal column held rigid throughout the movement (70). Proper spinal alignment is facilitated by maintaining a straight ahead or upward gaze, which reduces the tendency for unwanted flexion (15). Although some forward lean is sometimes necessary to maintain stability especially when performing deep squats, attempts should be made to keep the trunk as upright as possible to minimize shear. No lateral movement should take place at any time.

References

1. Adams, MA and Dolan, P. Forces acting on the lumbar spine. In: *Lumbar Spine Disorders: Current Concepts.* Aspden, RM and Porter, RW, eds. Singapore: World Scientific Publishing, 1995. pp. 22.
2. Adams, MA and Dolan, P. Recent advances in lumbar spine mechanics and their clinical significance. *Clin Biomech* 10: 3–19, 1995.
3. Adams, MA, May, S, Freeman, BJ, Morrison, HP, and Dolan, P. Effects of backward bending on lumbar intervertebral discs: Relevance to physical therapy treatments for low back pain. *Spine* 25: 431–437, 2000.
4. Andersen, HN and Dyhre-Poulsen, P. The anterior cruciate ligament does play a role in controlling axial rotation in the knee. *Knee Surg Sports Traumatol Arthrosc* 5: 145–149, 1997.
5. Ariel, BG. Biomechanical analysis of the knee joint during deep knee bends with heavy load. In: *Biomechanics IV.* Nelson, RC and Morehouse, C, eds. Baltimore: University Park Press, 1974. pp. 44–52.
6. Behm, DG and Sale, DG. Velocity specificity of resistance training. *Sports Med* 15: 374–388, 1993.
7. Bell, DR, Padua, DA, and Clark, MA. Muscle strength and flexibility characteristics of people displaying excessive medial knee displacement. *Arch Phys Med Rehabil* 89: 1323–1328, 2008.
8. Brinckmann, P, Biggermann, M, and Hilweg, D. Fatigue fracture of human lumbar vertebrae. *Clin Biomech* 3(Suppl 1): 1–23, 1988.
9. Buchanan, CI and Marsh, RL. Effects of exercise on the biomechanical, biochemical and structural properties of tendons. *Comp Biochem Physiol A Mol Integr Physiol* 133: 1101–1107, 2002.
10. Butler, DL, Noyes, FR, and Grood, ES. Ligamentous restraints to anterior-posterior drawer in the human knee. *J Bone Joint Surg* 62: 259–270, 1980.
11. Cappozzo, A, Felici, F, Figura, F, and Gazzani, F. Lumbar spine loading during half-squat exercises. *Med Sci Sports Exerc* 17: 613–620, 1985.
12. Caterisano, A, Moss, RF, Pellinger, TK, Woodruff, K, Lewis, VC, Booth, W, and Khadra, T. The effect of back squat depth on the EMG activity of 4 superficial hip and thigh muscles. *J Strength Cond Res* 16: 428–432, 2002.
13. Clarkson, HM and Gilewich, GB. *Musculoskeletal Assessment: Joint Range Of Motion And Manual Muscle Strength.* Baltimore, MD: Williams and Wilkins, 1999. pp. 374.
14. Dahlkvist, NJ, Mayo, P, and Seedhom, BB. Forces during squatting and rising from a deep squat. *Eng Med* 11: 69–76, 1982.
15. Dionisio, VC, Almeida, GL, Duarte, M, and Hirata, RP. Kinematic, kinetic and EMG patterns during downward squatting. *J Electromyogr Kinesiol* 18: 134–143, 2008.
16. Donnelly, DV, Berg, WP, and Fiske, DM. The effect of the direction of gaze on the kinematics of the squat exercise. *J Strength Cond Res* 20: 145–150, 2006.
17. Escamilla, RF. Knee biomechanics of the dynamic squat exercise. *Med Sci Sports Exerc* 33: 127–141, 2001.
18. Escamilla, RF, Fleisig, GS, Lowry, TM, Barrentine, SW, and Andrews, JR. A three-dimensional biomechanical analysis of the squat during varying stance widths. *Med Sci Sports Exerc* 33: 984–998, 2001b.
19. Escamilla, RF, Fleisig, GS, Zheng, N, Barrentine, SW, Wilk, KE, and Andrews, JR. Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Med Sci Sports Exerc* 30: 556–569, 1998.
20. Escamilla, RF, Fleisig, GS, Zheng, N, Lander, JE, Barrentine, SW, Andrews, JR, Bergemann, BW, and Moorman, CT. Effects of technique variations on knee biomechanics during the squat and leg press. *Med Sci Sports Exerc* 33: 1552–1566, 2001a.
21. Escamilla, RF, Lander, JE, and Garhammer, J. Biomechanics of powerlifting and weightflifting exercises. In: *Exercise and Sport Science.* Garrett, WE and Kirkendall, DT, eds. Philadelphia, PA: Lippincott Williams and Wilkins, 2000. pp. 585–615.
22. Fry, AC, Smith, JC, and Schilling, BK. Effect of knee position on hip and knee torques during the barbell squat. *J Strength Cond Res* 17: 629–633, 2003.
23. Gullett, JC, Tillman, MD, Gutierrez, GM, and Chow, JW. A biomechanical comparison of back and front squats in healthy trained individuals. *J Strength Cond Res* 23: 284–292, 2009.
24. Hattin, HC, Pierrynowski, MR, and Ball, KA. Effect of load, cadence, and fatigue on tibio-femoral joint force during a half squat. *Med Sci Sports Exerc* 21: 613–618, 1989.

25. Hay, JG, Andrews, JG, Vaughan, CL, and Ueya, K. Load, speed and equipment effects in strength-training exercises. In: *Biomechanics VIII-B*. Matsui, H and Kobayashi, K, eds. Champaign, IL: Human Kinetics Publishers, 1983. pp. 939–950.
26. Hemmerich, A, Brown, H, Smith, S, Marthandam, SS, and Wyss, UP. Hip, knee, and ankle kinematics of high range of motion activities of daily living. *J Orthop Res* 24: 770–781, 2006.
27. Hsieh, HH and Walker, PS. Stabilizing mechanisms of the loaded and unloaded knee joint. *J Bone Joint Surg* 58: 87–93, 1976.
28. Hung, YJ and Gross, MT. Effect of foot position on electromyographic activity of the vastus medialis oblique and vastus lateralis during lower-extremity weight-bearing activities. *J Orthop Sports Phys Ther* 29: 93–102, 1999.
29. Hwang, S, Kim, Y, and Kim, Y.Lower extremity joint kinetics and lumbar curvature during squat and stoop lifting. *BMC Musculoskel Disord* 2: 10–15, 2009.
30. Isear, JA Jr, Erickson, JC, and Worrell, TW. EMG analysis of lower extremity muscle recruitment patterns during an unloaded squat. *Med Sci Sports Exerc* 29: 532–539, 1997.
31. Kanamori, A, Woo, SL, Ma, CB, Zeminski, J, Rudy, TW, Li, G, and Livesay, GA. The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology. *Arthroscopy* 16: 633–639, 2000.
32. Kellis, E, Arambatzi, F, and Papadopoulos, C. Effects of load on ground reaction force and lower limb kinematics during concentric squats. *J Sports Sci* 23: 1045–1055, 2005.
33. Klein, K. The deep squat exercise as utilized in weight training for athletes and its effects on the ligaments of the knee. *J Assoc Phys Ment Rehabil* 15: 6–11, 1961.
34. Lattanzio, PJ, Petrella, RJ, Sproule, JR, and Fowler, PJ. Effects of fatigue on knee proprioception. *Clin J Sports Med* 7: 22–27, 1997.
35. Li, G, DeFrate, LE, Rubash, HE, and Gill, TJ. In vivo kinematics of the ACL during weight bearing knee flexion. *J Orthop Res* 23: 340–344, 2005.
36. Li, G, Most, E, DeFrate, LE, Suggs, JF, Gill, TJ, and Rubash, HE. Effect of the posterior cruciate ligament on posterior stability of the knee in high flexion. *J Biomech* 37: 779–783, 2004a.
37. Li, G, Rudy, TW, Sakane, M, Kanamori, A, Ma, CB, and Woo, SL. The importance of quadriceps and hamstring muscleloading on knee kinematics and in-situ forces in the ACL. *J Biomech* 32: 395–400, 1999.
38. Li, G, Zayontz, S, DeFrate, LE, Most, E, Suggs, JF, and Rubash, HE. Kinematics of the knee at high flexion angles: An in vitro investigation. *J Orthop Res* 27: 699–706, 2004b.
39. Li, G, Zayontz, S, Most, E, DeFrate, LE, Suggs, JF, and Rubash, HE. In situ forces of the anterior and posterior cruciate ligaments in high knee flexion: an in vitro investigation. *J Orthop Res* 22: 293–297, 2004c.
40. Lutz, GE, Palmitier, RA, An, KN, and Chao, EY. Comparison of tibiofemoral joint forces during open kinetic-chain and closed-kinetic-chain exercises. *J Bone Joint Surg* 75: 732–739, 1993.
41. Markolf, KL, Burchfield, DM, Shapiro, MM, Shepard, MF, Finerman, GA, and Slauterbeck, JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res* 13: 930–935, 1995.
42. Markolf, KL, Gorek, JF, Kabo, JM, and Shapiro, MS. Direct measurement of resultant forces in the anterior cruciate ligament: An in vitro study performed with a new experimental technique. *J Bone Joint Surg* 72: 557–567, 1990.
43. Markolf, KL, Slauterbeck, JL, Armstrong, KL, Shapiro, MM, and Finerman, GA. Effects of combined knee loadings on posterior cruciate ligament force generation. *J Orthop Res* 14: 633–638, 1996.
44. Matsumoto, H, Suda, Y, Otani, T, Niki, Y, Seedhom, BB, and Fujikawa, K. Roles of the anterior cruciate ligament and the medial collateral ligament in preventing valgus instability. *J Orthop Sci* 6: 28–32, 2001.
45. McCaw, ST and Melrose, DR. Stance width and bar load effects on leg muscle activity during the parallel squat. *Med Sci Sports Exerc* 31: 428–436, 1999.
46. McGill, S, Norman, RW, and Sharatt, MT. The effect of an abdominal belt on trunk muscle activity and intra- abdominal pressure during squat lifts. *Ergonomics* 33: 147–160, 1990.
47. Mirzabeigi, E, Jordan, C, Gronley, JK, Rockowitz, NL, and Perry, J. Isolation of the vastus medialis oblique muscle during exercise. *Am J Sports Med* 27: 50–53, 1999.
48. Miyamoto, K, Iinuma, N, Maeda, M, Wada, E, and Shimizu, K. Effects of abdominal belts on intra-abdominal pressure, intra-muscular pressure in the erector spinae muscles and myoelectrical activities of trunk muscles. *Clin Biomech* 14: 79–87, 1999.
49. Nagura, T, Dyrby, CO, Alexander, EJ, and Andriacchi, TP. Mechanical loads at the knee joint during deep flexion. *J Orthop Res* 20: 881–886, 2002.
50. Ninos, JC, Irrgang, JJ, Burdett, R, and Weiss, JR. Electromyographic analysis of the squat performed in self-selected lower extremity neutral rotation and 30° of lower extremity turn-out from the self-selected neutral position. *J Orthop Sports Phys Ther* 25: 307–315, 1997.
51. Nisell, R and Ekholm, J. Joint load during the parallel squat in powerlifting and force analysis of in vivo bilateral quadriceps tendon rupture. *Scand J Sports Sci* 8: 63–70, 1986.
52. Noyes, FR, Butler, DL, Grood, ES, Zernicke, RF, and Hefzy, MS. Biomechanical analysis of human ligament grafts used in knee ligament repairs and reconstructions. *J Bone Joint Surg* 66A: 344–352, 1984.
53. Paoli, A, Marcolin, G, and Petrone, N. The effect of stance width on the electromyographical activity of eight superficial thigh muscles during back squat with different bar loads. *J Strength Cond Res* 23: 246–250, 2009.
54. Potvin, JR, McGill, SM, and Norman, RW. Trunk muscle and lumbar ligament contributions to dynamic lifts with varying degrees of trunk flexion. *Spine* 16: 1099–107, 1991.
55. Race, A and Amis, AA. The mechanical properties of the two bundles of the human posterior cruciate ligament. *J Biomech* 27: 13–24, 1994.
56. Rasch, PJ and Burke, RK. *Kinesiology and Applied Anatomy* (5th ed.). Philadelphia, PA: Lea and Febiger, 1974.
57. Russell, PJ and Phillips, SJ. A preliminary comparison of front and back squat exercise. *Res Q Exerc Sport* 60: 201–208, 1989.
58. Sahli, S, Rebai, H, Elleuch, MH, Tabka, Z, and Poumarat, G. Tibiofemoral joint kinetics during squatting with increasing external load. *J Sport Rehabil* 17: 300–315, 2008.
59. Sakane, M, Fox, RJ, Woo, SL, Livesay, GA, Li, G, and Fu, FH. In situ forces in the anterior cruciate ligament and its bundles in response to anterior tibial loads. *J Orthop Res* 15: 285–293, 1997.
60. Sasaki, M, Horio, H, Wakasa, M, Uemura, S, and Osawa, Y. Influence of quadriceps femoris fatigue on low back load during lifting of loads at different distances from the toes. *J Phys Ther Sci* 20: 81–89, 2008.
61. Schaub, P and Worrell, T. EMG activity of six muscles and VMO:VL ratio determination maximal squat exercise. *J Sport Rehabil* 4: 195–202, 1995.
62. Senter, C and Hame, SL. Biomechanical analysis of tibial torque and knee flexion angle: Implications for understanding knee injury. *Sports Med* 36: 635–641, 2006.
63. Signorile, JF, Kwiatkowksi, K, Caruso, JF, and Robertson, B. Effect of foot position on the electromyographical activity of the superficial quadriceps muscles during the parallel squat and knee extension. *J Strength Cond Res* 9: 182–187, 1995.
64. Signorile, JF, Weber, B, Roll, B, Caruso John, Lowensteyn, I, and Perry, AC. An electromyographical comparison of the squat and knee extension exercises. *J Strength Cond Res* 8: 178–183, 1994.
65. Smith, LK, Weiss, EL, and Lehmkuhl, LD. *Brunnstrom's Clinical Kinesiology* (5th ed.). Philadelphia, PA: F.A. Davis Company, 1996.

66. Solomonow, M, Baratta, R, Zhou, BH, Shoji, H, Bose, W, Beck, C, and D'Ambrosia, R. The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *Am J Sports Med* 15: 207–213, 1987.

67. Stiene, HA, Brosky, T, Reinking, MF, Nyland, J, and Mason, MB. Comparison of closed kinetic chain and isokinetic joint isolation exercise in patients with patellofemoral dysfunction. *J Orthop Sports Phys Ther* 24: 136–141, 1996.

68. Stoppani, J. *Encyclopedia of Muscle and Strength.* Champaign, IL: Human Kinetics Publishers, 2006. pp. 151.

69. Stuart, MJ, Meglan, DA, Lutz, GE, Growney, ES, and An, KN. Comparison of intersegmental tibiofemoral joint forces and muscle activity during various closed kinetic chains exercises. *Am J Sports Med* 24: 792–799, 1996.

70. Toutoungi, DE, Lu, TW, Leardini, A, Catani, F, and O'Connor, JJ. Cruciate ligament forces in the human knee during rehabilitation exercises. *Clin Biomech* 15: 176–187, 2000.

71. Trafimow, JH, Schipplein, OD, Novak, GJ, and Andersson, GB. The effects of quadriceps fatigue on the technique of lifting. *Spine* 18: 364–367, 1993.

72. Vakos, JP, Nitz, AJ, Threlkeld, AJ, Shapiro, R, and Horn, T. Electromyographic activity of selected trunk and hip muscles during a squat lift. Effect of varying the lumbar posture. *Spine* 19: 687–695, 1994.

73. van Eijden, TM, Weijs, WA, Kouwenhoven, E, and Verburg, J. Forces acting on the patella during maximal voluntary contraction of the quadriceps femoris muscle at different knee flexion/extension angles. *Acta Anat* 129: 310–314, 1987.

74. Walsh, JC, Quinlan, JF, Stapleton, R, FitzPatrick, DP, and McCormack, D. Three-dimensional motion analysis of the lumbar spine during "free squat" weight lift training. *Am J Sports Med* 35: 927–932, 2007.

75. Watkins, J. Structure and Function of the Musculoskeletal System. Champaign, IL: Human Kinetics Publishers, 1999.

76. Wilk, KE, Escamilla, RF, Fleisig, GS, Barrentine, SW, Andrews, JR, and Boyd, ML. A comparison of tibiofemoral joint forces and electromyographic activity during open and closed kinetic chain exercises. *Am J Sports Med* 24: 518–527, 1996.

77. Wretenberg, P, Feng, Y, Lindberg, F, and Arborelius, W. Joint moments of force and quadriceps activity during squatting exercise. *Scand J Med Sci Sports* 3: 244–250, 1993.

78. Wright, GA, Delong, TH, and Gehlsen, G. Electromyographic activity of the hamstrings during performance of the leg curl, stiff-leg deadlift, and back squat movements. *J Strength Cond Res* 13: 168–174, 1999.

79. Yack, HJ, Washco, LA, and Whieldon, T. Compressive forces as a limiting factor of anterior tibial translation in the ACL-deficient knee. *Clin J Sports Med* 4: 233–239.