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REVIEW ARTICLE

Analysis of the Load on the Knee Joint and Vertebral Column with Changes in Squatting Depth and Weight Load

Hagen Hartmann · Klaus Wirth · Markus Klusemann

Abstract It has been suggested that deep squats could cause an increased injury risk of the lumbar spine and the knee joints. Avoiding deep flexion has been recommended to minimize the magnitude of knee-joint forces. Unfortunately this suggestion has not taken the influence of the wrapping effect, functional adaptations and soft tissue contact between the back of thigh and calf into account. The aim of this literature review is to assess whether squats with less knee flexion (half/quarter squats) are safer on the musculoskeletal system than deep squats. A search of relevant scientific publications was conducted between March 2011 and January 2013 using PubMed. Over 164 articles were included in the review. There are no realistic estimations of knee-joint forces for knee-flexion angles beyond 50° in the deep squat. Based on biomechanical calculations and measurements of cadaver knee joints, the highest retropatellar compressive forces and stresses can be seen at 90°. With increasing flexion, the wrapping effect contributes to an enhanced load distribution and enhanced force transfer with lower retropatellar compressive forces. Additionally, with further flexion of the knee joint a cranial displacement of facet contact areas with continuous enlargement of the retropatellar articulating surface occurs. Both lead to lower retropatellar compressive stresses. Menisci and cartilage, ligaments and bones are susceptible to anabolic metabolic processes and functional structural adaptations in response to increased activity and mechanical influences. Concerns about degenerative changes of the tendofemoral complex and the apparent higher risk for chondromalacia, osteoarthritis, and osteochondritis in deep squats are unfounded. With the same load configuration as in the deep squat, half and quarter squat training with comparatively supra-maximal loads will favour degenerative changes in the knee joints and spinal joints in the long term. Provided that technique is learned accurately under expert supervision and with progressive training loads, the deep squat presents an effective training exercise for protection against injuries and strengthening of the lower extremity. Contrary to commonly voiced concern, deep squats do not contribute increased risk of injury to passive tissues.

H. Hartmann (✉) · K. Wirth
Department of Human Movement Science and Athletic Training, Institute of Sports Sciences, Goethe-University, Ginnheimer Landstr. 39, 60487 Frankfurt am Main, Germany
e-mail: Hagen-Hartmann@online.de

M. Klusemann
Physiology Department, Australian Institute of Sport (AIS), Charles Sturt University, Leverrier Crescent, Bruce, ACT 2617, Australia

1 Introduction

The primary objective of performance-oriented strength training is the gain of muscle cross-sectional area (hypertrophy strength training) or the enhancement of inter- and intramuscular coordination (strength/power training), for which high training intensities are necessary [1]. For this purpose, the squat is one of the most effective exercises in athletic training. The Olympic barbell squat can be classified into three fundamental variations: the front squat [2], the high-bar back squat, and the low-bar back squat [2, 3] (Fig. 1). In the parallel squat, the knees are flexed until the inguinal fold is in a straight horizontal line with the top of the knee musculature [2] (Fig. 2). Depending on the squat variant, the knee angles vary between 60° and 70° in this position [2, 3]. The half squat is performed at 80°–100° [4, 5] (Fig. 3), and the quarter squat is executed at a knee angle

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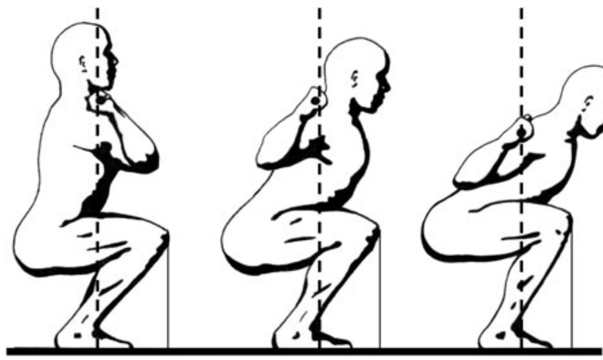

Fig. 1 Bar position determines trunk angle as demonstrated for the front squat (weightlifting style), the high-bar back squat, and the low-bar back squat. In the front squat the barbell is held in pronated posture and is positioned above the sternoclavicular joints, in the high-bar back squat, the barbell is positioned on the trapezius pars descendens below the seventh cervical vertebra. In the low-bar back squat, the weight is placed on the deltoid pars spinales. The bar remains balanced over the mid-foot in each variant and this requires that the trunk angle has to accommodate the bar position. This is the primary factor in the differences in technique between the three squatting variants (adapted from Rippetoe and Kilgore [7], with permission)

of 110°–140° [5, 6] (Fig. 4). The deep squat is performed at 40°–45° [3, 4] (Figs. 5, 6).

On the basis of electromyographic studies, Escamilla [8] and Schoenfeld [9] suggest that squatting depths below 90° do not induce greater neural stimuli of the quadriceps, because the activity signals remained constant. Abandonment of deep squats minimizes the magnitude of tibio- and patellofemoral compressive stress as well. Compared with deep squats, the quarter and half squat necessitate higher weights to induce effectual training stimuli to the hip and leg extensors due to the advantage of higher strength-developing conditions at larger extension angles. With increasing loads, a linear rise of compression load of the vertebral bodies [10] and of the intradiscal pressure occurs [11]. Higher weights require a greater degree of torso stabilization to counteract impairing spinal-shear forces. Higher weights result in increased tibio- [12, 13] and patellofemoral compressive forces [14]. These relationships have been ignored in recent publications that have discussed spinal- [9] and knee-joint forces [8, 9] at different squatting depths. It is unclear if utilizing half or quarter squats places less stress on the various joints used within the squatting motion than deep squats. In this article a literature-based load analysis of different squatting variants on spinal- and knee-joint stress is presented to assess whether squats with less knee flexion (half/quarter squats) are safer on the musculoskeletal system than deep squats.

Fig. 2 Parallel back squat

Fig. 3 Half back squat

2 Literature Search Methodology

A literature search of PubMed was conducted for the period from March 2011 to January 2013 (http://www.ncbi.nlm.nih.gov/pubmed). The main search terms were: 'weightlifting knee', 'squat knee', 'squat knee stress', 'squat knee force', 'squat knee biomechanics', 'squat patellofemoral biomechanics', 'squat tibiofemoral joint biomechanics', 'weightlifting spine', 'squat spine', 'exercise tendon adaptation', 'exercise ligament adaptation', 'weightlifting bone mineral density', 'in vivo measurements of intradiscal pressure', 'weightlifting injuries', 'mechanical loading cartilage explants', 'in vitro

Fig. 4 Quarter back squat

Fig. 5 Deep front squat

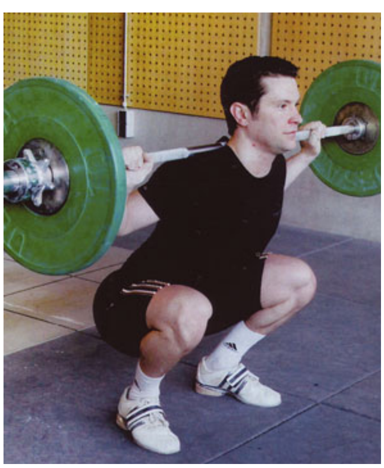

Fig. 6 Deep back squat (high bar)

mechanical loading tendon', 'in vitro mechanical loading ligament', 'in vitro patellar tendon properties', 'in vitro cruciate ligament properties', 'hip endoprosthesis pressure', 'gender differences cartilage volume knee'.

Exclusion criterion for the research was the bodyweight quarter squat, because no estimations of spinal-joint and knee-joint forces and tensile forces on tendons and ligaments of the knee with maximal loads are possible.

Inclusion criterion for the research was data about calculated knee-joint forces and tensile forces on tendons and ligaments of the knee in the half and deep squat with additional load, which were extrapolated to lower knee extension angles. Data from deep and half bodyweight squats were included to compare with data from loaded squats in equivalent knee-joint angles.

3 Mechanical Loading of the Knee (and its Biological Structure) with the Patellofemoral and the Tibiofemoral Joints During the Different Squats

There are three main studies [15–17] that are relevant for the recommendations of different squatting depths. Nisell and Ekholm [15] calculated the tibio- and patellofemoral compressive forces of a powerlifter (110 kg) that occurred in the low-bar deep back squat (ascent phase) with a load of 250 kg [2.27 × bodyweight (bw)]. At the turning point of approximately 50°, these values were between 6,750 and 7,000 Newton (N) (6.3–6.5 × bw), at 90°, they were between 6,000 and 6,250 N (5.6–5.8 × bw) and at 120°, the values decreased to 5,000–5,500 N (4.6–5.1 × bw). Because tendofemoral compressive forces between the quadriceps tendon and the femoral groove peaked to 6,000 N (5.6 × bw) at the turning point, Escamilla [8] recommends avoiding squatting depths below 90° (parallel and deep squats) to minimize the risk of injury for the tendofemoral complex.

Escamilla et al. [16] calculated the patellofemoral compressive forces in a half back squat to 80° with a load of 70–75 % 1-RM (146.0 ± 30.3 kg, 1.57 × bw). Subjects ($n = 10$) were experienced in strength training. In the descent phase of the lift, the highest compressive forces were evaluated at 95° (4,548 ± 1,395 N, 4.99 × bw). In the ascent phase, the highest compressive forces were evaluated at 85° (4,042 ± 955 N, 4.43 × bw). Escamilla et al. [17] were able to confirm these results. In both studies, the lowest rise of the calculated compressive forces was seen in the descent phase to 130°. For that reason, Escamilla [8] recommends the execution of the quarter back squat at 130° if degenerative changes of the patella-tendon complex are present. However, these calculated knee-joint forces, which are based on the half and deep back squat, cannot be extrapolated readily to the quarter back squat, because they do not take into account (a) the influence of the reversal of motion with lower retropatellar contact zones [18–20] and lower or missing tendofemoral support surface [18, 20–23] and (b) the different loads of the particular squat variant that can be tolerated, as detailed below:

(a) Based on biomechanical calculations [22, 23] and measurements of cadaver knee joints [18, 20, 21], the highest retropatellar compressive forces (in Newton, N) [20, 22, 23] and greatest compressive stresses (N/mm^2; in Megapascal, MPa) [18, 20] are observed at 90°. With increasing flexion, the additional contact between the quadriceps tendon and the intercondylar notch as the tendofemoral support surface (‘wrapping effect’, Bandi [22], p. 24) contributes to an enhanced load distribution and enhanced force transfer (pressure measurement to 60° [18, 20], force calculation to 60° [23], to 50° [22], force calculation on the basis of the pressure measurement to 60° [20], to 50° [21]). Therefore, a lower risk of injury may be expected. Because lower weights are used in deep squats and regular strength training practice leads to functional adaptations of the passive tissue, concerns about degenerative changes of the tendofemoral complex are unfounded and unproven. Additionally, with further flexion of the knee joint, a cranial displacement of facet contact areas with continuous enlargement of the retropatellar articulating surface occurs (measurement to 60° [19, 20, 24], to 40° [18]). In connection with the wrapping effect, declines of the retropatellar compressive forces [20–23] and reductions of the retropatellar compressive stresses are the result [18, 20]. Findings of magnetic resonance imaging (MRI) verify these retropatellar increments of deformation zones ($n = 28$) in vivo for the execution of a bodyweight parallel squat (to 60°) [24]. In contrast, at flexion angles beyond 120° [25, 26] and 90° [21], reduced contact areas of the articulating facets have been observed. These findings are likely due to different measuring methods and different magnitudes of force application and force duration [27]. However, after 80°–90°, declines of the retropatellar compressive forces can be seen due to the wrapping effect [21]. The magnitude of the articulating contact area is highly dependent on the compressive force [18], thus, force inclinations lead to a more wide-area deformation (determination to 90° [25, 26], to 50° [21], to 40° [18]). Female cadaver knees have been shown to possess 33 % lower contact areas of the patellofemoral joint than male samples at 120° and 90° knee flexion, although the femoral epicondylar widths were only 10 % smaller for the female than for the male samples [25]. The authors therefore assume gender differences in the bony geometry of the patellofemoral joint [25] that cannot be verified by the literature. MRI of knee joints (in vivo) in the static squat position (180°–120° knee angle) confirm no significant gender differences in retropatellar contact areas when these were normalized to patella dimensions (width × height) [28]. Patella width and patella height show only moderate and low correlations to body height ($r = 0.50$, $p \leq 0.05$ and $r = 0.64$, $p \leq 0.01$) and weight ($r = 0.22$ and $r = 0.19$). But tibial head diameter has been shown to significantly account for patella width ($r = 0.74$, $p \leq 0.001$) and patella height ($r = 0.82$, $p \leq 0.001$) of both genders [29] and to the surface areas of the whole knee for males ($r = 0.71$, $p \leq 0.001$) and females ($r = 0.65$, $p \leq 0.001$) [30]. Apparently for the patella, femur and medial and lateral tibia of males, the significantly higher articular surface areas are based in part on their greater tibial diameter (joint size) compared with females, even when matched for bodyweight or height [30]. But when normalized to the product of bodyweight and height, gender differences in cartilage surface areas of the same joint plates are no longer existent [31]. On the tibia plateau, the deformation zone (including intact menisci) depends on the joint angle and the compression load [32–37]. Measurements of articulating contact areas of the menisci confirm declines with increasing flexion. However, increasing the compression loads produced enlargement of the articulating contact zone [33]. These ex vivo measurements occurred to 90° flexion only [33]. In addition, cartilage deformation zones of the tibia plateau, determined with MRI to 60°, show continuous enlargements with increasing knee flexion in vivo (bodyweight lunge) [38]. The highest dynamic peak forces occur after the change into concentric muscle action [39, 40]. The execution of the half squat (80°–100°) cannot be recommended because the turning point is initiated in a knee-joint angle amplitude, where the highest patellofemoral compressive forces [20, 22, 23] and greatest compressive stresses [18, 20] occur with only a minor tendofemoral support surface [18, 20–23]. Pernitsch and Brunner [40] evaluated the patellofemoral (F_{cp}) and tibiofemoral compressive forces (F_{ct}) based on the same calculations as Nisell and Ekholm [15] but for the front squat. The subject (69 kg) lifted 80 kg (1.16 × bw). At the turning point of the half squat of 85° higher compressive forces were computed ($F_{cp} = 4{,}781$ N, $F_{ct} = 4{,}893$ N; 7.1 and 7.2 × bw) than at the turning point of a parallel squat of 58° ($F_{cp} = 3{,}995$ N, $F_{ct} = 4{,}437$ N; 5.9 and 6.6 × bw). Knee-joint loads remained almost constant during flexion and extension from 88° to 58° and again to 90°. For comparison: the bodyweight squat jump, started at a 108° knee angle, resulted in similar knee-joint forces to the half squat ($F_{cp} = 4{,}352$ N, $F_{ct} = 4{,}708$ N; 6.4 and 7.0 × bw). The subject initiated the ascent phase of all exercises with maximal explosive effort to reach maximal acceleration [40].

(b) With decreased knee flexion in the squat, the weight loads that can be lifted increase. The influence of increasing weight loads have not been considered in joint-force estimations of different squatting depths. For the quarter back squat (120° knee angle) physical education students were shown to be capable of lifting 3.89-fold (±0.33) their bodyweight [41]. For professional soccer players, Hoff and Helgerud [42] recommend values of 2.75-fold of bodyweight in the half back squat. Higher weights result in increased tibio- [12, 13] and patellofemoral compressive forces [14]. These data indicate that the force values for half and deep back squats calculated by Escamilla et al. [16] and Nisell and Ekholm [15], which were intended to be extrapolated to lower extension angles up to 120°–130°, are too low. For the quarter back squat, physical education students were shown to be capable of moving weights of 4.02-fold (±1.59) the weight of a deep back squat and 4.38-fold (±1.02) the weight of a deep front squat, although the majority of the subjects had minimal strength-training experience [41]. Well trained athletes are capable of lifting much greater weights. Based on the weight of the deep back squat (250 kg, 2.27 × bw) that was moved by the powerlifter described in Nisell and Ekholm [15], this weight would correspond to an astronomical 1005 kg. This weight is out of question for training practice, because such a high barbell load is unlikely to be stabilized by the back musculature. The back, not the legs, would be performance-limiting, thus providing no training effect for the legs.

McKean et al. [43] determined the movement pattern with regards to the timing when maximum angles of the hip and knee occurred while performing the parallel back squat. Sixteen men and twelve women executed the squat with no additional weight or with 0.5-fold bodyweight. Independent of gender, phase (ascent or descent) and load, the subjects demonstrated their maximum hip and knee angles within 2 % of the deepest position whereby subjects had to move their knees forward of the toes by 6.4–6.5 cm in men and 9.3–9.7 cm in women. The difference in the forward movement of the knees between genders could be explained by the greater body height of the men (167 vs. 179 cm). The subject's height and tibial length has been shown to account for 69 % of the explained variances in male subjects being capable of keeping their heels on the ground while performing the parallel back squat [44]. These anthropometric factors have an influence on the torso inclination and hence the forward movement of the knees [43]. In training and practice of strength training it is sometimes suggested that the tibia should move anteriorly only to the point where the knee joints and the toes form a vertical line to minimize the tibiofemoral shear forces. This recommendation is based on video analyses of parallel back squats (mean weight 201.85 kg, 2.23 × bw) of 12 male weightlifters. Data of only three subjects are presented. The subject with the greatest forward movement of the knees while performing the squat had the greatest tibiofemoral shear forces [45]. The restriction of the forward knee displacement will result in changes to the knee-hip coordination [43] with greater forward leaning [46] and ventral flexion of the thoracic and lumbar spine [47]. This evasive movement elicits greater anterior shear forces on intervertebral discs [48] and causes tensile forces on intervertebral ligaments [48, 49]. For that reason, instructions about a restriction of the forward knee displacement have to be strictly avoided. This recommendation is based on a misinterpretation of existing data and should be removed in future practical literature. It is not clear from the data of Ariel [45] whether the calculated shear forces acted in an anterior or posterior direction. The subject with the greatest anterior knee movement, who showed the highest shear forces, squatted to a knee angle of 90° only. The remaining subjects squatted to 61° and 69° knee angles and possessed lower shear forces. The literature provides evidence for these calculations: When initiating the turning point at 80°–90°, higher (relative) posterior shear forces are the result [16, 17, 50] compared with the deep squat at 50° [15] (chapter 5). Additionally, the calculations described by Ariel [45] do not provide any information about the squatting technique of the subjects. The high shear forces could result from a very upright posture, which can be demonstrated by the following calculations of the compressive forces in the patellofemoral joint. For the parallel back squat, Wretenberg et al. [3] evaluated higher mean patellofemoral compressive forces for eight weightlifters than for six powerlifters (4,700 ± 290 vs. 3,300 ± 1,700 N, 5.84 × bw vs. 3.87 × bw), despite the weightlifters lifting lower weights than the powerlifters (101.9 ± 27.7 vs. 154.2 ± 21.1 kg, 1.24 × bw vs. 1.85 × bw). Considering the biomechanical calculations of Nisell and Ekholm [23], this result was due to the higher torque values in the knee joint, which were developed via a more upright posture (high- vs. low-bar position of the barbell) [3], see Fig. 1.

There are no calculations for knee flexion angles beyond 50° in the deep squat that permit an accurate estimation of knee-joint forces because these studies did not take the influence of the wrapping effect [51] or the influence of soft tissue contact between the back of the thigh and the calf into account [51–53]. Evaluations of Reilly and Martens [51] and Dahlkvist et al. [52] showed patellofemoral compressive forces (7.6 × bw and 7.62 × bw) in the turning point of the bodyweight deep squat (40°) that exceeded those of Nisell and Ekholm [15] (6.3 × bw), Pernitsch and Brunner [40] (5.9 × bw) and Wretenberg et al. [3] (weightlifter 5.84 × bw, powerlifter 3.87 × bw). It is surprising that despite only using bodyweight loading, Reilly and Martens [51] and Dahlkvist et al. [52] measured higher patellofemoral compressive forces than in the studies of Nisell and Ekholm [15] and Wretenberg et al. [3] using up to 70 % 1-RM loads in the deep (50°) and parallel squat (64°–69°). The subjects described by Dahlkvist et al. [52] experienced difficulties in maintaining their balance while performing the squat. However, an uncontrolled execution leads to distinctly higher knee-joint forces [45]. The same contradictions between the studies of Dahlkvist et al. [52] and Nisell and Ekholm [15] exist for the calculated posterior shear forces.

The influence of soft tissue contact between the back of the thigh and the calf plays a prominent role in reducing the knee-joint forces beyond 40° of knee extension [54–56]. The calculation of knee-joint forces, which did not take into account the soft tissue contact in deep flexion angles (25°) [53], must be examined critically. This soft tissue contact depends upon the cross-sectional area of the hamstring and the calf muscles and can begin at higher degrees of extension from approximately 60° [57], thereby reducing tibiofemoral [54–56] and patellofemoral joint forces [54, 55]. The structures of the knee (such as meniscal and cartilage tissue and ligaments) that benefit from soft-tissue contact remain unclear. The apparent higher risk for degenerative changes, such as osteochondrosis dissecans of the odd facet in deep squats [8], seems unfounded and is unproven. Recommendations for half or quarter squats to avoid degenerative changes in the knee joint may be counterproductive. If the cartilage tissue of the odd facet is inadequately stressed then it receives insufficient nourishment that leads to consequent degeneration and atrophy [58]. This result is confirmed by animal studies in which cartilage tissue was exposed to hypopression by unweighting the extremities [59, 60].

3.1 Adaptation Effects and Damage of Meniscal and Cartilage Tissue of the Knee Joint

The training of weightlifters confronts the athlete with deep front/back squats and exercises in deep knee bends with acceleration and deceleration of high barbell loads. With ten training sessions per week at the international level [61], it could be suggested that these athletes are predisposed to a high prevalence of acute and chronic knee injuries with long training interruptions. However, from 27 weightlifters from different US Olympic training centres, Calhoon et al. [62] determined 3.3 injuries per 1000 training hours over a 6-year period. Of missed training time due to knee problems, 95.3 % lasted 1 day or less, and in the remaining cases, lasted less than 1 week. These training interruptions were primarily related to tendinitis due to chronic overuse and in very few cases to muscle strains due to acute injuries [62]. A 4-year retrospective study of 1,109 weightlifters (age 12–20 years), who participated in national or international competitions, did not show any injuries (i.e. at the epiphyseal joints) that required surgery or hospitalization [63]. An examination of questionnaires completed by weightlifters aged 13–16 years ($n = 1{,}634$) and based on 168,551 hours of training, indicated that the injury rate of American weightlifters is 0.0017 per 100 hours of training. This prevalence is much lower than the injury rate in US basketball players (0.03), US track-and-field athletes (0.57), American football players (0.10) and US gymnasts (0.044) [64]. According to a questionnaire conducted by 80 weightlifters, knee problems were not common in the deep squat. There have been no meniscectomies among the lifters. No knee clicks or pops were reported. Acute knee injuries like sprains usually occurred in the catch phase in the deep-squat position of the clean and jerk with high loads and not in the deep-squat exercise [65].

According to the findings of Pernitsch and Brunner [40], for knee joint forces in the range of 60°–110° the acceleration achieved determines the compressive forces more than the weight load. The higher the lowering speed in the descent phase, the higher the developing deceleration phase to avoid a 'dipping movement' and hence a rising increase of tibiofemoral shear and compressive forces in the turning point of the squat [45]. For that reason, care should be taken to complete a slow and controlled execution [45] comprising a descent phase of 3 and 4 seconds in the deep squat corresponding to an average angular velocity in the knee joint of 46.66 and 35°/sec, respectively. For comparison, international-level weightlifters produced a 10-fold and 13-fold higher average angular velocity in the knee joint of 465.67°/sec in the snatch (1-RM) in the descent movement under the barbell (descent phase 238 msec) [66]. At the turning point (knee angle 17.17°) [66], this will cause increases of the knee-joint forces accordingly. The following calculations of the clean and jerk with a weight of 120 kg are indicative of this point. For an elite weightlifter (71 kg) Collins [57] calculated the tibiofemoral compressive forces at the turning point (knee angle 35°–40°), which accounted for maximal values of 24-fold the bodyweight of the athlete. However, cross-sectional findings, determined using MRI, verify that knee joints of professional weightlifters ($n = 7$) possess significantly ($p < 0.01$) higher cartilage thickness for the same patellar contact area of 14 %, in average, compared with controls ($n = 14$) who are not strength-trained [67]. When loaded, an increased retropatellar cartilage thickness shows an increased stiffness that corresponds to an increased mechanical stress tolerance [68]. Although neither contact area nor cartilage thickness of the tibia plateau and the femur condyles demonstrated significant differences [67], in vivo evidence from cross-sectional [69] and longitudinal studies [70, 71] with humans show that increased activities can lead to anabolic biochemical and structural adaptations of the cartilage tissue, causing increased mechanical stress tolerance and hence protective effects against degenerative changes [70, 72, 73]. There were no tibiofemoral or patellofemoral cartilage defects in the weightlifters [67]. These athletes had begun their weightlifting training at between 7 and 13 years and continued until the point of data evaluation [67]. For active weightlifters at a national and international level and a mean training experience of 17 years ($n = 13$, mean age 35.3 years), the prevalence of degenerative cartilage changes of the patellofemoral and tibiofemoral joints (grade 2–4) is not higher than in men of the general population and of similar age ($n = 162$) [74].

Laboratory studies with animal [75, 76] and human preparations [77], and animal experiments (in vivo) [78], confirm that the menisci are responsive to anabolic metabolic processes and functional structural adaptations that are induced by dynamic loading [75–77] and increased activity [78]. Evaluations of former professional weightlifters ($n = 29$, mean age 59.3 years) provide evidence for this adaptation potential in humans: when compared with former professional soccer players ($n = 31$, mean age 56.5 years), patellofemoral arthrosis dominated for the weightlifters. For the soccer players, tibiofemoral arthrosis was dominant. The prevalence of gonarthrosis in the weightlifters (31 %) did not differ from the soccer players (29 %), but both cohorts demonstrated a higher percentage than former professional long-distance runners (14 %) ($n = 28$, mean age 59.7 years) [79]. However, the misdiagnosis of functional and degenerative cartilage changes must also be noted [80].

Based on the facts presented, it can be assumed that many years of strength training in the full range of motion and using the correct movement pattern results in functional adaptations of the articulating cartilage and meniscal tissue.

4 In Vitro Pressure Thresholds of Cartilage and Meniscal Tissue and Their Application to In Vivo Conditions

It has been suggested that the deep squat with high loads may exceed the compression threshold of articular cartilage of the knee joint [81]. Pernitsch and Brunner [40] calculated at a knee-joint angle of 66° in the bodyweight squat jump (starting angle 60°) tibiofemoral compressive forces of 5.1-fold bodyweight (3,449 N). In cadaver knees, the same relative force value at 60° knee angle resulted in a measured tibiofemoral compression value of 26.6 MPa [81]. The methodological procedure has to be critically examined. Ex vivo conditions cannot guarantee the natural joint movement, which is subject to muscular support in vivo. This can lead to unnatural local peak forces. Indenter [82] and impactor compression [83] elicited chondrocyte death in the cell matrix at 15–20 MPa of bovine cartilage explants [82] and at 20–30 MPa of human bone cartilage explants [83]. Repetitive indenter compression of bovine bone cartilage explants already caused initial chondrocyte death at over 6 MPa due to summed microscopic trauma [84]. Joint loading of hip endoprosthesis of living persons has been shown to result in 8.9 MPa when rising from a chair [85]. This means that (repetitive) rising from a chair or jumping from the parallel squat position would already exceed the compression threshold of hip and tibiofemoral articular cartilage. The influence of the acceleration on the patellofemoral stress can be seen when comparing the calculations of the compressive stress in the bodyweight squat jump (94°) and in the loaded front squat with 80 kg (86°), eliciting comparable compressive stress (Table 1). However, the mechanical properties and composition vary between cartilage tissue of articular joint areas (i.e. femoral-trochlear vs. patellar) [86]. Furthermore, the composition, structure, relative thickness [87] and mechanical properties [88, 89] of cartilage tissues show species-specific differences [87–89]. Whether the experimentally based threshold values of in vitro studies, which were determined with indenter [82, 84] or impactor devices [83], can be extrapolated to the in vivo (human) condition must therefore be critically examined. In articular cartilage of intact joints, the load is transmitted by hydrostatic compression and not by elastic deformation [90]. When two opposing articulating cartilage explants of bovine knee joints were compressed in a compression cylinder, there was no cellular trauma even at a hydrostatic pressure of 50 MPa compared with the pure impactor compression that already caused cell death at 10 MPa [91]. These factors clarify the issue of making general statements on compression threshold values of articular cartilage.

To the authors' knowledge, there are only two studies that provide information on compression test-induced cellular traumata of meniscal explants. Findings of animal experiments point to comparable mechanical tolerance thresholds of bovine medial [92, 93] and lateral [93] meniscal explants compared with cartilage tissue (in vitro). In addition to cartilage tissue, the mechanical properties of meniscal tissue show species-specific differences [94, 95]. Joshi et al. [94] and Sweigart et al. [95] did not confirm any statistically significant differences in the compression behaviour of bovine and human medial meniscus explants. However, Sweigart et al. [95] revealed significant differences in the permeability between both species. The extent to which the findings of Nishimuta and Levenston [93] and Kisiday et al. [92] can be extrapolated to humans must be investigated.

Table 1 Calculations of the peak patellofemoral (*PF*) compressive stress in the parallel and half front squat, half back squat, and in the squat jump at different starting angles

Reference	N	Activity	Turning point/ starting angle	Barbell load (relative strength)	PF stress
Pernitsch and Brunner [40]	1	Squat jump	–/60°	0	8 MPa at 66° (ascent)
		Squat jump	–/94°	0	11 MPa at 94° (ascent)
		Squat jump	–/108°	0	11 MPa at 108° (ascent)
		Parallel front squat	66°/–	30 kg (0.43)	10 MPa at 68° (turning point)
		Parallel front squat	58°/–	80 kg (1.16)	10 MPa at 58° (turning point)
		Half front squat	86°/–	80 kg (1.16)	12 MPa at 85° (turning point)
Escamilla et al. [16]	10	Half back squat	80°–90°/–	146 ± 30 kg (1.57)	11.6 MPa at 90° (turning point)

These exercises of Pernitsch and Brunner [40] were performed with maximal explosive effort. Calculations of the patellofemoral compression stress from Escamilla [8] (p. 135) are based on the averaged knee-joint forces with a controlled execution of the half back squat [16]. Calculations of both studies are based on the same joint contact areas from Huberti and Hayes [20]

MPa Megapascal, *relative strength* kilogram barbell load per kilogram body mass

5 Calculated Tibiofemoral Shear Forces in Different Squatting Variants Considering Mechanical and Morphological Properties of Ligaments (In Vitro) and Their Adaptations to Physical Activity

Tibiofemoral compressive forces have been shown to be an important component of knee stabilization [96–99]. In cadaver knees, antagonistic hamstring load in conjunction with quadriceps load minimized shear forces by limiting tibiofemoral translation and hence tensile forces on the anterior cruciate ligament (ACL) [100–102]. Furthermore, the magnitude of shear forces in the tibiofemoral joint is highly dependent on correct execution and hence on the level of movement pattern, which was determined with weightlifters in the parallel back squat [45].

In the context of this manuscript, the anterior shear forces are defined as those forces that act on the ACL, and the posterior shear forces as those forces that act on the posterior cruciate ligament (PCL) [103]. In the present manuscript, tensile strength (N) is defined as the ultimate load that ligaments and tendons can maintain tension before failure. The tensile strength of the PCL was based on measurements of 53- to 98-year-old cadaver samples that led to average values of 1,620 ± 500 N. Information about bodyweight is absent [104]. For individuals under 26 years, the PCL strength of 4,000 N is extrapolated from measurements of the ACL of cadaver samples of varying ages [105]. Tensile tests for the ACL of 16- to 35-year-olds (ex vivo) showed a strength of 1,730 ± 660 N (3.35 × bw [105]) and 2,160 ± 157 N (bw not available) [106].

Increases of the barbell load (0.5–1.2 × bw) led to increases of the calculated posterior shear forces in the half back squat (90°) [13]. The subjects ($n = 20$) described by Sahli et al. [13] achieved, in the lowest position of the descent phase with the highest load tested (95 kg, 1.2 × bw), calculated posterior shear forces of 535 N (0.7 × bw), on average. Extraordinarily high posterior shear forces in the squat exercise were reported in the study of Toutoungi et al. [107]. At the turning point of 80°, these authors calculated PCL forces of 2,704 N (3.5 × bw), on average. These data confirm the calculated posterior shear forces of 2,913 N (3.88 × bw), on average, of Dahlkvist et al. [52], although in both studies, subjects were not confronted with any additional weight. In the bodyweight squat, Sahli et al. [13] calculated at the turning point of 90° mean posterior shear forces of 225 N (0.29 × bw) only. Pernitsch and Brunner [40] calculated, at the turning point of 68° (30 kg burden) and of 58° (80 kg burden) of the parallel front squat, maximum posterior shear forces of 792 and 787 N (1.16 × bw and 1.17 × bw). Nisell and Ekholm [15] calculated the highest posterior shear forces, while lifting a barbell load of 250 kg (2.27 × bw), to be 1,800 N (1.67 × bw) at the turning point of the squat (50°). Escamilla et al. [16, 17] and Wilk et al. [50] calculated, for the half back squat (turning point 80°–90°) with lower weights (133–147 kg, 1.43–1.58 × bw), greater (relative) posterior shear forces than Nisell and Ekholm [15] in the deep back squat. Shear and PCL forces were between 1,783 and 2,066 N (1.96–2.25 × bw) in the ascent phase between 90° and 117° knee angle [16, 17, 50]. Pernitsch and Brunner [40] calculated, for the half-front squat with maximal-explosive effort (turning point 85°) with 80 kg (1.16 × bw), posterior shear forces of 621 N (0.92 × bw).

These calculated posterior shear forces and PCL forces with loads of 0.43- to 2.27-fold bodyweight in the parallel and deep squat account for between 19.68 and 45 % of the extrapolated tensile strength of a PCL for under 26-year-olds [104]. The calculated posterior shear forces in the half squat with loads of 1.16- to 2.25-fold bodyweight account for between 13.39 and 51.65 %. The findings of Toutoungi et al. [107] and Dahlkvist et al. [52] reach estimated values of 67.6 and 72.83 %. These values have to be critically examined. The calculated posterior shear forces of the bodyweight squat (90°) of Sahli et al. [13] account for 5.63 % of the extrapolated tensile strength of the PCL only.

While Escamilla et al. [16, 17] and Wilk et al. [50] did not calculate any anterior shear forces for the half back squat, Toutoungi et al. [107] calculated a very low ACL force of 28 N (0.03 × bw), on average, in the ascent phase of the bodyweight squat. In the deep squat (250 kg burden), Nisell and Ekholm [15] calculated the highest anterior shear forces in the final 30° of knee extension of approximately 500 N (0.46 × bw). Pernitsch and Brunner [40] calculated in the parallel front squat (80 kg), in the ascent phase of the lift, the highest anterior shear forces of 251 N (0.37 × bw) at 158° (maximal explosive effort). In the half front squat, in which a higher acceleration occurred, the same load resulted in 2.86-times higher anterior shear forces of 719 N (1.06 × bw) at 138° knee angle.

The calculated anterior shear forces and ACL forces with loads of 1.16- to 2.27-fold bodyweight in the parallel and deep squat account for between 11.62 and 28.9 % of the tensile strength of an ACL of 16- to 35-year-olds [105, 106]. The calculated anterior shear forces in the half squat with a load of 1.16-fold bodyweight accounts for between 33.29 and 41.56 %.

Based on these calculations, in deep squats, neither posterior nor anterior shear forces may reach magnitudes that could harm an intact PCL and ACL.

Training interventions of 8- to 21-weeks duration confirm that parallel [108] and deep back squats [109, 110] do not cause any negative effects on knee ligament stability. Measurements of knee stability followed immediately by the execution of parallel back squats with 1.6-fold bodyweight show no significant changes (powerlifters, $n = 24$) when compared with 19 % ($p \leq 0.01$) after a 6.2-mile road race (distance runners, $n = 12$) or 19 % ($p \leq 0.01$) after basketball training (basketball players, $n = 10$) [111]. In contrast, cross-sectional studies with 28 weightlifters and 27 powerlifters confirm significantly ($p \leq 0.005$) higher knee stability compared with 30 controls with little or no strength training experience [109].

The adaptation potential of ligaments is known from training studies with animals [112], i.e. the increased tensile strength and enhanced stiffness and enlarged elasticity modulus of the ACL [113]. MRI of knee joints of active professional weightlifters ($n = 9$, mean age 26.1 years) show an impressive significantly larger cross-sectional area (CSA) ($p \leq 0.05$) of the ACL (71.7 vs. 40.56 mm^2) and PCL (64.48 vs. 44.98 mm^2) compared with age-matched untrained controls ($n = 19$), with no significant group differences in body height and weight. These athletes had already begun weightlifting at an age of between 9 and 12 years [114]. The CSA of the ACL and PCL of the weightlifters were 61.49 and 50 % larger than those of the cadaver studies [104, 105]. Higher tensile strength, and hence lower risk of injury of these structures, may thus be expected. Possible improvements of the material properties that are unrelated to CSA, as is known for the elasticity modulus from animal training studies [113], remain yet unconsidered.

6 Mechanical and Morphological Properties of Tendons (In Vitro and In Vivo) and Their Adaptations to Physical Activity

Performing deep squats has been suggested to increase the risk of suffering a distal tendinopathy of the patella tendon. Based on MRI of 24 knee joints from individuals with distal tendinopathy, Johnson et al. [115] suggest that an impingement of the patella tendon at the apex inferior could be responsible for the injury localization. The authors attribute the pathogenesis to a general impingement in deep flexion and not to chronic overuse because there were no significant anatomical differences when compared with asymptomatic knee joints of a control group. The measurements were performed to a knee angle of 120° only. Schmid et al. [116] were unable to verify this assumption with MRI of 19 symptomatic and 32 asymptomatic knee joints in deeper joint positions to 80°. As with the proximal tendinopathy [117], the pathogenesis of the distal tendinopathy was ascribed to chronic overuse [116].

Strength training [118, 119] and cross-sectional studies of badminton, fencing, volleyball, long-distance running and weightlifting athletes [120–122], show that the Achilles and patella tendons are responsive to increases in CSA which is consistent with an enhanced stiffness [118–120]. In ex vivo comparative tests with 10-mm-wide preparations (central third) ($n = 16$, mean age 24.9 years), Stäubli et al. [123] determined a tensile strength of 2,173 ± 618 N for the quadriceps tendon and 1,953 ± 325 N for the patella tendon. For the tensile strength of the patella tendon, Cooper et al. [124] revealed both a linear relationship to the width and to the CSA of the preparations. For the central third of the patella tendon (width 15 mm) of young persons (mean age 28 years), Cooper et al. [124] measured a tensile strength of 4,389 N, on average. For male cadavers, the width of the quadriceps tendon and patella tendon is 50 and 35.8 mm, on average [23], thus the tensile strength of these tendons is much greater. Adams et al. [125] determined that intact preparations from 52-year-old cadavers showed a 90 % higher tensile strength of the quadriceps tendon (3,660 ± 830 N, $p \leq 0.05$) compared with the patella tendon (1,920 ± 330 N), which is based on the fact that the quadriceps tendon is 25–30 % wider and thicker than the patella tendon [23]. These clearly different values of these age groups [123–125] may be explained by differences in testing methods, bodyweight and activities during the lifetime of the cadavers [123].

A bony detachment of the patella tendon was calculated at a tensile force of 14,500 N (18 × bw of an 82.5 kg heavy male athlete) during weightlifting in the jerk phase [126]. A bony avulsion of the quadriceps tendon was observed for a powerlifter at the turning point of the parallel back squat (382.5 kg burden) [15]. Nisell and Ekholm [15] extrapolated a tensile strength of this tendon from 10,900 to 18,300 N (11–19 × bw). The extrapolated tensile strength of the patella tendon was 8,000–13,100 N (8–16 × bw). These high extrapolated values of Nisell and Ekholm [15] and Zernicke et al. [126] support the conclusion that increases in tensile strength is caused by regular, long-term strength training, which is based on increases in the CSA, but not only in its insertion sites [118–120]. In high-performance weightlifters ($n = 9$, mean age 26.1 years), Grzelak et al. [122] verified 37.1 % ($p < 0.05$) larger CSA of the patella tendon in its midsubstance compared with age-matched untrained controls ($n = 19$), with no significant group differences in body height and weight. Because an increased tendon CSA leads to enhanced tensile strength [124, 127, 128], it enhances the stiffness [118–120, 129], which reduces the risk of injury of the tendon in the long term. However, increases in tendon stiffness are also possible via training-induced enhancements of the elasticity modulus [118, 119, 130], which is unrelated to the CSA. On the basis of animal studies, this adaptation phenomenon is explained as increments of collagen content and of collagen cross-links [129]. These structural changes were also noted in MRI of individuals with proximal tendinopathy after strength training and were attributed to a remodelling process [131].

Studies confirm that in addition to tensile stimuli [132, 133], compression stimuli [134–136] stimulate or influence the synthesis of extracellular proteins in tendons and tendon cells. Human cadaver preparations demonstrate that tendons and ligaments, which wrap around bony structures, develop compression areas with cartilaginous composition [137]. Therefore, it is obvious that relative increases in the thickness of the quadriceps tendon occur when the increasing wrapping effect in deep squats causes higher compression of this tendon.

7 Adaptation Effects and Damage of Passive Tissues in the Spinal Joints

To the best of the authors' knowledge, there are no calculations about the compression loads of the lumbar spine between different squatting depths with maximal loads. These estimations are therefore extrapolated from existing literature. Cappozzo et al. [10] calculated the compression loads on the L3–L4 segment on four subjects, who performed half and quarter back squats. Weights of between 0.8- to 1.6-fold bodyweight resulted in compression loads of 6- to 10-fold bodyweight at the turning point of the squat (3,100–7,324 N). With increasing loads, these authors calculated elevations in compression loads acting on the spine [10]. Physical education students lifted 1.26- (±0.23) and 1.41- (±0.3) fold their bodyweight in deep back and deep front squats [41]. Based on the calculations of Cappozzo et al. [10], the compression loads of the lumbar spine were within their previously calculated scores. Relative strength values in the quarter back squat were, on average, 3.89- (±0.33) fold bodyweight [41]. Based on the calculations of Cappozzo et al. [10], compression loads acting on the L3–L4 segment may have exceeded 20 times the bodyweight. While in the load combination of high-axial compressive and shear forces in ventral flexion intervertebral discs prolapse [138], in axial compression, the vertebral body is the weakest link and is the initial structure to show compression failure in the fracture of the endplate [139, 140]. However, high axial compression loads that act on the spine during long-term weightlifting training result in functional adaptations of the vertebral bodies including enhanced compression tolerance through increased bone mineral density (BMD) [141]. In the present manuscript, the compressive strength (N) is defined as the ultimate load a vertebral body or segment can tolerate under axial compression before failure. In ex vivo measurements, a positive and linear correlation of $r = 0.82$ ($p < 0.00001$) was found between BMD and compressive strength of the vertebral bodies (L3, $n = 101$) [142]. Hansson et al. [143] determined a compressive strength of a vertebral body of 11,000 N.

The compressive strength (ex vivo) of an L4/L5 vertebral segment for a 22-year-old man was 8,800 N, that of an L3/L4 vertebral segment of a 22-year-old woman was 6,200 N and of a 39-year-old man was 8,200 N [140]. Based on the calculations of Cappozzo et al. [10], the compression loads in the quarter back squat would have exceeded the compression tolerance limit [41]. Therefore, these extrapolations must be examined critically. However, these comparisons lead to the serious question of how performance-oriented strength training with comparatively supra-maximal loads in the quarter squat may increase the risk of injury of the spinal column, in particular for female athletes. Females possess significant lower compressive strength of vertebral bodies (L3) [144] because of their significantly lower end-plate CSA than their male counterparts [145–147]. This means the female lumbar vertebral body is exposed to higher axial compressive stress than a male spine when subjected to an equivalent load [145].

Twenty-five elite weightlifters (mean age 17.4 years) with an average training experience of 2.5 years possessed 133 % ($p \leq 0.05$) higher BMD of the L2–L4 vertebrae than age-matched controls with no significant group differences in body height or weight. In addition, these values significantly exceeded the reference values of 400 adult men of between 20 and 39 years by 113 % ($p \leq 0.05$) [141]. One year later, an additional measurement demonstrated a further increase in the BMD of these weightlifters [1]. Furthermore, Neumann et al. [148] determined a positive and linear correlation of the BMD of lumbar vertebrae (in vitro) to both the tensile strength ($r = 0.84$, $p \leq 0.05$) and to the stiffness of the anterior longitudinal ligament ($r = 0.78$, $p \leq 0.05$). Increases in tensile strength and stiffness could cause a greater passive stability of the vertebral segments in vivo. Combined with an increased BMD and a well developed muscle corset, regularly practiced strength training can be attributed a protective effect.

For a man with a bodyweight of 70 kg, lifting a beer crate (20 kg) close to the body from a squat position resulted in measured intradiscal (L4/L5) compression values of 1.7 MPa [149]. At 2.5 MPa, human discus explants demonstrated increases in proteoglycan synthesis but showed a reduction at 7.5 MPa. Intermediate values were not evaluated [150]. In vivo experiments with rats underpin an adaptation potential of the caudal nucleus and annulus pulposus when subjected to 2 weeks of dynamic compression at 1 MPa, which amounts to 3-fold the bodyweight of the animal [151]. Because there are differences between humans and rats in the cell content of the intervertebral disc, extrapolations to humans must be made with a certain degree of reserve [152]. On an international level, weightlifting includes ten training sessions per week, involving 400 repetitions and much more in phases of high training volume with 70–90 tons lifted [61]. However, for weightlifters ($n = 25$, mean age 31.5 years) the extent of radiographically determined degenerative changes of the spine was not higher when compared with track-and-field athletes ($n = 25$, mean age 27.0 years) [153]. MRI of the spine (T6–T7, L5–S1) of elite athletes (mean age 26 years) showed that weightlifters ($n = 21$) and ice-hockey players possessed the highest prevalence of degenerative disc abnormities when compared with wrestlers ($n = 13$) and orienteers ($n = 18$). These comparisons led to no statistically significant group differences, which was not even significant to untrained control persons ($n = 21$, mean age 28 years). Fifteen years later, an additional measurement demonstrated further deteriorations of the previously diagnosed degenerative diseases in 88 % of the elite athletes, the most of which were seen in the ice-hockey players [154]. The investigators did not find any statistical correlation between back pain and number of affected discs or any type of abnormality on MRI [154]. Epidemiological studies over 4- to 6-year periods did not report any serious injuries of the spine in competitive weightlifters [62–64]. A questionnaire-based survey of former weightlifters ($n = 13$, age 40–61 years) emphasized a lower prevalence of low-back pain (23 vs. 31 %) compared with a general population ($n = 716$, age 40–47 years) [155]. MRI of lumbar vertebrae of former high-performance athletes confirm that weightlifters ($n = 29$, mean age 59.4 years) possess no statistically significant group differences in disc height (L1–L2, L5–S1) compared with long-distance runners ($n = 27$, mean age 59.6 years), shooters ($n = 28$, mean age 61.1 years) or soccer players ($n = 30$, mean age 56.6 years). In addition, there were no statistically significant group differences in the lumbar mobility [156]. Based on MRI, the prevalence of disc abnormities such as reduced disc height (T6–T7, L5–S1) is not higher in elite weightlifters ($n = 10$, mean age 42 years) compared with untrained control persons ($n = 10$, mean age 43 years) [154].

Therefore, it is obvious that human intervertebral discs are responsive to training-induced increases of their compression tolerance in the long term.

Walsh et al. [157] performed a three-dimensional motion analysis of the lumbar spine during the half back squat. Subjects ($n = 48$) had strength training experience and lifted three submaximal-intensity loads (40–80 % 1-RM). With increasing load, subjects hyperextended their lumbar spines significantly. These authors were concerned about the increased pressure in the posterior annulus that was analyzed by Adams et al. [158] in the combination of axial compression and extension. These findings are based on ex-vivo measurements of the L4–L5 segment. During axial compression, 2° of extension led to significantly increased compression stress within the posterior annulus compared with neutral loading [158]. The expressed fears of Walsh et al. [157] during the execution of the squat exercise are unfounded because the hip-joint angle changes. On the contrary, the deep squat involves the risk of dissolving the lordotic curvature in the turning point [159]. McKean et al. [159] determined a lower range of movement at the sacrum ($p \leq 0.01$) with a larger range of lumbar flexion ($p \leq 0.01$) in 18 males compared with 12 females while performing the descent phase in the parallel back squat with a narrow stance (pelvic width). In contrast, females demonstrated less range of lumbar flexion and more anterior tilt of the sacrum compared with males. Women have a lower stiffness and greater range of motion between motion segments of the lumbar spine than men [160]. McKean et al. [159] therefore assume that the females were capable of developing more muscular stabilization of the lumbar spine that may explain their greater sacrum movement due to higher flexibility of the lumbar sacral region. Another contributing factor could be the greater hamstring flexibility that females have been shown to possess compared with males [161]. Subjects squatted with a barbell load of 50 % bw only [159]. Larger weights may also predispose females to a higher risk of ventral flexion of the lumbar spine while squatting. Lander et al. [162] determined five male subjects with strength training experience in performing parallel back squats (75–80 % 1-RM) with an eight repetition maximum load. Completion near muscular failure caused greater forward leaning, induced by exhaustion. Quadriceps fatigue has been shown to affect lifting technique with regard to performing a back lift rather than a squat lift [163]. Forward leaning during the execution of the squat increases the risk of ventral flexion of the lumbar spine [47], in particular in fatiguing lifting condition. The load combination of high-axial compressive and shear forces in ventral flexion increases the risk of a spinal disc herniation [138]. To minimize the magnitude of this ventral flexion, it is necessary to induce a lumbar extension manoeuvre before motion reversal. This movement leads to increased activity of the erector spinae and joint closing of the apophyseal joints, causing reduced shear forces on the intervertebral discs [48]. Weightlifters are trained to maintain the lordotic posture during movement onset in the clean and jerk, also. Calhoon et al. [62] calculated the injury prevalence of 27 Olympic weightlifters to 3.3 events per 1000 hours of training over an observation period of 6 years. Missed training time caused by lower back injuries lasted for 1 day or less for 87.3 % of cases and less than 1 week in the remaining cases.

8 Conclusion

For elite athletes, the perennial training structure in deep front and back squats obtains target values between 1.5- to 2-fold bodyweight [164]. It is unclear why higher risk of injury of passive tissues in deep squats is hypothesized [5, 81], although considerably lower weights are accomplished in this variant. When compared with half and quarter squats, in the deep squat, lower knee joint and spinal joint stress can be expected. Provided that the technique is learned accurately under expert supervision and with progressive training loads, the deep squat presents an effective training exercise for protection against injuries and strengthening of the lower extremity.

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