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REVIEW ARTICLE

A Systematic Review of in Vivo Anterior Cruciate Ligament Loading During Static, Slow-Speed and Athletic Tasks

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Abstract

Purpose An anterior cruciate ligament (ACL) injury is one of the most common severe knee injuries in sports. The purpose of this review was to summarize the studies that directly quantified in vivo ACL loading as a function of knee motion in healthy individuals during static, slow-speed and athletic tasks.

Methods A systematic review of the literature in multiple databases was conducted using different combinations of the terms "anterior cruciate ligament" or "ACL" and "in vivo" combined with "tensile" "strain" "stress" "force" and "loading".

Results A total of 27 studies were identified. Eleven studies utilized strain sensors, while 16 studies applied imaging techniques. The numbers of studies for static or semi-static postures, slow-speed tasks, and athletic tasks were 12, 11 and 4, respectively.

Conclusion There were strong and negative correlations between ACL elongation and knee flexion angles across different tasks. Peak ACL elongation mostly occurred when the knee flexion angle was minimal. Increased tibial anterior shear forces and patellar tendon forces would increase ACL loading when the knee is kept at a constantly small angle. In addition, a high patellar tendon force could be generated by preparatory quadriceps activation to load the ACL even when the lower extremity was not in contact with the ground during athletic tasks. Furthermore, while exercise modalities might affect peak ACL loading, the relationship between exercise intensities and ACL loading was complex and should not be assumed to be linear.

Keywords ACL · ACL injury · Loading mechanism · Strain · Risk factor

Introduction

An anterior cruciate ligament (ACL) injury is one of the most common severe knee injuries in sports that typically occur during cutting, landing, and other sudden deceleration maneuvers [10, 33, 40]. There are approximately 200,000 ACL injuries and 100,000 ACL reconstructions in the United States each year [17, 37]. Although up to 80% of athletes following ACL reconstructions can return to sports, ACL re-injury rates are extremely high, particularly in young athletes (18%–37%) [7, 31, 52]. In addition, ACL injuries may cause long-term negative effects such as pain, functional impairments and knee osteoarthritis, resulting in a substantial economic problem and decreased quality of life [36, 41].

From a mechanical perspective, ACL injuries occur when excessive loading is sustained by the ACL [53]. Biomechanical analyses are crucial in developing clinical treatments and prevention programs [10]. Most previous studies used in vitro measurements or computer modeling methods to quantify ACL forces and strains [1, 6, 9, 30, 32, 38, 46]. These studies significantly contributed to the understanding of ACL's biomechanical functions and loading mechanisms through non-invasive and low-risk approaches. On the other hand, in vivo quantification of ACL loading provides excellent external validity to generalize the findings of ACL loading mechanisms to actual ACL injury events. Two techniques have been utilized to quantify ACL loading in vivo, including implantable strain sensors and innovative imaging techniques [54]. The strain sensors directly measure the ACL strain by invasively implanting the sensor on the anteromedial band of the intact ACL [2, 26]. The innovative imaging technique is non-invasive and achieved

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by matching three-dimensional (3D) knee models obtained from magnetic resonance imaging (MRI) to fluoroscopic knee images during static or dynamic activities. The 3D femur and tibia are translated and rotated with six degrees of freedom until they match the fluoroscopic images to quantify ACL length based on the origin and insertion locations of the ACL [12, 34].

Previously, researchers have reviewed the techniques and general findings of in vivo ligament/tendon loading studies, but there is a lack of summary of how movement biomechanics may affect in vivo ACL strain during a variety of tasks [5, 23, 43, 44]. One article reviewed in vivo ACL strain measured using a transducer in studies published before 2005 [39]. However, recently developed non-invasive imaging techniques and their application in athletic tasks were not included. Another article summarized techniques for measuring in vivo ACL strain [54] and included findings during rehabilitation exercises such as squatting, cycling, and stepping. But the effects of knee motion on ACL strain during athletic tasks were not included. Understanding in vivo ACL strain during a variety of tasks will provide information for understanding ACL loading mechanisms as a function of task difficulties and speeds. While ACL strain during slow-speed tasks might inform exercise selection for post-injury rehabilitation, ACL loading during athletic tasks could provide insight into ACL injury mechanisms and injury risk factors. Furthermore, the biomechanical variable that can cause increases in ACL strain across different tasks is likely to be the primary loading mechanism, which may be targeted for ACL injury risk screening via movement analyses and injury prevention through technique modification and neuromuscular training.

Therefore, the purpose of this systematic review was to summarize the studies that directly quantified in vivo ACL loading as a function of knee motion in healthy individuals during static, slow-speed, and athletic tasks. The summaries can provide direct information regarding in vivo ACL loading mechanisms to better understand ACL injury mechanisms and prevent ACL injuries.

Methods

Search Strategy

A systematic review of the literature was performed to identify studies evaluating in vivo ACL loading during various tasks. The search identified all articles containing the terms: "anterior cruciate ligament" or "ACL" and "in vivo" combined with "tensile", "strain", "stress", "force" and "loading". Terms including "Humans", "Full text", and "English" were used as search filters in the PubMed, CINAHL, SPORTDiscus, Cochrane Library and Web of Science databases. In addition, the reference list from each identified article was examined for any missed relevant studies.

Selection Criteria

Included studies were: (1) peer-reviewed studies published in English; (2) studies performed on humans who had intact ACLs; (3) studies directly measuring in vivo ACL loading and reporting knee mechanics. The exclusion criteria were: (1) studies only involving nonhuman subjects, computer models, robotic simulations or cadaveric testing; (2) studies didn't report in vivo ACL loading variables as a function of knee mechanics; (3) review articles or conference abstracts. Studies measuring the strain of an ACL graft were not included as the strain behavior of an ACL graft was likely to be different from an intact ACL and might be affected by ACL reconstruction techniques.

Risk of Bias and Quality

Methodological quality and risk of bias were assessed using the Appraisal tool (AXIS). Each of AXIS 20 components was rated as yes, not applicable, or no.

Searching Results

The flow diagram of searching the articles is shown in Fig. 1. This search yielded 452 results. The additional search of reference lists identified six relevant studies that have been missed through database searching. A total of 193 studies were excluded because of duplication, and 232 studies were excluded after reviewing the titles and abstracts. A total of 33 full texts were included for further discussion, and 6 of 33 studies were excluded based on the exclusion criteria. The remaining 27 studies were included in the systematic review. Eleven studies utilized strain sensors, while 16 studies applied imaging techniques. The numbers of studies for static or quasi-static postures (Table 1), slow-speed tasks (Table 2), and athletic tasks (Table 3) were 12, 11 and 4, respectively. Overall, all studies were considered fair studies with a low risk of bias based on the AXIS appraisal tool. One limitation was that no study reported the justification for the small sample size and how the sample was selected from the targeted population (Appendix A).

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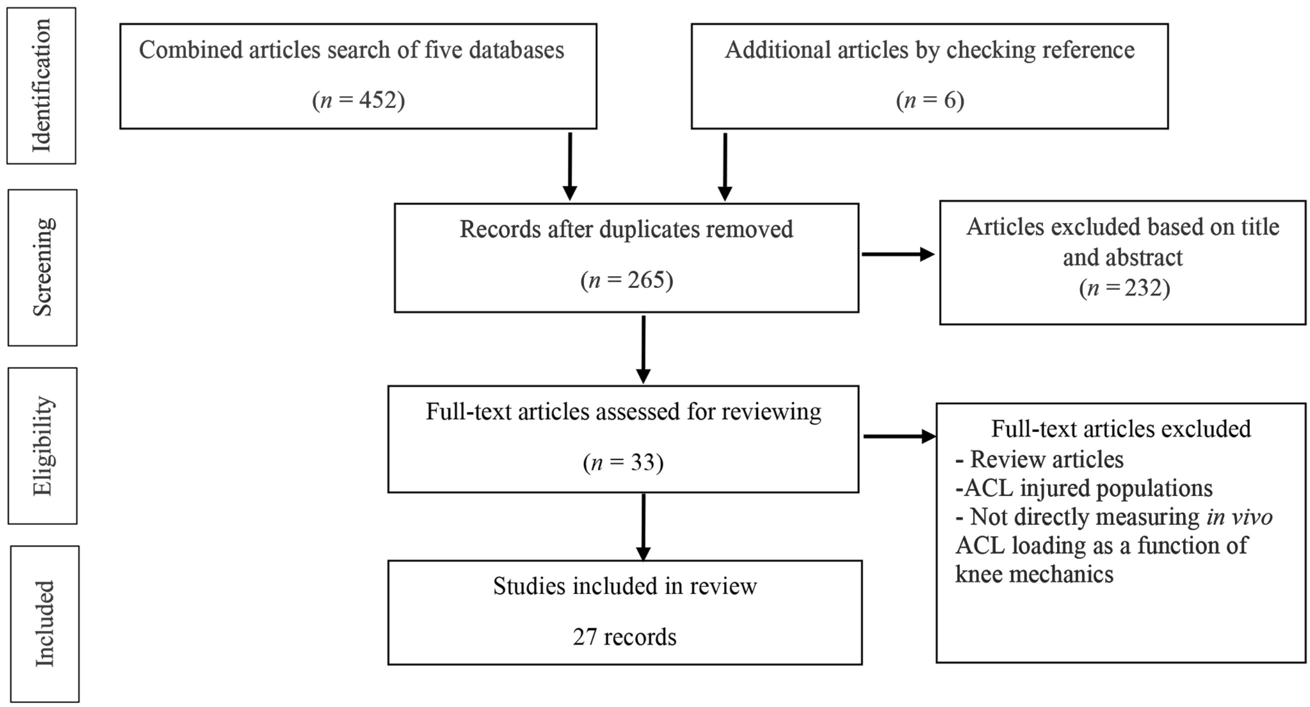

Fig. 1 Flow graph: Study selection process

Discussion

ACL Loading During Static or Quasi-static Tasks

Strain Transducer Studies

Howe et al. [26] used strain transducers to quantify the strain of the anterior-medial bundle of the ACL under a 150-N tibial anterior shear load at knee flexion of 30° and 90° and under isometric quadriceps contraction of 27 N·m at knee flexion of 30° and 90°. A greater ACL strain was observed when the knee was at 30° of flexion compared to 90° of flexion under the tibial anterior shear load. In addition, isometric quadriceps contraction resulted in increased ACL strain at 30° but not at 90° of knee flexion compared to no contraction conditions. The findings were supported by Beynnon et al. [2], who applied similar measuring techniques. The researchers found that a 150-N tibial anterior shear load produced greater ACL strain at 30° of knee flexion compared to 90° of flexion. In addition, the ACL was only strained when the knee flexion angle was less than 50° during active knee flexion-extension but was not strained during passive knee flexion-extension. Beynnon et al. [3] studied the effect of external loading (45-N weight applied at lower leg) on ACL strain during active knee flexion and extension movements. The external loading increased ACL strain, which was the greatest close to knee extension. The strained-unstrain transition occurred between 40° and 50° of knee flexion for the loaded condition and between 30° and 40° of knee flexion for the unloaded condition. In addition, isometric quadriceps activation to generate 15 and 30 N·m external moments loaded the ACL at 15° and 30° of knee flexion, while isometric quadriceps and hamstring co-activation without generating external moments also loaded the ACL at 15° of knee flexion.

Fleming et al. [18] used similar strain transducer methods to quantify the relationship between anterior tibial translation and ACL strain. The anterior-posterior tibial translation was recorded under an anterior shear load between 0 and 150 N at 30° or 90° of knee flexion. A significant correlation between anterior tibial translation and ACL strain was observed only at 30° but not at 90° of knee flexion. In addition, the anterior tibial translation also decreased at 30° of flexion compared to 90° of flexion under a 150 N tibial anterior shear force. Fleming et al. [21] further evaluated the effect of weight-bearing (40% body weight) and three-dimensional knee loading on ACL strain at 20° of knee flexion. A knee-loading fixture was used to apply isolated external loads to the tibia, including anterior-posterior shear force (– 90 N to 130 N), internal-external torque (– 9 N·m to 9 N·m), and varus-valgus moments (– 10 N·m to 10 N·m). The weight-bearing condition had greater ACL strain than the non-weight-bearing condition during all the external loading conditions, likely due to increased muscle activities and compressive loads. For the weight-bearing conditions, ACL strain increased linearly with tibial anterior

Table 1 ACL loading during static or quasi-static activities

Studies	Samples (Sex)	Techniques	Tasks	Findings
Strain Transducer				
Howe et al. [26]	5 Men	HEST	1. Anterior-posterior shear testing (150 N) at 30° and 90° of knee flexion 2. Isometric quadriceps contractions (27 N·m) at 30° and 90° of knee flexion	1. Anterior shear force of 150 N applied at 30° of knee flexion (4.0% ± 0.795%) produced a significantly higher strain than knee at 90° (2.3% ± 0.813%) 2. ACL strain significantly increased at 30° of knee flexion during isometric quadriceps contractions compared to the load was 0 N·m. However, no significant change when knee is held at 90° of flexion 3. Strain at 30° of knee flexion during isometric quadriceps contractions was significantly greater than knee at 90°
Beynnon et al. [2]	10 Men	HEST	1. Anterior shear force testing (150 N) at 30° and 90° of knee flexion 2. Isometric quadriceps contractions (27 N·m) at 30° and 90° of knee flexion	1. ACL strain was greater in 30° (3.7% ± 0.8%) than 90° of knee flexion (1.8% ± 0.9%) 2. Isometric quadriceps contraction produced higher ACL strain at 30° (3.2% ± 1.3%) than the value at 90° of knee flexion (–2.5% ± 3.2%) 3. ACL strain during active knee extension between 10° and 40° was significantly greater than knee is passively taken through 10° to 40°
Beynnon et al. [3]	11 (Not reported)	HEST	Active range of flexion-extension (5–90° of flexion) of the knee with (45 N) and without weight	1. A significant greater in the peak ACL strain values during with weight condition (3.8% ± 0.5%) compared to without weight condition (2.8% ± 0.6%) 2. The mean knee flexion angle at strained-unstrained transition in active motion with weight shifted 10° more for extension portion and 12° more for flexion portion compared without weight 3. With the knee at 15° of flexion simultaneous quadriceps and hamstrings contraction produced a significant ACL strain (2.8%)compared with the relaxed state, while no differences at 30° (0.4%), 60° (0%) and 90° of knee flexion (0%)
Fleming et al. [18]	5 Men	HEST	Anterior shear force (0–150 N) applied at 30° and 90° of knee flexion	When knee flexion at 30° with an ACL load at 90 or 150 N, ACL strain were 2.2 ± 1.2% and 3.0 ± 1.4%. When knee flexion at 90° with an ACL load at 90 or 150 N, ACL strain were 0.3 ± 0.4% and 0.9 ± 0.7%. ACL strain was a 30% decrease in 150 N of anterior shear load at 30° of knee flexion compared to loads at 90° of knee flexion
Fleming et al. [21]	2 Women and 9 Men	DVRT	1. 20˚ Knee flexion with simulated weight-bearing (40% bodyweight), and non-weight bearing 2. 20˚ Knee flexion with three externally applied loadings conditions (anterior-posterior (A-P) shear forces (between − 90 and 130 N); internal-external (I-E) torques (between − 10 N·m and 10 N·m); varus-valgus (V-V) moments (between − 10 N·m and 10 N·m)).	1. Weight-bearing significantly increased mean ACL strain values (2.1% ± 1.78%) in comparison to non-weight-bearing (− 2.0% ± 1.78%) 2. All three loadings could strain ACL, but A-P load could influence on strain for both the weight-bearing and non-weight-bearing conditions; I-E load could increase ACL strain when I-E torque was between − 2 to − 9N·m; Both ACL strains under weight-bearing (2%) and non-weight-bearing conditions (− 2%) kept consistent during application of V-V moments
Fleming et al. [22]	2 Women and 4 Men	DVRT	Electrical muscle stimulation at knee position at 5°− 30° of flexion	ACL strain values increased when the isolated contractions of the gastrocnemius muscle were contracted, and these decrease as the knee was flexed. Co-contraction of the quadriceps and gastrocnemius muscles produced higher ACL strain values than the isolated contraction of quadriceps or gastrocnemius muscles. Co-contraction of the hamstrings and gastrocnemius muscles increased ACL strain values relative to isolated contractions of hamstrings muscle
Imaging techniques				

Table 1 (continued)

Studies	Samples (Sex)	Techniques	Tasks	Findings
Li et al. [34]	5 (Not reported)	MRI and biplanar fluoroscopy	A static single leg lunge with full extension, 30°, 60° and 90° flexion	Compared to knee at full extension (32.5 ± 2.8 mm) and at 30° of flexion (32.5 ± 3.7 mm), the length of anterior-medial bundle of ACL at 60° decreased slightly to 30.7 ± 1.8 mm and a slightly decrease was found at 90° of flexion (30.2 ± 2.0 mm). The posterior-lateral bundle significantly decreased at 30 (26.3 ± 4.1 mm), 60 (24.1 ± 2.9 mm) and 90° (23.5 ± 2.3 mm) of knee flexion compared to full knee extension (27.6 ± 5.2 mm)
Englander et al. [12]	8 Men	MRI and biplanar fluoroscopy	A single legged static lunge position at various flexion angles (0–90° in increments in 15°)	ACL length (range from 27.8 ± 2.4 mm to 24.0 ± 1.9 mm) and patellar tendon tibial shaft angle (range from 23.1 ± 4.2° to − 1.9 ± 2.9°) decreased with knee flexion, and ACL length had positive correlation with patellar tendon tibial shaft angle (rho = 0.94 ± 0.07, $P < 0.05$)
Li et al. [35]	5 (Not reported)	MRI and biplanar fluoroscopy	A static single leg lunge with full extension, 30˚, 60˚ and 90˚ flexion	ACL length, elevation angle and deviation angle decreased with flexion. The elevation of the ACL at full extension, 30°, 60° and 90° flexion were 64.9° ± 10.7°; 52.6° ± 8.2°; 47.3° ± 6.1°; 43.2° ± 4.0°, respectively. The deviation angle was 11.1° ± 7.6° for full extension°; 11.2° ± 11.6° for 30°; 5.7° ± 7.4° for 60°; 2.3° ± 8.6° for 90° of flexion. ACL internal rotation angle increased as knee flexion angles increased (9.8° ± 10.2° at full extension, 17.4° ± 21° at 30° of flexion, 37.0° ± 31.5° at 60° of flexion, and 43.9° ± 34.5° at 90° of flexion)
Jordan et al. [29]	7 (Not reported)	MRI and biplanar fluoroscopy	A quasistatic single-legged lunge at selected angles (0°, 30°, 60°, 90°, 120° and 135°)	The length of anteromedial bundle increased with knee flexion. The longest length was 37.6 ± 3.1 mm at 0° and 37.9 ± 2.7 mm at 30°, and shortest length was 32.2 ± 2.4 mm at 135° of flexion. Posterolateral bundle was longest at 0° (30 ± 2.8 mm) and shortest at 135° of flexion (21.7 ± 3.5 mm). The elevation angles of the anteromedial bundle decreased consistently with flexion between 0°– 135° (range from 56.2° ± 6.1° to 23.2° ± 4.0°). The deviation angle of anteromedial bundle decreased from 16.3° ± 6.1° at 0° to 11.7° ± 5.4° at 90° of flexion
Hosseini et al. [28]	5 Women and 4 Men	MRI and biplanar fluoroscopy	A single leg at four flexion angles with no load condition (< 10 N) and full body weight condition (0°, 15°, 30°, 45°)	The ACL elongated increased as the load from no load (< 10 N) (27.1 ± 2.3 mm) to full body weight (27.5 ± 2.4 mm), and significant increased at 15° (5.9% ± 3.4%) and 30° of flexion (4.4% ± 3.4%) under body weight condition. If ACL was modeled as multiple surface fiber bundles, there was a dramatic difference between the relative elongations among the surface fiber bundles
Utturkar et al. [50]	8 Men	MRI and biplanar fluoroscopy	Single-legged positions: full extension, 30°of flexion, and a valgus collapse position (30° of flexion with 10°of external rotation of the tibia and maximal internal rotation at the hip)	Both functional bundles of the ACL followed similar trends with regards to decreases in length in each of the three positions. The knee position at full extension had a maximal ACL length (30.2 ± 2.6 mm), and the length decrease to 27.1 ± 2.2 mm when the knee flexion angle was 30°. When the knee was at the valgus collapse position, the ACL length was the least compared to other two conditions (25.6 ± 2.4 mm)

Note: *DVRT* Differential Variable Reluctance Transducer, *HEST* Hall Effect Strain Transducer, *MRI* Magnetic Resonance Imaging, *ACL* Anterior Cruciate Ligament

shear forces. In addition, internal rotation torques resulted in greater ACL strain than external rotation torques, while the ACL was strained but did not change significantly as a function of knee abduction or adduction moments [21]. Fleming et al. [22] also assessed ACL strain under electrically stimulated muscle contraction at three knee flexion angles (5°, 15° and 30°) in six healthy participants. ACL was strained under isolated contraction of the gastrocnemius when the knee flexion was less than 15°, and the gastrocnemius activation level positively correlated with ACL strain at knee flexion of 15°. At 15° and 30° of knee flexion, the addition of quadriceps activation to gastrocnemius activation could further increase ACL strain, while the addition of hamstring activation could decrease ACL strain [22]. Overall, these findings support the significant contribution of quadriceps activation, tibial anterior shear forces, and tibia translation on ACL loading at a small knee flexion angle. In addition, tibial internal rotation torques and gastrocnemius muscle forces could also load the ACL. Hamstring forces might decrease ACL loading with a constant quadriceps force, but quadriceps and hamstring co-activation without generating external moments might increase ACL loading.

Table 2 ACL loading during slow-speed activities

Studies	Samples (Sex)	Techniques	Tasks	Findings
Strain transducer				
Beynnon et al. [4]	8 (Not reported)	DVRT	Active flexion-extension with non-weight-bearing exercise, squatting with and without Sport Cord	The maximum ACL strain value produced at extension and decreased with the knee flexion. Maximum ACL strain values were similar among the three conditions (active flexion-extension were 3.8% ± 1.0%, squatting with weight-bearing exercise were 3.6% ± 1.3% and without Sport Cord were 4.0% ± 1.7%)
Fleming et al. [19]	2 Women and 6 Men	DVRT	Bicycling with three power levels (75, 125, and 175 W), each of which was performed at two cadences (60 and 90 rpm)	ACL strain had correlation with knee flexion angle over all conditions. The highest compressive force occurred at the end part of down stroke and increased with power level and decreased with cadence. Compressive force had significantly correlation with ACL strain but there were no differences in peak strain across the six conditions
Fleming et al. [20]	5 Men	DVRT	Stair climbing with cadences at 80 and 112 steps per minute	ACL strain increased when knee flexion angle was extended in both cadences. The peak ACL strain for the two cadences was not significantly different (2.69% ± 2.89% for the 80 steps per minute condition; 2.76% ± 2.68% for the 112 steps per minute condition)
Heijne et al. [27]	4 Women and 5 Men	DVRT	Step-up, step-down, the lunge, and the one-legged sit to stand	ACL strain increases when the knee was extended for each exercise. However, the mean peak ACL strain values were not significantly different between exercises. (step-up, 2.5% ± 0.36%; step-down, 2.6% ± 0.34%; the lunge 1.8% ± 0.62% for extension portion; 2.0% ± 0.5% for flexion portion; and the one-legged sit to stand, 2.8% ± 0.23% for flexion portion; 2.8% ± 0.27% for extension portion).
Imaging techniques				
Wu et al. [51]	2 Women and 6 Men	MRI and biplanar fluoroscopy	Gait (walk on treadmill at a speed of 0.67 m/s)	The relative elongation of the two bundles during the stance phase decreased similarly with increasing flexion, 12% ± 7% for the AM bundle and 13% ± 15% for the PL bundle. Sagittal and coronal elevation decreased but transverse plane deviation increased as knee flexion angles increased
Taylor et al. [49]	1 Woman and 7 Men	MRI, fluoroscopy, and marker-based motion capture	Normal level walking	The peak in ACL length (34.5 ± 1.4 mm) occurred approximately 60% of stance phase when knee flexion angle was near the lowest. There was an significantly inverse relationship between knee flexion angle and ACL length
Nagai et al. [42]	5 Women and 5 Men	MRI and biplanar fluoroscopy	Walking and running at self-selected pace	Both bundles relative elongations had similar pattern during walking and running. The maximum AM bundle relative elongation (6.7% ± 2.1%) during running was significantly greater than during walking (5.0% ± 1.7%). The maximum PL bundle relative elongation during walking (3.4% ± 2.3%) was significantly greater than running (1.1% ± 2.1%)
Englander et al. [11]	3 Women and 1 Man	MRI and biplanar fluoroscopy	Gait (walk on dual belt treadmill at a speed of 1 m/s)	Both average AM ($R^2 = 0.99$, $P < 0.01$) and PL bundle lengths ($R^2 = 0.99$, $P < 0.01$) were positivity correlated with overall ACL length. Flexion angle was a significant predictor of ACL length. ACL length decreased with decreasing extension (rho = − 0.68, $P < 0.01$)
Englander et al. [14]	3 Women and 7 Men	MRI and biplanar fluoroscopy	Walking on dual belt treadmill at a speed of 1 m/s)	The peak ACL length and peak strain were found at midstance (length 28.5 ± 2.6 mm, strain 5% ± 4%) and heel strike (length 30.5 ± 3.0 mm, strain 12% ± 5%) with low flexion angles (−2° ± 8° and − 4° ± 10°, respectively). Both bundles length negatively correlated with knee flexion during the stance phase (rho = −0.86, $P < 0.001$)
Englander et al. [15]	6 Men	MRI and biplanar fluoroscopy	Gait (walk on dual belt treadmill at a speed of 1 m/s)	ACL subdivisions that derived from randomly generated locations on the attachment site of ACL were highly correlated with ACL length throughout the full gait cycle (rho = 0.91 ± 0.1, $P < 5 \times 10^{-5}$)
Rao et al. [45]	6 Women and 16 Men	MRI and biplanar fluoroscopy	Single-legged lunge (0°−120°)	ACL strains decreased with flexion angle increased. As the knee angle is at 30°, AM had a 0.1% ± 3.9% strain and PL bundle had a − 9.6% ± 5.2% strain. When knee flexion angle was above 30, both bundles had negative strains showing minimal constraints

Note: *DVRT* Differential Variable Reluctance Transducer, *HEST* Hall Effect Strain Transducer, *MRI* Magnetic Resonance Imaging, *ACL* anterior cruciate ligament, *AM* anterior-medial, *PL* posterior-lateral

Table 3 ACL loading during athletic activities

Studies	Samples (Sex)	Techniques	Tasks	Findings
Strain transducer				
Cerulli et al. [8]	1 Man	DVRT	Single-leg forward hop (1.5 m)	ACL strain started to increase during the flight phase. The peak ACL strain occurred while ground reaction force reached peak (5.47% ± 0.28%). The strain still remained high level during the knee extension of landing
Imaging techniques				
Taylor et al. [48]	8 Men	MRI, fluoroscopy, and marker-based motion capture	Jump landing task: jump forward (0.47 m), then jump vertically with maximal effort	The peak ACL strain occurred 55 ± 14 ms prior to impact when ACL length was 12% ± 7% longer than an MRI based reference length. The length of the ACL during the jump landing decreased with increasing flexion angle
Englander et al. [49]	8 Men	MRI and biplanar fluoroscopy	Single-legged jump	Mean ACL strain had inverse relationship with flexion angle (rho = -0.73, $P < 0$ 0.001). Peaks ACL strain occurred before toe-off (7% ± 6%) and 55 ± 35 milliseconds before initial ground contact (7% ± 4%). Both the anterior-medial and posterior-lateral bundle lengths were positively correlated with overall ACL length (rho > 0.85, $P < 0.001$)
Englander et al. [16]	8 Men	MRI and biplanar fluoroscopy	Single-legged jump	Flexion angle was inversely related to patellar tendon angle ($R^2 = 0.95$, $P = 3.4 \times 10^{-9}$). ACL strain was positively related to patellar tendon strain ($R^2 = 0.65$, $P = 4.8 \times 10^{-4}$)

Note: *DVRT* Differential Variable Reluctance Transducer, *MRI* Magnetic Resonance Imaging, *ACL* anterior cruciate ligament

Imaging Studies

Li et al. [34] utilized MRI and dual fluoroscopic-based 3D models for measuring ACL elongation during a quasi-static single leg lunge at full extension, 30˚, 60˚ or 90˚of knee flexion in five participants. The length of the anterior-medial bundle of the ACL was generally similar among all the knee flexion angles, while the ACL appeared to be longer at full extension and 30° of flexion compared to the 60° and 90° of flexion. In addition, the posterior-lateral bundle significantly decreased at 60° and 90° of knee flexion compared to full knee extension. Li et al. [35] also modeled the ACL length, elevation angles, deviation angles, and twist angles as a function of knee flexion during a single-leg lunge. The findings showed that ACL length, elevation angle, and lateral deviation angles decreased, and the ACL internal rotation angle increased as knee flexion angles increased. Similarly, Jordan et al. [29] modeled the ACL as multiple bundles and showed that both anterior-medial and posterior-lateral bundles generally decreased their lengths and elevation angles as the knee angle increased from full extension to 90° of flexion. Hosseini et al. [28] measured ACL elongation with no load (< 10 N) and full body weight condition during single-leg lung with 0°, 15°, 30° and 45° of knee flexion. When the ACL was modeled as a single-bundle, ACL length increased under the full body weight condition and decreased as the knee flexion angle increased. However, when the ACL was modeled as eight bundles, different bundles demonstrated nonhomogeneous changes as a function of body weight and knee flexion. Englander et al. [12] quantified the correlation between ACL length, knee flexion angles, and patellar tendon shaft angles during a single-leg lunge. The results showed that patellar tendon tibial shaft angle and ACL length both had a negative correlation with flexion angle. In addition, there was a positive correlation between the patellar tendon tibial shaft and ACL length. When the knee angle was near extension, the patellar tendon shaft angle was the greatest. These studies suggested that the length of the entire ACL or the two bundles of the ACL generally decreased as the knee flexion angles increased. In addition, the ACL elevation angles also decreased as the knee flexion angle increased. With a constant tibial anterior shear force, an increased ACL elevation angle indicates that a greater ACL force is needed to generate the same amount of posterior force to resist the tibial anterior shear force. Furthermore, the patellar tendon shaft angle increased as the knee flexion angles decreased. With a constant patellar tendon force, an increased patellar tendon shaft angle indicated a greater tibial anterior shear force applied by the patellar tendon to the tibia, which may also increase ACL loading. The direct changes in ACL length and the indirect changes in ACL elevation and patellar tendon shaft angles all make the ACL susceptible to high loading at low knee flexion angles.

In addition to different knee angles, Utturkar et al. [50] evaluated the effect of different knee positions on ACL elongation in a single-leg lunge. The three knee positions included full extension, 30° of flexion, and a valgus collapse position (30° of flexion with 10° of external rotation of the tibia and maximal internal rotation at the hip). The knee position at full extension had a maximal ACL length, and the length decreased when the knee flexion angle was 30°. When the knee was at the valgus collapse position, the ACL length was the least compared to the other two conditions [50]. While the decreased ACL length associated with increased knee flexion was consistent with previous studies, these findings also suggested that knee valgus, which involved tibial external rotation, could decrease ACL length.

In summary, the above-mentioned studies measured ACL load under selected static or quasi-static postures with or without bearing weight-bearing. Both strain transducer and imaging studies showed that ACL loading reached the peak values as the knee near extension position and increased with increased tibial anterior shear force and decreased with increased knee flexion. A small knee flexion directly results in increased ACL length under no weight-bearing conditions and could amplify ACL loading through increased ACL elevation angles and increased patellar tendon shaft angles under constant patellar tendon loading. While ACL was primarily responsible for restraining tibial anterior shear forces, the tibial internal rotation torque might load the ACL, while external rotation might unload the ACL.

ACL Loading During Slow-Speed Tasks

Strain Transducer Studies

Beynnon et al. [4] compared ACL strain during active knee flexion-extension movements, bodyweight squats, and bodyweight squats with additional external loads. The findings showed the greatest ACL strain when the knee was close to extension for all three exercises, but the peak strain was similar among the three exercises despite the different external loading. Fleming et al. [19] quantified ACL strain during bicycling under three power levels and two cadence conditions. Peak ACL strain was observed when the knee flexion angle was small for all the conditions. However, higher power levels and lower cadence resulted in increased compressive forces on the pedal but similar ACL strain. Similarly, Fleming et al. [20] observed the maximum ACL strain values occurred when the knee was near extension during stair climbing, but the peak ACL strain was not significantly different between the two climbing cadences. Heijne et al. [27] investigated ACL strain during four closed kinetic chain activities: step-up, step-down, lunges, and one-legged sit-to-stand. Similarly, the greatest ACL strain was observed when the knee flexion angle was minimal, but a similar peak ACL strain was observed among the four tasks. Again, these studies found that peak ACL strain mostly occurred when the knee flexion angle was small during slow-speed tasks. However, increased whole-body external loading imposed by resistance bands, movement power and cadences, and exercise modalities did not appear to directly result in increased peak ACL strain.

Imaging Studies

Wu et al. [51] measured ACL elongation and orientation in three planes during treadmill walking using the MRI and dual-fluoroscopic imaging techniques. Both anterior-medial and posterior-lateral bundles of the ACL reached their peak elongation during the middle of the stance phase when the knee was close to full extension under full body weight. Consistently, a strong negative correlation was found between the elongation of the ACL and knee flexion angle during the stance phase. In addition, sagittal and coronal elevation angles decreased, and transverse plane deviation angles increased with increasing knee flexion. Taylor et al. [49] integrated MRI imaging, dual fluoroscopic techniques, and mark-based motion tracking technique to reproduce in vivo ACL kinematics during normal level walking. Similarly, the peak ACL strain occurred during the middle of the stance phase when the knee was close to full extension. Strong and negative correlations were observed between ACL strain and knee flexion angles. Nagai et al. [42] quantified in vivo ACL elongation during both treadmill walking and running. Peak ACL elongation occurred during the middle of the stance phase when the knee flexion angles were small for both walking and running. The anterior-medial bundle demonstrated greater elongation during running, while the posterior-lateral bundle showed greater elongation during walking. Both bundle elongation positively correlated with the tibia anterior-posterior translation and negatively correlated with knee flexion and extension during the stance phase. Englander et al. [11] utilized similar imaging techniques and applied an automatic registration algorithm to match the 3D surface model to the biplanar radiographs to quantify ACL length during walking. The findings showed that both anterior-medial and posterior-lateral bundles demonstrated similar patterns of changes and reached their peaks when the knee flexion angles were minimal during both the swing and stance phases. Again, knee flexion angles strongly and negatively correlated with ACL length. Englander et al. [14] further demonstrated that the overall ACL length, the anterior-medial bundle length, and the posterior-lateral bundle length were highly correlated with each other and all negatively correlated with the knee flexion angle throughout the gait cycle. Englander et al. [15] found that the strong correlation between the overall ACL length and different ACL bundles still existed even when 1000 virtual bundles were modeled by randomly sampling locations on the attachment site of the ACL. Overall, ACL subdivisions were all elongated when the knee was extended during gait, and different bundles of the ACL are not likely to demonstrate different patterns of changes as a function of knee flexion. Rao et al. [45] quantified ACL length at different knee flexion angles during a single-leg lunge and showed that both the anterior-medial and posterior-lateral bundles decreased their lengths at the knee flexion angle increased. The anterior-medial bundle was not strained when the knee flexion was greater than 30°, while the posterior-lateral bundle was not strained above 15° of knee flexion.

In summary, previous strain transducer and imaging technique studies in slow-speed tasks all demonstrated that peak ACL strain occurred when the knee flexion angle was close to minimal. In addition, most studies supported that the anterior-medial and posterior-lateral bundles demonstrated similar changing patterns and had strong negative correlations with knee flexion angles. The increased ACL elevation angle as a function of decreased knee flexion in walking was also consistent with the findings in semi-static postures. However, while exercise modalities such as running and walking might affect peak ACL loading, the relationship between the intensity of different tasks and ACL loading was not likely to be linear. These findings appeared to be inconsistent with previous studies that supported the role of weight-bearing and increased quadriceps forces on ACL loading, but several explanations may be provided. First, increased external loads or activity intensity might not directly increase quadriceps activation. A previous study showed that knee joint moments did not increase linearly as the external load increased during bilateral squats, and the increases in the ankle and hip moments were greater than the knee moments [24]. This could be due to participants' changes in body postures so that the weight vector of the external load stayed closer to the knee and imposed similar knee loading [25]. However, previous studies did not directly quantify knee joint moments or quadriceps activation when assessing ACL loading during tasks with different intensities. Second, increased quadriceps and tibial anterior shear forces could increase ACL loading at small knee flexion angles, but tasks with different intensities might result in peak ACL loading at different knee flexion angles. The potential differences in knee flexion angles might contribute to the lack of correlation between exercise intensity and peak ACL loading. Third, different from static or semi-static postures, dynamic tasks involve many muscle groups around the joint. The complex activation, co-activation, and coordination of quadriceps, hamstring, and gastrocnemius muscle forces, as well as tibiofemoral compressive force, could all contribute to the changes of the tibial anterior shear force to load the ACL but were not quantified in previous studies. Last, the small sample size and relatively large variability among participants might also limit the statistical power to detect significant differences.

ACL Loading During Athletic Tasks

Strain Transducer Studies

Cerulli et al. [8] assessed ACL strain during a single-leg forward hop task and showed that ACL strain started to increase during the flight phase and reached the peak strain shortly after initial contact with the ground while the ground reaction force arrived at the maximal value. These findings indicated that the ACL strain that occurred before the landing phase might result from the preparatory activation of the quadriceps along with small knee flexion angles. The increased quadriceps muscle forces and ground reaction forces at small knee flexion angles resulted in the peak ACL strain during the early landing phase.

Imaging Studies

Taylor et al. [48] used imaging techniques in combination with marker-based motion analysis techniques to quantify ACL strain during a double-leg jump-landing task. Interestingly, peak ACL length occurred approximately 55 ms prior to the initial contact of landing when the knee flexion angle was minimal. It appeared that ACL length decreased immediately before ground contact due to increases in knee flexion and slightly increased during the early landing phase, likely due to the increased quadriceps forces and ground reaction forces. Englander et al. [13] used MRI and high-speed biplanar radiography to determine ACL strain during a vertical single-leg jump-landing task. One peak in ACL strain was observed before the takeoff in the jumping phase, and another peak in ACL strain was found 55 ms prior to the initial contact landing, as shown in the previous study. However, ACL strain decreased during the landing phase, most likely due to the increased knee flexion angles after the initial contact of landing. In addition, both the anterior-medial and posterior-lateral bundle lengths positively correlated with the ACL length, which negatively correlated with knee flexion angles in both the jumping and landing phases. Englander et al. [16] further quantified the patellar tendon angles and strain, knee flexion angles, and ACL strain during a single-leg jump-landing task. In addition to the negative correlation between the ACL strain and knee flexion angle, strong and positive correlations were found between the patellar tendon strain and ACL strain. The patellar tendon demonstrated more horizontal orientation and high strain approximately 61 ms before initial contact of landing, which corresponded to the peak ACL strain. The findings support that the quadriceps could be highly activated in the flight phase. In the meantime, the patellar tendon angle was the greatest when the knee was close to extension, resulting in greater tibial anterior shear force to the tibia with a constant quadriceps force to load the ACL.

In summary, the findings in athletic tasks further supported the importance of knee flexion in affecting ACL loading. The increases in patellar tendon angle with decreased knee flexion could be a major factor in correlating knee flexion angles and ACL loading. In addition, ACL strain could be high or reach its peak during the pre-landing due to the high preparatory activation of the quadriceps at small knee flexion angles. ACL strain might decrease during the landing phase as the knee flexion angle increases. While the findings might appear to be inconsistent with the observation of ACL injuries, which commonly occur during the early phase of landing [47], it should be noted that the peak ACL strain during the flight phase was observed during non-injury events. The increases in knee flexion angles near the initial contact and after the initial contact could be a safe preparatory strategy that participants used to decrease ACL loading. However, with a small time window (50–60 ms) between the peak ACL strain and initial contact, midflight perturbation and unanticipated events might compromise the motor planning and control of this safe preparatory strategy. High quadriceps activation and impact ground reaction forces when the knee stays close to an extension during the early-landing phase might mimic the dangerous scenario when ACL injuries occur.

Summary and Implications

The current study has summarized in vivo ACL loading measured by strain transducers or imaging techniques during various static or quasi-static, slow-speed, and high-speed athletic tasks. First, there were strong and negative correlations between ACL elongation and knee flexion angles across tasks with different speeds and intensities. Peak ACL elongation mostly occurred when the knee flexion angle was minimal. A small knee flexion directly results in increased ACL length and can indirectly amplify the effects of quadriceps forces on ACL loading through increased ACL elevation angles and patellar tendon shaft angles. In addition, most studies supported that both the anterior-medial and posterior-lateral bundles changed similarly as a function of knee flexion angles. Second, the ACL was primarily resisting anterior tibia translation, and increased tibial anterior shear forces and patellar tendon forces would increase ACL loading when the knee is kept at a constantly small angle. Additionally, a high patellar tendon force could be generated by preparatory quadriceps activation to load the ACL even when the lower extremity was not in contact with the ground. Peak ACL elongation could occur during the pre-landing phase if the knee flexion starts to increase before the initial contact of landing during non-injury events. Third, while exercise modalities might affect peak ACL loading, the relationship between intensities and ACL loading during the same activity was complex and should not be assumed to be linear. Last, tibial internal rotation and knee adduction/abduction may make secondary contributions to ACL loading, while tibial external rotation might unload the ACL.

The information gained from this review may assist clinicians and researchers in better designing ACL injury screening tasks, ACL injury prevention programs, and ACL rehabilitation exercises to reach the long-term goal of reducing ACL injuries. First, ACL injury screening tasks should include primary ACL loading mechanisms, including knee flexion angles and quadriceps forces/activation, as well as secondary ACL loading mechanisms, such as tibial internal rotation and knee abduction. In addition, the pre-landing phase, during which peak ACL strain might occur in non-injury situations, should also be analyzed along with the early-landing phase to extract key variables for predicting future ACL injuries. Second, ACL injury prevention programs need to incorporate technique training that encourages increased knee flexion and soft landing, particularly for high-risk populations during activities with midflight perturbation and unanticipated events. Technique training under high-risk scenarios will better prepare individuals to adopt safe techniques in real sports situations. Third, exercises that involve increased knee flexion and decreased quadriceps activation at a slower speed might be considered during the early phase of post-ACL injury rehabilitation. Engaging the knee close to full extension with high quadriceps activation at a fast speed might be considered during the late phase of rehabilitation and assessed prior to return to play.

In the meantime, future studies to quantify in vivo ACL loading should consider other athletic tasks, such as cutting and pivoting tasks, with a larger sample size to compare different populations and task difficulties. It would be important to understand how sports environments and task demands might affect ACL strain to connect laboratory-based findings to the real world. In addition, more studies are needed to quantify the potential loading differences between original ACL and ACL grafts to better inform post-surgery rehabilitation and ACL re-injury mechanism. As non-invasive techniques continue to advance, future screening tasks might incorporate real ACL strain as a tool to differentiate the high-risk and low-risk populations.

Conclusion

Peak ACL elongation mostly occurred when the knee flexion angle was minimal. A high patellar tendon force could be generated by preparatory quadriceps activation to load the ACL even when the lower extremity was not in contact with the ground during athletic tasks. The relationship between exercise intensities and ACL loading was complex and should not be assumed to be linear. These findings may assist clinicians and researchers in better designing ACL injury screening tasks, ACL injury prevention programs, and ACL rehabilitation exercises.

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Data Availability Data sharing is not applicable to this review article as no datasets were generated or analyzed during the current study.

Declarations

Competing Interests The authors have no competing interests to declare that are relevant to the content of this article.

Ethical Approval/Consent to Participate/Consent to Publish. This is a review article. No ethical approval and consent are required.

Conflict of interest On behalf of all authors, the corresponding author states that there is no conflict of interest.

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