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LOADING MECHANISMS OF THE ANTERIOR CRUCIATE LIGAMENT

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Abstract

This review identifies the three-dimensional knee loads that have the highest risk of injuring the anterior cruciate ligament (ACL) in the athlete. It is the combination of the muscular resistance to a large knee flexion moment, an external reaction force generating knee compression, an internal tibial torque, and a knee abduction moment during a single-leg athletic manoeuvre such as landing from a jump, abruptly changing direction, or rapidly decelerating that results in the greatest ACL loads. While there is consensus that an anterior tibial shear force is the primary ACL loading mechanism, controversy exists regarding the secondary order of importance of transverse-plane and frontal-plane loading in ACL injury scenarios. Large knee compression forces combined with a posteriorly and inferiorly sloped tibial plateau, especially the lateral plateau—an important ACL injury risk factor—causes anterior tibial translation and internal tibial rotation, which increases ACL loading. Furthermore, while the ACL can fail under a single supramaximal loading cycle, recent evidence shows that it can also fail following repeated submaximal loading cycles due to microdamage accumulating in the ligament with each cycle. This challenges the existing dogma that non-contact ACL injuries are predominantly due to a single manoeuvre that catastrophically overloads the ACL.

Keywords

anterior cruciate ligament; knee; injury; mechanism

INTRODUCTION

An injury to the anterior cruciate ligament (ACL) can be debilitating for the athlete. For over 60 years, research has focused on trying to understand the risk factors (for recent review, see Pfeifer et al.[115]) and pathomechanics of non-contact ACL injuries,[12] with the overarching goal of preventing these injuries. This is critical given that non-contact ACL injuries account for nearly 80% of all ACL injuries.[121]

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To understand the aetiology of non-contact ACL injuries, scientists and clinicians have studied the anatomy of the ACL (for example, Duthon et al.[23]), its material properties,[14] and its function in stabilising the knee joint. They have investigated the forces and moments about the knee during various athletic activities associated with ACL injury, as well as the ACL's heterogeneous response to such loading.[77] To do so, various experimental approaches have been utilised, ranging from quantitative video analyses of actual ACL injuries (for example, Della Villa et al.[20]) to dynamic *in vitro* simulations of various injury scenarios employing animals[69] and human models (for example, Wojtys et al.[155]), to computer simulations (for example, Shin et al.[127]). The factors that can affect these forces and moments about the knee, and those applied to the ACL, include the geometry of the articulating surfaces of the tibiofemoral joint,[8] the tibiofemoral muscle forces,[126] and the friction between the athlete's shoe and the playing surface.[134] With the considerable amount of attention given to the ACL and ACL injuries, the body of research has grown exponentially. Hence, the purpose of this narrative review is to synthesise this literature to better understand the knowledge gained, to identify areas of consensus and those of debate, and to identify potential knowledge gaps. This will help guide future research aimed at better understanding the pathomechanics of non-contact ACL injuries towards the goal of better preventing these injuries.

ACL ANATOMY

ACL Attachment Locations

The ACL originates on the posterior-superior portion of the medial facet of the lateral femoral condyle and inserts anteriorly and slightly medially to the tibia's intercondylar eminence (Figure 1).[1] The shape of the ACL's femoral origin (or colloquially 'footprint') has often been described as oval or elliptic.[48,62,78,97,129,141,146,158] However, a more accurate description is that of a "segment of a circle" with a straight anterior border and a convex posterior border"[25]—a larger footprint that also includes the "fan-like extension fibres" or "indirect fibres" as described by Iriuchishima et al.[46] and Suruga et al.,[140] that extend and splay posteriorly to the articular cartilage of the femoral condyle (Figure 1A). For this reason, inconsistencies exist in the literature in terms of the surface area of the ACL's femoral footprint. Often, it has been quantified as smaller than that of the tibial footprint[28,45,47,62,66,125] if one excludes these indirect fibres. With their inclusion, however, the surface area of the femoral footprint is undoubtedly greater,[143] as effectively shown by Iriuchishima et al.[46] The tibial insertion of the ACL has been described as triangular,[104] mostly C-shaped, but ovoid and double-C-shaped as well,[125] duck-foot shaped[118] and most recently as paw-shaped, but sometimes quadrangular or irregular (Figure 1B).[62] The factors that lead to these variations in shape are as yet not understood.

The Structure of the ACL

The ACL is most often described as a "two-bundle" structure comprising the fibres of the antero-medial (AM) and the postero-lateral (PL) bundles,[112,114,132,133,137,146,161] named for the relative location of these fibres' insertion on the tibial plateau. A third bundle, the intermediate bundle, which is situated between the AM and PL bundles as its name suggests, has also been described, but less commonly.[62,79,104,145] Although controversy exists in the literature as to the number of bundles, what is of greater clinical importance is that the ACL is actually comprised of a continuum of fibres, each varying in tension during three-dimensional (3D) knee rotations.[1] Although the fibre bundles function in a complementary manner,[40] the AM fibres play a primary role in resisting anterior tibial translation,[17,87] whereas the PL fibres play a primary role in resisting internal tibial rotation.[160] The fibre contributions are also dependent on knee flexion angle, with a gradual transition from the anteriorly located fibres resisting peak loads at moderate knee flexion angles (i.e., 30°–60°) to the posteriorly located fibres resisting peak loads near full extension (i.e., 0°–15°).[32,43,49,98,123] This is because the location from where the PL fibres originate on the femur rotates toward the attachment site on the tibia as the knee flexes, thereby shortening the PL fibres. Also, the bundle of PL fibres, which is narrower and shorter,[53] elongates more than the AM fibres during weight bearing with a knee near full extension.[43] It is not surprising, therefore, that isolated ruptures of the PL fibres, which account for 20–41% of all partial ACL ruptures,[27,106,135,157] have been associated with pivoting events that include an axial rotational component; whereas isolated ruptures of the AM fibres, which account for 59–80% of all partial injuries,[27,106,135,157] have been associated with predominately anterior-directed forces applied to the knee joint.[130,159] It should be noted that the rate of isolated PL bundle ruptures are probably underestimated due to the difficulty in diagnosing these ruptures with conventional magnetic resonance imaging and arthroscopy.[13,113,136] These isolated ruptures of the PL and AM fibres were also associated with a "less energetic" injury and a "more explosive-type" knee trauma, respectively.[130,159] Accordingly, we suspect that the PL fibres, especially near the femoral enthesis,[13,135,159] are at a greater risk of a non-contact injury during athletic manoeuvres which often include an axial rotation component with a knee near full extension. Also, these types of injuries frequently occur during unremarkable, routine manoeuvres—perhaps defined as "less energetic"—that the athlete has performed countless times before (refer to 'ACL Failure Resulting from a Single Overload Event vs. Repetitive Submaximal Loading Events' subsection for a discussion on this topic).

BIOMECHANICAL FUNCTION OF THE ACL

Many years of research have shown that the main roles of the ACL are to resist primarily an anterior shear force on the proximal tibia relative to the distal femur, and secondarily an internal tibial torque relative to the femur and a knee abduction moment. To gain this insight, a large body of research focused on assessing ACL behaviour in response to simple, quasi-static knee loading scenarios.[11,34,71,85,93] This allowed for the examination of the ACL's response to single-plane loading (for example, Kennedy et al.[55]). While such loading scenarios may not be physiologically relevant to ACL injuries (i.e., no or minimal axial compression force, no significant trans-knee muscle forces, quasi-static loading, unrealistic magnitude and/or timing of applied forces), they can isolate the effect of each force/moment on ACL behaviour. This section will review this literature to summarise the ACL's role in resisting an anterior tibial shear force, an internal tibial torque, and a knee abduction moment. This section will also focus on the function of the ACL at small knee flexion angles (0–30°) given that most non-contact ACL injuries occur with the knee close to full extension (i.e., 10–30°).[20,60] Hence, unless otherwise stated, loading scenarios are reported for a knee at 0–30° of flexion.

Sagittal Plane Loading

There is widespread agreement in the literature that the ACL is the primary restraint to anterior translation of the proximal tibia relative to the distal femur, irrespective of knee flexion angle.[4,11,52,55,59,85,123,142,160] It provides 70–87% of the resistance to an anterior tibial shear force applied to the knee at small flexion angles (0–30°), and somewhat less of this resistance (62–85%) at larger flexion angles (60–90°).[4,11,59,142] In earlier human *in vitro* studies aimed at determining the role of the ACL in stabilising the knee, two main types of studies were conducted: (1) those that measured ACL strain or force in response to quasi-statically applied load(s) and (2) those that measured displacement of the tibia relative to the femur in response to a quasi-statically applied load(s) before and after transection of the ACL. The change in tibial displacement in response to the constant load (or similarly, the change in load to achieve the same tibial displacement) after ACL transection was inferred to be a measure of ACL function. In the former type of studies, an anterior tibial shear force significantly increased ACL strain,[26,55] ACL length,[28,42] and ACL force[74,82,123,142] in ACL-intact human knee specimens. In the latter type of studies, anterior tibial translation significantly increased after ACL transection in response to the same magnitude of anterior shear force applied to the tibia.[31,52,59,85,124,160] Similarly, after transection of the ACL, the anterior tibial shear force required to anteriorly translate the tibia the same distance significantly decreased.[4,11,75,124] In short, there is consensus that the ACL is the primary restraint to an anterior tibial shear force and the resulting anterior tibial translation, especially at small knee flexion angles. As we shall discuss later (‘Quadriceps Contraction‘ subsection), the primary source of that anterior shear force is the patellofemoral mechanism: the resultant force produced by large quadriceps and the patellar ligament tensile forces pushes the distal femur posteriorly relative to the proximal tibia, while drawing the proximal tibia anteriorly relative to the distal femur, thereby straining the ACL.

Transverse Plane Loading

Evidence of the ACL’s role in maintaining transverse-plane rotational stability of the knee, however, is less consistent. Some studies show that the ACL plays a secondary role in resisting internal tibial rotation at small knee flexion angles,[55,59,71,75,83,96,122,127] providing 10–20% of the resistance to an internal tibial torque;[4,59] meanwhile other studies show a minimal[85,93,103] or negligible role.[34,64,74,99,100,156] All but one of the studies that measured ACL loading in response to quasi-statically applied internal tibial torque reported a significant increase in ACL strain[26,55,83,127] or ACL force.[35,96] There is consensus, therefore, that an internal tibial torque significantly loads the ACL but only at small knee flexion angles (0–30°). As the flexion angle increases, the ACL’s resistance to an internal tibial torque decreases to a trivial role.[59,83,96] The disagreement between studies is among those that measured changes in internal tibial rotation in response to a constant internal tibial torque (or similarly, the change in internal tibial torque to achieve the same magnitude of internal tibial rotation) before and after ACL transection. While several studies showed a significant increase in internal tibial rotation (or a decrease in internal tibial torque),[4,31,33,50,59,71,75,107,122] a few studies showed a small increase[85,93,103] and others showed no increase.[34,64,99,100,156] Part of the disagreement in the literature may relate to how the change in tibial displacement following ACL transection is interpreted. Although most investigators interpret it as a measure of ACL function, we argue that it is mostly a measure of the tissue that is left intact in the ACL-deficient knee. Part of the disagreement may also relate to how an internal tibial torque loads the ACL. In the presence of an internal tibial torque, coupled anterior tibial translation occurs (see ‘Mechanical Coupling Between Knee Loads and Displacements’ section below for biomechanical explanation).[33,34,74,156] The consensus that internal tibial torque significantly strains the ACL, therefore, may be due to this coupled anterior tibial translation and to a smaller extent, internal tibial rotation. Consequently, ACL transection would cause large increases in anterior tibial translation[50] and smaller increases in internal tibial rotation in response to a constant internal tibial torque.[34,64,85,93,99,100,103,156] Hence, studies that assess the role of the ACL in resisting internal tibial torque via ACL transection are probably underestimating it, partly due to the mechanical coupling of anterior tibial translation with this axial torque.

Frontal Plane Loading

Evidence of the ACL’s role in maintaining frontal-plane rotational stability of the knee is contradictory. Some studies show that the ACL plays a minor but significant role in resisting a knee abduction moment,[4,26,42,83,88,127] providing 10% of the resistance at small flexion angles;[4] meanwhile other studies show a minimal[85,103] or negligeable role.[31,34,55] Most studies that measured ACL loading in response to a quasi-statically applied knee abduction moment reported a significant increase in strain[26,83,120,127] or length;[42] meanwhile Kennedy et al.[55] reported a mild relaxation of the ACL under knee abduction loading. The effect of a knee abduction moment on ACL strain/length, however, may be dependent on knee flexion angle, with a trend of increasing effect with increasing knee flexion angle.[42,96] For example, Miyasaka et al.[96] reported a significant increase in ACL strain when the knee was flexed at 30°, but no increase at smaller flexion angles (i.e., 0° and 15°). Since most non-contact ACL injuries occur with the knee flexed at 10–30°,[20,60] knee abduction loading may play a role in the mechanism of only a fraction of these injuries—those for which the knee flexion angle is at the greater end of this range. Similar to transverse-plane loading, there is disagreement among those studies that measured changes in knee abduction angle in response to a constant abduction moment before and after ACL transection. While two studies showed a significant increase in knee abduction angle,[4,88] two studies showed a minor increase[85,103] and two others showed no increase.[31,34] Mechanical coupling of knee motions also occurs in the presence of a knee abduction moment (see ‘Mechanical Coupling Between Knee Loads and Displacements’ subsection below for details). The significant increases in ACL strain reported under a knee abduction moment, therefore, seem to be largely due to a coupled anterior tibial translation and to a smaller extent, coupled internal tibial rotation. Consequently, ACL transection can cause significant increases in anterior tibial translation[30,51] and internal tibial rotation[30,51,88] that can parallel and perhaps cause those increases in knee abduction angle.[88]

Simultaneous Multiplane Loading

Given that an anterior tibial shear force, an internal tibial torque, and a knee abduction moment can all load the ACL, it is not surprising that the combination of these forces and moments induces the greatest load on the ACL (Figure 2).[81] This loading has also produced the greatest anterior subluxation of the medial and lateral tibial plateau during a simulated pivot shift in human knee specimens at about 25°of knee flexion.[105] Additionally, the combination of an internal tibial torque and a knee abduction moment produces significantly greater ACL forces and strain than either an internal tibial torque[51] or a knee abduction moment alone.[81,127] The magnitude of the effect of multiplanar loading on ACL loads, however, appears to be dependent on knee flexion angle, with internal tibial torque generating larger effects at very small knee flexion angles (−5° (hyperextension) to 10°) and knee abduction moments generating larger effects at larger angles (≥20°).[82] In short, multiplanar knee loading produces the greatest load on the ACL but appears to be moderated by the knee flexion angle at which loading is applied.

Mechanical Coupling Between Knee Loads and Displacements

Although one can attempt to isolate the effect of a single force or moment on ACL behaviour with simple quasi-static loading scenarios using a single-plane approach, the resulting knee displacement will almost always be multiplanar. This is because of the complex interaction between loads applied to the knee and the knee's kinematic response (Table 1). For instance, an anterior tibial shear force induces not only anterior tibial translation but also internal tibial rotation[29,32,34,42] with minimal to no knee abduction rotation (Table 1).[32,42,68] In addition, an internal tibial torque induces coupled anterior tibial translation[33,34,74,156] due to the location of the centre of rotation of the tibia in combination with the geometry of the articulating surfaces of the tibiofemoral joint. With a centre of rotation located at the medial tibial plateau, internal tibial rotation includes a component of anterior translation of the centre and lateral margin of the tibial plateau (Figure 3).[33] Under a knee abduction load, coupled anterior tibial translation[30,32,51] and internal tibial rotation[30,32,42,51,88,120] occurs (Table 1). Ren et al.[120] showed that a pure knee abduction moment causes an internal tibial torque, and thus a significant increase in ACL strain. It has also been proposed that internal tibial rotation can cause knee abduction rotation by tilting the longitudinal axis of the tibia in the frontal plane, thereby inducing abduction of the tibia relative to the femur. This is because of the posterior-inferior-directed slope of the tibial plateau.[88] Not surprisingly, therefore, a combination of an internal tibial torque and a knee abduction moment induces large coupled anterior tibial translation.[32,52,160] Mechanical coupling of knee displacements is also induced when an axial compressive force is applied to the knee, such as during a landing. This force can induce anterior tibial translation and internal tibial rotation due to the geometry of the articulating surfaces of the tibiofemoral joint (Figure 4). How and why this occurs is explained later in this review (see both 'Axial Knee Compressive Force' subsections).

A REVIEW OF ACL LOADING MECHANISMS

Most non-contact ACL injuries occur during athletic manoeuvres that include sudden dynamic loading of the knee joint, such as landing from a jump, cutting, pivoting, and decelerating.[20,101,138,149] To study these manoeuvres and how they load the ACL, *in vitro* models have been used that allow for precise measurement of the external forces and moments at the knee joint, as well as the resulting ACL strain and knee displacements. Such models aim to replicate knee loading that occurs during *in vivo* athletic manoeuvres. They provide a more accurate understanding of non-contact ACL injury mechanisms than the quasi-static models presented in the previous section ('Biomechanical Function of the ACL') that ignore physiologically relevant loading. Hence, this section will be utilising data obtained from these dynamic impulsive *in vitro* loading models to gain insight into how the three main forces/moments identified in the previous section—anterior tibial shear force, internal tibial torque, and knee abduction moment—can be produced during athletic manoeuvres.

Anterior Tibial Shear Force

Axial Knee Compressive Force.—During athletic manoeuvres such as jump landings, large axial compressive forces act on the tibiofemoral joint. These forces are produced by trans-knee muscle contractions, especially of the quadriceps to oppose the large flexion moment that occurs during these manoeuvres, and by the large ground reaction force acting across the knee, which also contributes to the flexion moment. The vertical ground reaction force reaches 3–4 times bodyweight during landings in a controlled laboratory environment,[36,110,144] and can exceed this magnitude in the field. Evidence indicates that axial knee compressive forces can significantly load the ACL.[80,84,94,150] This is because of the geometry of the articulating surfaces of the tibiofemoral joint: these joint compressive forces induce coupled anterior tibial translation,[80,84,94,151] as well as internal tibial rotation[84,94] (Table 1) and some knee abduction[84]—all knee displacements that are known to load the ACL (see 'Biomechanical Function of the ACL' section for details). Although these compression forces induce the greatest knee displacements at larger knee flexion angles (30–50°), significant displacements also occur at smaller flexion angles (10–30°)[80,84] when most non-contact ACL injuries occur.[20,60] The knee joint compression force acting along the longitudinal axis of the tibia has an anterior tibial shear force component which produces anterior tibial translation as a result of the posterior-inferior-directed slope of the lateral and medial tibial plateau (Figure 4A). In fact, Wang et al.[151] found that the magnitude of anterior tibial translation produced under an axial knee compression force was positively correlated with both the medial and lateral slopes of the tibial plateau: as the slopes increased, so did tibial translation. It is not surprising, therefore, that extensive evidence exists that the magnitude of the posterior-inferior-directed slope of the tibial plateau is a risk factor for ACL injury, especially that of the lateral tibial plateau.[10,24,37,38,119,131,139,148,162] ACL-injured knees have a larger lateral tibial slope than uninjured control knees.[10,24,37,38,119,131,139,147,148] Also, a greater slope of the lateral tibial plateau has been associated with greater anterior tibial acceleration,[92] and therefore greater ACL strain *in vitro*.[72,92]

Quadriceps Contraction.—Another sagittal-plane loading mechanism of the ACL is via the activation and stretch of the quadriceps during athletic manoeuvres.[2,102] For example, when the tensile force on an active muscle exceeds the force that the muscle can generate, the muscle is forcibly lengthened in a so-called 'eccentric' contraction. This lengthening can increase the muscle tension by 70% above its maximum isometric value,[54] a phenomenon that athletes exploit to create high muscle forces, such as during a stretch-shortening cycle. When the quadriceps is forcibly stretched as the knee flexes slightly during an athletic manoeuvre, therefore, a large quadriceps force is produced. There is consensus that the quadriceps muscle force acts via the patellofemoral mechanism to significantly increase anterior tibial translation[21,41] and internal tibial rotation,[41] thereby significantly loading the ACL.[9,22,70,86,150,153] This occurs because of the angle at which the quadriceps inserts onto the tibia via the patellar tendon; the anterior-superior-directed tensile force developed by the quadriceps at its tibial insertion results in an anterior tibial shear component and an axial knee compression component, both of which can strain the ACL (Table 1, Figure 4A). Near full knee extension (0–20°), the anterior tibial shear force component is greatest because of the relatively large insertion angle at the tibia. At larger knee flexion angles (>20°), this component is smaller, and the axial compressive force component is larger.[126] Hamstrings co-contraction can reduce ACL loading[22,70,111,154] by developing a posterior tibial shear force to oppose the anterior tibial shear force. Near full knee extension (~0°), however, the hamstrings are much less efficient because of the small angle at which they insert at the tibia, and therefore they develop relatively small posterior tibial shear forces.[111] On that account, can a large quadriceps force produce ACL loading high enough to injure it? Some disagreement exist in the literature,[21,89,90,116] but the general consensus is that this cannot occur under physiological knee loading scenarios.[90,116] In landing and deceleration tasks, in the sagittal-plane, peak ACL forces result from the complex interaction between the anterior tibial shear force produced by quadriceps contraction and knee compression, and the posterior tibial shear force induced by the ground reaction force applied to the lower leg.[90,116]

Internal Tibial Torque

Similar to the quasi-static knee loading models presented earlier ('Biomechanical Function of the ACL' section), *in vitro* dynamic knee loading models have consistently revealed that the application of an internal tibial torque significantly loads the ACL.[108] These models simulated landing manoeuvres by applying realistic knee loads consisting of an impulsive knee compression force and knee flexion moment, as well as trans-knee muscle forces.[56,57,72,108] They showed that adding an internal tibial torque, thus simulating a pivot landing, significantly increased ACL strain,[56,57,72,108] as well as coupled anterior tibial translation,[72,108] in comparison to landing without pivot. Such an internal tibial torque can be produced upon initial ground contact when the resultant ground reaction force is directed posteriorly and medially with respect to the shank (Figure 5). This can occur in a number of scenarios with varying combinations of joint mechanics at initial ground contact, such as suddenly decelerating and changing direction or axially rotating the trunk towards the ipsilateral knee prior to and upon ground contact.[18] The magnitude of this torque is moderated by two main factors: (1) axial knee compressive force interacting with the geometry of the articulating surfaces of the tibiofemoral joint and (2) coefficient of friction between an athlete's shoe and the playing surface.

Axial Knee Compressive Force.—Large knee compressive forces can significantly increase ACL loads[80,84,94,150] not only because of their anterior shear force component but also because of the coupled internal tibial rotation[41,84,94] due to the geometry of the articulating surfaces of the tibiofemoral joint (Table 1). This is especially true at larger knee flexion angles (30–50°), but also at smaller flexion angles (10–30°)[80,84] when most non-contact ACL injuries occur.[20,60] The coupled internal tibial rotation results from the steeper posterior-inferior-directed slope of the lateral tibial plateau in comparison with that of the medial tibial plateau. The lateral femoral condyle essentially pushes the steeper sloped lateral tibial plateau anteriorly more than the medial femoral condyle pushes the medial tibial plateau thereby causing internal tibial rotation (Figure 4B).[155] This concept is supported by McLean et al.,[91] who found that during a dynamic *in vivo* single-leg land-and-pivot manoeuvre, peak internal tibial rotation was significantly associated with the medial-to-lateral tibial slope ratio, which was governed by the lateral tibial slope. In addition, Wang et al.[151] found that the magnitude of internal tibial rotation produced under an axial compression force was positively correlated with the lateral tibial slope as well as with the lateral-to-medial tibial slope difference: the steeper the slope and the greater the difference between slopes, the greater the rotation.[151] This explains why a steep slope of the tibial plateau, especially of its lateral compartment, has been frequently identified as a significant risk factor for ACL injury.

Effect of Friction Between Shoe and Ground.—Upon initial ground contact during an athletic manoeuvre with an axial rotation component (e.g., pivot landing; plant-and-pivot task), the magnitude of internal tibial torque that can be applied to the knee joint is also governed by the coefficient of friction between the athlete's shoe and the ground. A higher coefficient can induce higher knee torques. For example, artificial grass with sand/rubber infill and soccer shoes with blade-type cleats produced greater axial torques of the lower leg than natural grass and stud-type cleats, respectively.[134] Therefore, one might expect athletes to generate greater internal tibial torque, thereby leading to greater risk of ACL injuries, under those high-friction shoe-playing surface conditions.[39,63,76] For instance, ACL injury rates in the NFL were found to be 67% higher on artificial surfaces (FieldTurf) than on natural grass surface.[39] Also, high school American football players using a cleat design shoe with higher rotational traction were more than three times more likely to injure their ACL than in a cleat with lower rotational traction.[63] It must be noted, however, that not all evidence supports a higher risk of ACL injury on artificial turf.[3,44] Nevertheless, biomechanical principles tell us that the coefficient of friction between the athlete's shoe is an important factor in the production of internal tibial torque. The controversy in the literature simply reveals the complex interaction between the factors that determine the magnitude and 3D direction of the knee and ACL loads.

Knee Abduction Moment

Amongst studies utilising *in vitro* dynamic knee loading models with impulsive loading previously described, there appears to be agreement that the application of a knee abduction moment significantly loads the ACL.[56,57,127,152] For instance, Withrow et al.[152] demonstrated that adding a knee abduction moment to a neutral landing (impulsive knee compression force, knee flexion moment, trans-knee muscle forces) increased peak ACL strain by 30%. The presence of a knee abduction moment not only increases the range of abduction rotation upon landing,[57] but it also increases anterior tibial translation and internal tibial rotation due to mechanical coupling.[30,32,51,109,120,155] It begs the question, which knee displacement dominates in terms of ACL loading? Although this coupled internal tibial rotation with a knee abduction moment has been demonstrated frequently, evidence against it exists.[57] This disagreement in the literature is discussed in greater detail below ('Worst-Case Knee Loading Direction for the ACL' subsection). Generally, a knee abduction moment is produced upon initial ground contact when the resultant ground reaction force is directed laterally with respect to the knee joint centre[117] to help the athlete accelerate in a contralateral direction (Figure 5).

Effect of Knee Flexion Angle

The angle of knee flexion can affect the magnitude of ACL loading, with this relation being modulated by the knee loading scenario. For instance, without any loading applied to a knee joint specimen being moved quasi-statically from full extension (0°) to 90°, ACL forces are highest in full extension and sharply decrease as the flexion angle increases with minimal forces measured at 30° and beyond.[96] Given that these associations between knee flexion angle and ACL loading are presented throughout this review within their respective sections in terms of loading scenario, a mere summary is presented here in tabular format (Table 2). In short, ACL loading tends to increase as the knee flexion angle decreases, which may explain why most non-contact ACL injuries occur with the knee close to full extension.[20,60]

Worst-Case Knee Loading Direction for the ACL

There is consensus that the 3D combination of knee joint compression and a knee flexion moment with an anterior tibial shear force, internal tibial torque, and a knee abduction moment (Figure 5) results in the greatest loads in the ACL during single-leg athletic manoeuvres, such as jump landings, abrupt turns, and sudden decelerations, performed with a knee close to full extension (i.e., 0–30°) than any of the forces or moments alone.[5,7,56,58,67,109,127]

Controversy exists, however, whether a knee abduction moment is secondary to an internal tibial torque with regard to its effect on ACL loading or vice versa[5,7,58,109] (both are secondary to an anterior tibial shear force which is the primary ACL loading mechanism, as presented in the 'Biomechanical Function of the ACL' section). Oh et al.[109] argued that it is the internal tibial torque that primarily causes large peak ACL strain. First, their *in vitro* dynamic knee loading model of a single-leg pivot landing (i.e., impulsive knee compression force, knee flexion moment, trans-knee muscle forces, as well as an axial tibial torque and a frontal-plane moment) was very sensitive to the direction of axial tibial torque[108,109] but not to the direction of the frontal-plane moment.[109] Peak ACL strain was found to be significantly greater when an internal tibial torque was applied in comparison with an external tibial torque, regardless of whether the frontal-plane moment was abduction or adduction. On the other hand, peak ACL strain remained the same whether a knee abduction or adduction moment was applied in the presence of an axial tibial rotation, regardless of its direction.[109] Second, the application of a knee abduction moment increased peak ACL strain by generating a coupled internal tibial rotation before a medial knee joint space opening could occur (see 'Mechanical Coupling Between Knee Loads and Displacements' subsection for biomechanical explanation and evidence). If internal tibial rotation was not coupled with a knee abduction moment, a larger medial knee joint opening would have to occur to strain the ACL. However, a concomitant injury to the medial collateral ligament injury with ACL injury only occurs 4–27% of the time.[65,95] Therefore, evidence supports the notion that it is the internal tibial rotation induced mainly by the knee compression forces and an internal tibial torque, and also by a knee abduction moment, that plays a greater role in loading the ACL (secondary to an anterior tibial shear force, the primary ACL loading mechanism).

Conversely, a group working with a different *in vitro* dynamic knee loading model argued the opposite—that internal tibial rotation is secondary to knee abduction motion and anterior tibial translation in increasing ACL loading.[7,56–58,67] First, while investigating the timing sequence of multiplanar knee kinematics during an *in vitro* simulated landing, Kiapour et al.[58] found that internal tibial rotation increased and peaked significantly later than knee flexion, anterior tibial translation, knee abduction, and ACL strain. However, such time lag in internal tibial rotation has not been reported in other *in vitro*-simulated dynamic landing models[107] or in *in vivo* single-leg landings.[61] Second, minimal coupled internal tibial rotation was found with a knee abduction moment, but significant coupled knee abduction motion was present with an internal tibial torque.[57] This result contradicts much of the literature, which shows a strong mechanical coupling between a knee abduction moment and internal tibial rotation,[30,32,42,51,109] as discussed earlier (see 'Mechanical Coupling Between Knee Loads and Displacements' subsection). Additionally, there is concern with how the external moment were applied to the knee in this *in vitro* dynamic landing model. Prior to the simulated ground impact, the knee is preloaded with an external anterior tibial shear force, knee abduction moment, and internal tibial torque, in addition to the trans-knee muscle forces.[6] It is unusual for the knee to be placed under static external pre-loads prior to ground contact during a landing *in vivo*. In contrast, the *in vitro* dynamic knee loading model previously discussed applied impulsive axial compression, knee flexion moment, internal tibial torque, and knee abduction moment.[107–109] Hence, evidence appears to favour an internal tibial torque as the second most critical to ACL loading, with a knee abduction moment having a tertiary order effect.

ACL Failure Resulting from a Single Overload Event vs. Repetitive Submaximal Loading Events

Current dogma considers implicitly that an ACL tear happens under a single loading cycle that exceeds the ultimate tensile strength (UTS) of the healthy ligament (one cycle on x-axis above UTS* on y-axis in Figure 6).[20,101,128] While such a single overload injury event can and does happen, newer research suggests that partial or complete ACL tears can also occur after repetitive near-maximal loading events (below the healthy ligament's UTS), each causing enough microdamage to accumulate and coalesce to weaken the structure. Unable to be repaired in time, the ligament fails under a submaximal load in a seemingly normal manoeuvre that has been performed numerous times before without injury (Figure 6).[15,19,73,155] While little can be done to modulate the intensity of an athlete during competition, it is possible to adjust their training to limit the number and intensity of the unnecessarily strenuous (for the ACL) repetitions in training, especially of the likely very small subset of manoeuvres known to significantly load the ACL to near maximum levels. If these could be recognised and counted, they could be limited to allow sufficient time for any microdamage to repair and for the ultimate strength of the ACL to increase back to that of a healthy ligament. Based on this review, the loading mechanism most likely to induce ACL failure, whether in a single or under repetitive loading, would be a manoeuvre that gives rise to the combination of knee joint compression, anterior tibial shear, internal tibial torque and knee abduction moment during a jump landing, abrupt change of direction, or sudden stop.

CONCLUSIONS

In summary, there is consensus that the 3D combination of knee joint compression and a knee flexion moment combined with an anterior tibial shear force, internal tibial torque, and a knee abduction moment results in the greatest loads in the ACL during single-leg athletic manoeuvres including jump landings, abrupt turns, and sudden decelerations performed with a knee close to full extension (i.e., 0–30°). While there is consensus that an anterior tibial shear force is the primary loading mechanism of the ACL, especially at small knee flexion angles, controversy still exists regarding the secondary order of importance of transverse-plane and frontal-plane loading in ACL injury scenarios. In addition, mechanical coupling between 3D knee loads and displacements adds to the complexity of characterising ACL loading mechanisms. Due to the geometry of the articulating surfaces of the tibiofemoral joint, single-plane loading usually causes multiplane knee displacements. For instance, large knee compression forces combined with a posteriorly and inferiorly sloped tibial plateau, especially with a steeper lateral plateau, causes anterior tibial translation and internal tibial rotation and increases ACL loading. Accordingly, a steep lateral tibial slope has been identified repeatedly as an important risk factor for ACL injury. Lastly, recent evidence has emerged suggesting that while the ACL can fail under a single supramaximal loading cycle, it can also fail under repetitive submaximal loading due to microdamage accumulating with each successive loading cycle, thereby weakening the ligament. This challenges the widely accepted view that an ACL injury only occurs during a single loading event and has implications for better ACL injury prevention in the future.

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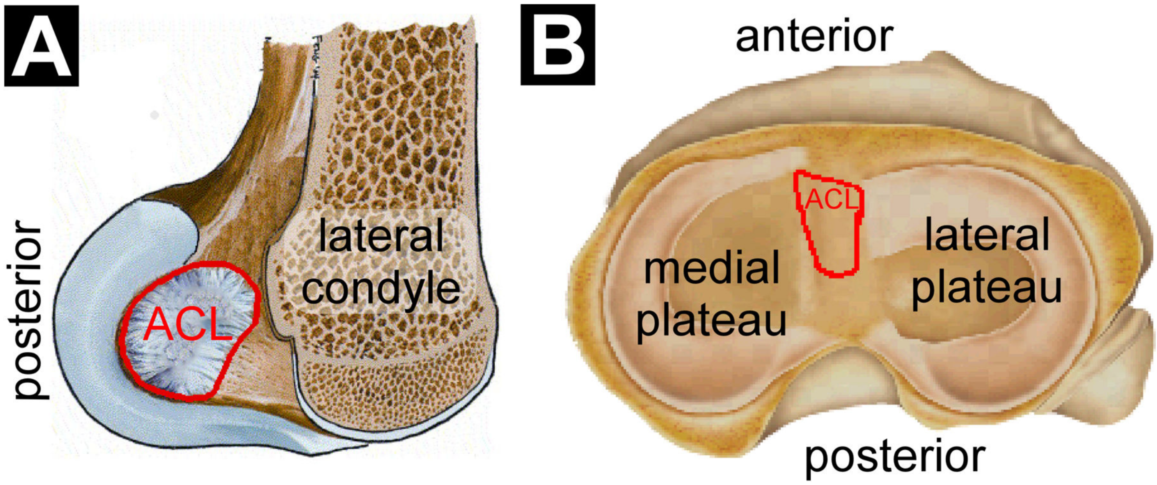

Figure 1.

Illustration of the (A) femoral origin and (B) tibial insertion sites of the ACL (outlined in red). Modified from Chhabra et al.[16] and Lord et al.[75] with permission from Elsevier.

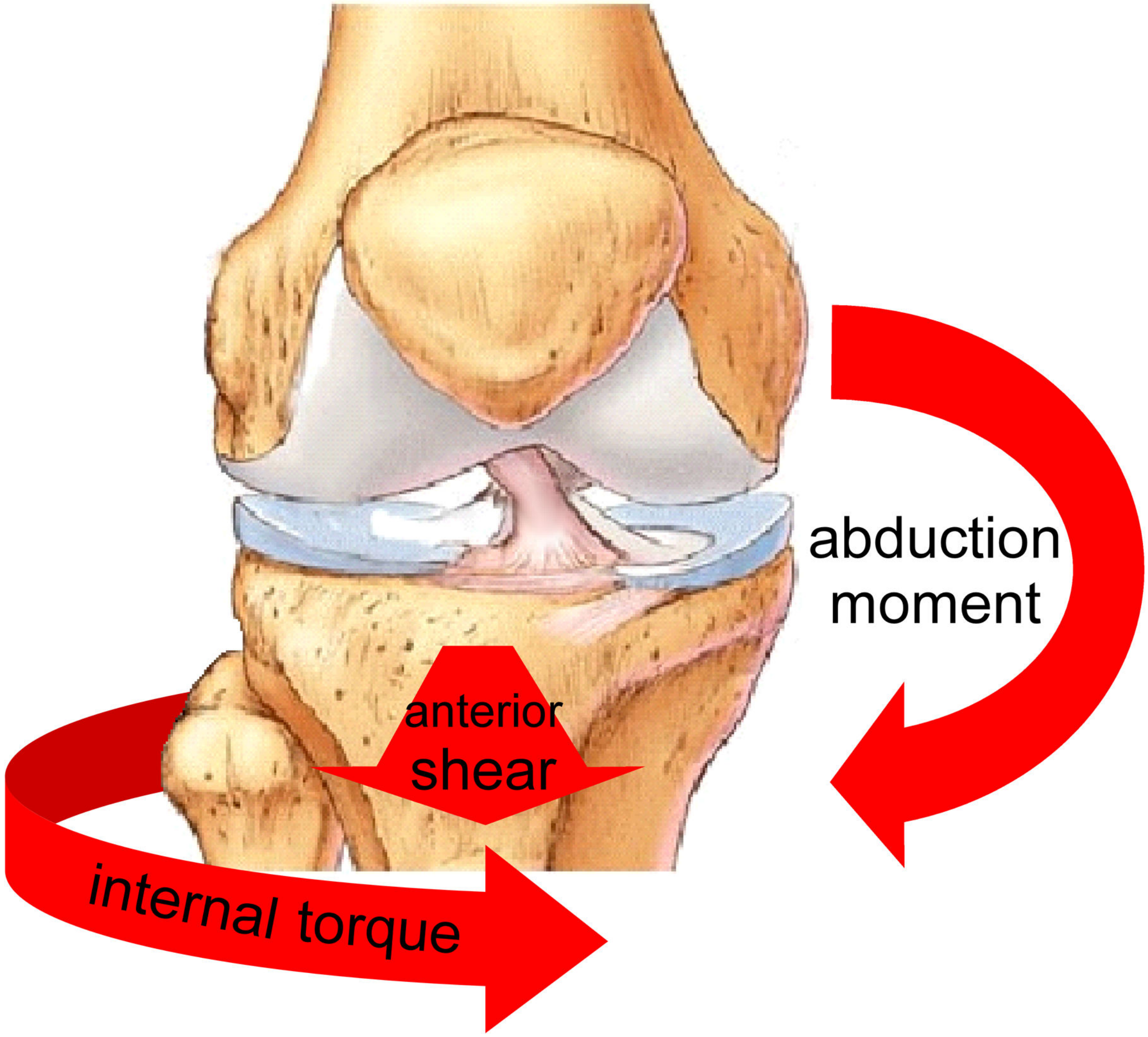

Figure 2.
The combination of an anterior tibial shear force, an internal tibial torque, and a knee abduction moment induces the greatest load on the ACL.

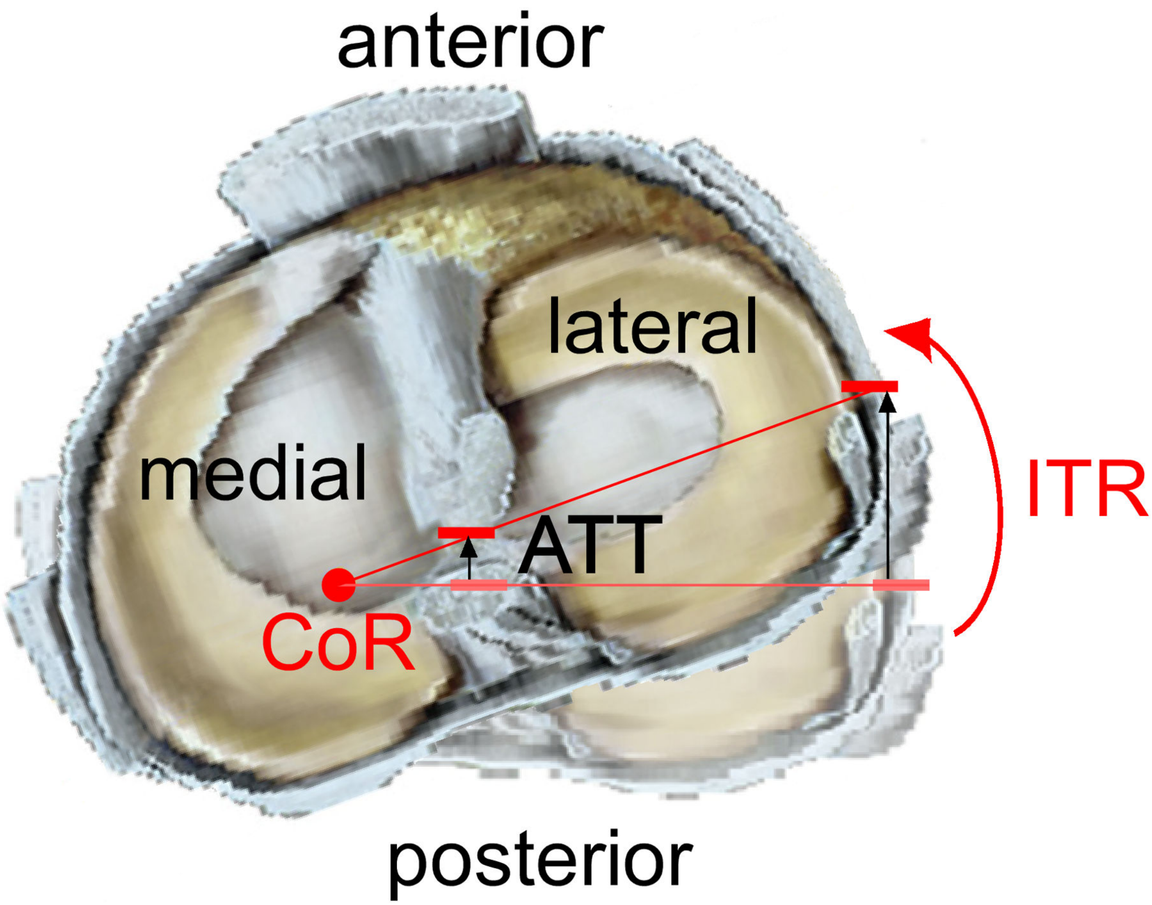

Figure 3.

Superior view of a right knee illustrating how an internal tibial torque can produce internal tibial rotation (ITR) and coupled anterior tibial translation (ATT). Because the centre of rotation (CoR) of the tibia is located at the medial tibial plateau, the centre and lateral margin of the tibial plateau translate anteriorly as the tibia rotates internally relative to the femur. Modified from Gardner et al.[33] with permission from Elsevier.

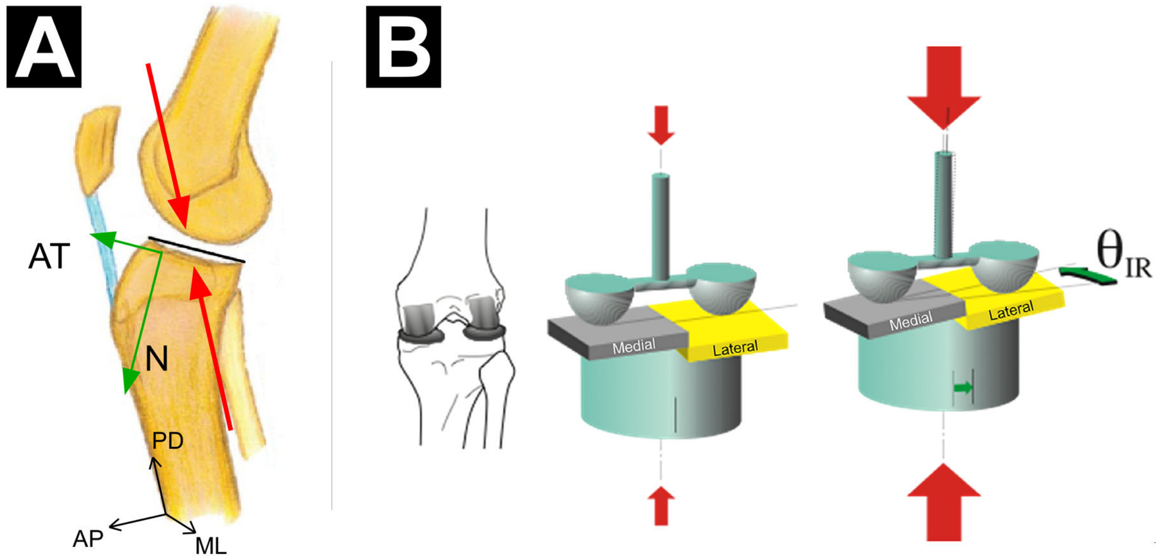

Figure 4.
(A) Sagittal-plane view of a right knee illustrating how an axial knee compression force can induce coupled anterior tibial translation because of the geometry of the articulating surfaces of the tibiofemoral joint. As a result of the posterior-inferior-directed slope of the lateral and medial tibial plateau, the knee compression force (red arrows, acting along the proximal-distal (PD) axis of the tibia) has a component that induces an anterior tibial shear force (green arrow, AT), in addition to the normal force (green arrow, N, force component perpendicular to the surface of the tibial plateau). (B) Postero-superior three-quarter view of a right knee illustrating how a large axial knee compression force (large red arrows) can induce coupled internal tibial rotation. As a result of the steeper posterior-inferior-directed slope of the lateral tibial plateau in comparison with that of the medial tibial plateau, the lateral femoral condyle essentially pushes the lateral tibial plateau anteriorly more than the medial femoral condyle pushes the medial tibial plateau thereby causing internal tibial rotation (IR). Reproduced from Wojtys et al.[155]

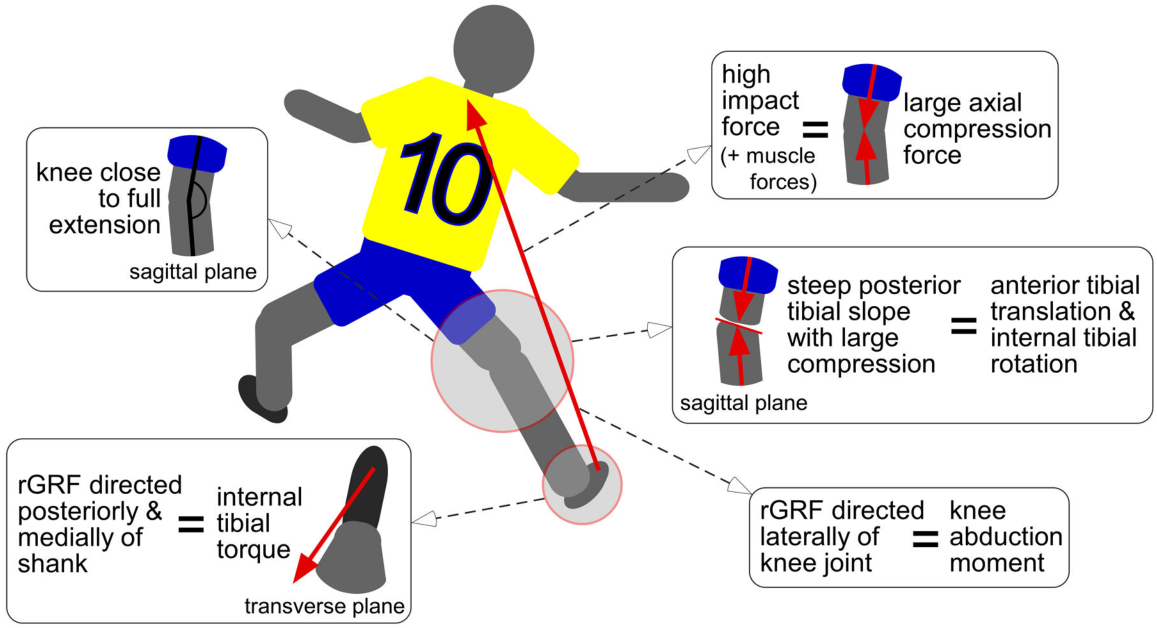

Figure 5.

A rear-view illustration of an athlete landing and cutting to the left. The combination of the muscular resistance to a large knee flexion moment and a large upward external reaction force (rGRF) generating knee compression, an internal tibial torque, and a knee abduction moment during a single-leg athletic manoeuvre such as landing from a jump, abruptly changing direction, or rapidly decelerating with a knee close to full extension has the highest risk of injuring the ACL in the athlete. The large knee compression force resulting from the rGRF and the trans-knee muscle forces can induce coupled anterior tibial translation and internal tibial rotation due to the posterior-inferior-directed slope of the tibial plateau, especially that of the lateral compartment. An internal tibial torque can be produced when the resultant ground reaction force (rGRF) applied under the ball of the foot is directed posteriorly and medially with respect to the shank as the athlete decelerates and pushes laterally. In the frontal plane, a knee abduction moment is produced when the rGRF is directed laterally with respect to the knee joint centre.

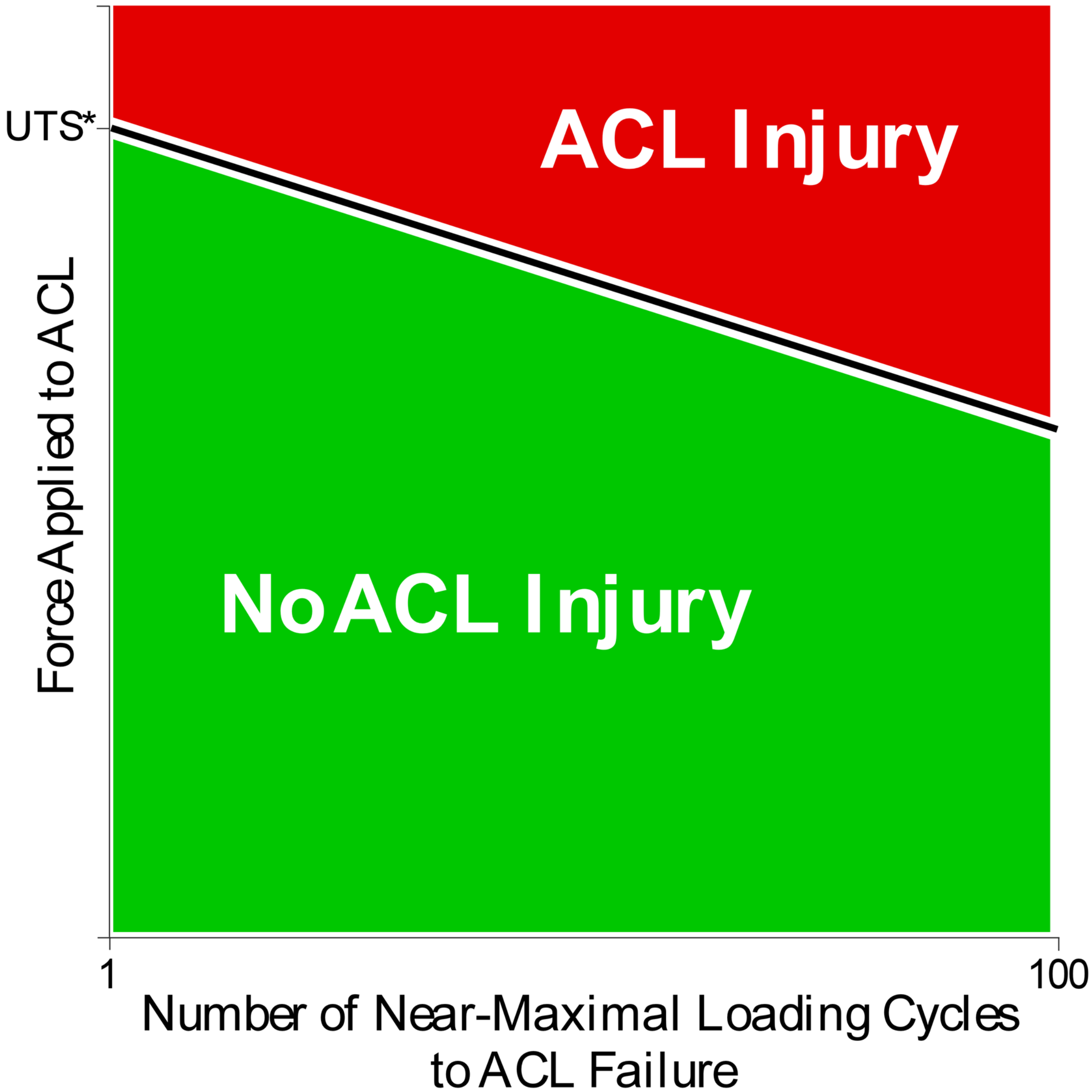

Figure 6.

Diagram illustrating the inverse relationship between force applied to the ACL during athletic manoeuvres and the number of near-maximal loading cycles it can withstand before failing. Evidence suggests that as the force applied to the ACL decreases per loading cycle, the number of loading cycles leading to ACL failure increases, and vice versa. The precise nature of this relationship is unknown, however, and most likely depends on many factors such as age, the time interval between cycles and the ACL's structural properties. It should be noted that these near-maximal ACL loading cycles represent a very small fraction of the total ACL loading cycles that occur in a typical practice session. UTS* refers to the ultimate tensile strength of a healthy ligament.

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Table 1.

List of knee loads and the associated coupled knee displacements.

Knee Loading	Coupled Knee Displacements	Figure Illustrating Coupling
anterior tibial shear force	anterior tibial translation internal tibial rotation	N/A
internal tibial torque	internal tibial rotation anterior tibial translation	Figure 3
knee abduction moment	knee abduction rotation anterior tibial translation internal tibial rotation	N/A
axial knee compression force	anterior tibial translation internal tibial rotation	Figure 4A Figure 4B

Table 2.

Relation between knee flexion angle (0–90°) and the magnitude of ACL loading for various knee loading scenarios. The section of the review where this information is discussed is also presented.

Knee Loading	Relation Between Flexion Angle and ACL Loading	Review Section
anterior tibial shear force	greatest ACL loading at 15–30°; moderate at 0° and 60°; lowest at 90°	Biomechanical Function of the ACL / Sagittal Plane Loading
internal tibial torque	ACL loading increases as knee flexion decreases; significant loading at 0–30°	Biomechanical Function of the ACL / Transverse Plane Loading
knee abduction moment	ACL loading decreases as knee flexion decreases	Biomechanical Function of the ACL / Frontal Plane Loading
quadriceps contraction	ACL loading increases as knee flexion decreases; significant loading at 0–20°	A Review of ACL Loading Mechanisms / Anterior Tibial Shear Force / Quadriceps Contraction
axial knee compressive force	ACL loading, ATT, ITR and ABD decrease as knee flexion decreases;	A Review of ACL Loading Mechanisms / Anterior Tibial Shear Force & Internal Tibial Torque / Axial Knee Compressive Force

ATT: anterior tibial translation; ITR: internal tibial rotation: ABD: knee abduction angle