SUPPLEMENT

Mechanisms of non-contact ACL injuries

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In soccer one of the most common knee injuries is the anterior cruciate ligament (ACL) tear, which usually occurs through non-contact mechanisms. Female soccer players are at higher risk of sustaining non-contact ACL injuries than male soccer players. A good understanding of ACL loading mechanisms is the basis for a good understanding of the mechanisms of non-contact ACL injuries, which in turn is essential for identifying risk factors and developing prevention strategies. Current literature demonstrates that sagittal plane biomechanical factors, such as small knee flexion angle, great posterior ground reaction force and great quadriceps muscle force, are the major ACL loading mechanisms. A great posterior ground reaction force may be associated with a great quadriceps muscle force, which would cause great anterior draw force at the knee. A small knee flexion is associated with a large patella tendon-tibia shaft angle and ACL elevation angle, which would result in great ACL loading. Current literature also demonstrates that the ACL is not the major structure of bearing knee valgus-varus moment and internal-external rotation loadings. Knee valgus-varus moment and internal-external rotation moment alone are not likely to result in isolated ACL injuries without injuring other knee structures.

Anterior cruciate ligament (ACL) injury is one of the most commonly seen injuries in sport and has a devastating influence on patients' activity levels and quality of life. Gottlob *et al*[21] estimated that approximately 175 000 primary ACL reconstruction surgeries were performed annually in the USA with an estimated cost of over US$2 billion. Complete ACL rupture can induce other pathological knee conditions including knee instability, damage to menisci and the chondral surface, and osteoarthritis. Studies have repeatedly shown that patients with complete ACL rupture have chronic knee instability and secondary damage to menisci and chondral surfaces.[19 28 30]

ACL injuries that occur without physical contact between athletes are referred to as non-contact ACL injuries[8 17 18] and most occur through a non-contact mechanism of injury in sports in which sudden deceleration, landing and pivoting manoeuvres are repeatedly performed.[8] Female athletes had a higher incidence of ACL injuries compared with their male counterparts.[2 18] Studies have shown that the incidence in female athletes is two to eight times higher than in males in soccer, basketball and volleyball.[2 7 18 22 25 26 36]

As with other sports injuries, understanding injury mechanisms is a key component of preventing non-contact ACL injuries.[36a] The research effort to determine the risk factors for sustaining non-contact ACL injuries is increasing as concerns grow about the larger number of incidents, the greater treatment costs and the serious consequences of non-contact ACL injuries. Prospective cohort studies commonly use epidemiological research designs for determining injury and disease risk factors,[48] and are being used for determining the risk factors for sustaining non-contact ACL injuries.[24] The results of epidemiological studies with cohort designs, however, are descriptive in nature and do not examine the cause-and-effect relationship between identified risk factors and the injury.[48] Without a good understanding of the injury mechanisms, the risk factors for sustaining non-contact ACL injuries identified from epidemiological studies could be misinterpreted and lead to the selection of non-optimal injury prevention programs. The purpose of this review was to examine biomechanical studies relating to the mechanisms of non-contact ACL injuries.

Mechanically, ACL injury occurs when an excessive tension force is applied on the ACL. A non-contact ACL injury occurs when a person themselves generates great forces or moments at the knee that apply excessive loading on the ACL. Therefore, an understanding of the mechanisms of ACL loading during active human movements is crucial for understanding the mechanisms and risk factors for non-contact ACL injuries. Berns *et al*[5] investigated the effects of combined knee loading on ACL strain on 13 cadaver knees. The strain of the anterior medial bundle of the ACL was recorded using liquid mercury strain gauges at 0° and 30° knee flexion. The results of this study showed that the anterior shear force on the proximal end of the tibia was the primary determinant of the strain in the anterior medial bundle of the ACL, while neither pure knee internal-external rotation moment or pure knee valgus-varus moment had significant effects on the strain of the anterior medial bundle of the ACL. The results of this study further showed that anterior shear force at the proximal end of the tibia combined with a knee valgus moment resulted in a significantly greater strain in the anterior medial bundle of the ACL than did the anterior shear force at the proximal end of the tibia alone.

Markolf *et al*[39] also investigated the effects of anterior shear force at the proximal end of the tibia and knee valgus, varus, internal rotation and external rotation moments on the ACL loading of cadaver knees. A 100-N anterior shear force and 10-Nm knee valgus, varus, internal rotation and external rotation moments were added to cadaver knees. The ACL loading was recorded as the knee was extended from 90° of flexion to 5° hyperextension. The results of this study showed that an anterior shear force on the tibia generated significant ACL loading, while the knee valgus, varus and internal rotation moments also generated significant ACL loading only when the ACL was loaded by the anterior shear force at the proximal end of the tibia. The results of this study further showed that the ACL loading due to the anterior shear force combined with either a valgus or varus moment to the knee was greater than that due to the anterior shear force alone, while the ACL loading due to the anterior shear force combined with a knee external rotation moment was lower than that due to anterior shear force alone. The knee valgus and external rotation moment loadings are elements of dynamic valgus that many current ACL injury prevention programs are trying to avoid.[24] The results of the study by Markolf *et al*[39] also showed that ACL loading due to the combined knee varus and internal rotation moment loading was greater than that due to either knee varus moment loading or internal rotation moment loading alone, and that the ACL loading due to combined knee valgus and external rotation moment loading was lower than that due to either knee valgus or external rotation moment loading alone. Finally, the results of this study showed that the ACL loading due to the anterior shear force and knee valgus, varus and internal rotation moments increased as the knee flexion angle decreased.

Fleming *et al*[20] studied the effects of weight bearing and tibia external loading on ACL strain. They implanted a differential variable reluctance transducer on the anterior medial bundle of the ACL of 11 subjects. ACL strains were measured in vivo when a subject's leg was attached to a knee loading fixture that allowed independent application of anterior-posterior shear force, valgus-varus moments and internal-external rotation moments to the tibia and simulation of a weigh-bearing condition. The anterior shear force was applied on the proximal end of the tibia from 0 N to 130 N in 10-N increments. The valgus-varus moments were applied to the knee from −10 Nm to 10 Nm in 1-Nm increments. The internal-external rotation moments were applied to the knee from −9 Nm to 9 Nm in 1-Nm increments. The knee flexion angle was fixed at 20° during the test. The results of this study showed that ACL strain significantly increased as the anterior shear force at the proximal end of the tibia and the knee internal rotation moment increased, while knee valgus-varus and external rotation moments had little effect on ACL strain under the weight-bearing condition.

The above-mentioned studies consistently showed that the anterior shear force at the proximal end of the tibia is a major contributor to ACL loading, while the knee valgus, varus and internal rotation moments may increase ACL loading when an anterior shear force at the proximal end of tibia is applied. According to these ACL loading mechanisms, a small knee flexion angle, a strong quadriceps muscle contraction or a great posterior ground reaction force can increase ACL loading.

Quadriceps muscles are the major contributor to the anterior shear force at the proximal end of the tibia through the patella tendon. Arms *et al*[3] and Draganich and Vahey[16] found that quadriceps force can significantly increase the in vitro ACL strain from 0° to 45° of knee flexion. Durselen *et al*[15] showed that the applied quadriceps force causes the ACL to sustain a high level of strain from full extension to 30° of knee flexion and that the ACL strain then starts to decrease as the knee flexion angle increases to more than 30°. Beynnon *et al*[6] had similar results in that the in vivo isometric quadriceps force significantly strained the ACL at knee flexions of 15° and 30° relative to the relaxed condition. Shoemaker *et al*[50] showed that the quadriceps had a significant effect on the tibial anterior displacement and the ACL graft tension. The largest graft tension due to the quadriceps occurred at 35° of knee flexion. A recent study by DeMorat *et al*[14] showed that a 4500-N quadriceps muscle force could cause ACL injuries at 20° of knee flexion. Eleven cadaver knee specimens were fixed to a knee simulator and loaded with a 4500-N quadriceps muscle force. Quadriceps muscle contraction tests at 400 N (Q-400 tests) and KT-1000 tests were performed before and after the 4500-N quadriceps muscle force loading. Tibia anterior translations were recorded during the Q-400 and KT-1000 tests. All cadaver knee specimens were dissected after the tests to determine the ACL injury states. Six of the 11 specimens had confirmed ACL injuries (three complete ACL tears and three partial tears). All specimens showed increased tibia anterior translation in the Q-400 and KT-1000 tests.

Decreasing knee flexion angle increases the anterior shear force at the proximal end of the tibia by increasing the patella tendon-tibia shaft angle. With a given quadriceps muscle force, the anterior shear force at the proximal end of the tibia is determined by the patella tendon-tibia shaft angle, defined as the angle between the patella tendon and the longitudinal axis of the tibia.[43] With a given quadriceps muscle force, the greater the patella tendon-tibia shaft angle, the greater the anterior shear force on the tibia. Nunley *et al*[43] studied the relationship between the patella tendon-tibia shaft angle and the knee flexion angle with weight bearing. Ten male and ten female college students without a known history of lower extremity injuries were recruited as subjects. Sagittal plane *x* ray films were taken for each subject at 0°, 15°, 30°, 45°, 60°, 75° and 90° knee flexions bearing 50% of the body weight. Patella tendon-tibia shaft angles were measured from the *x* ray films. Regression analyses were performed to determine the relationship between patella tendon-tibia shaft angle and knee flexion angle, and compare the relationship between genders. The results of this showed that the patella tendon-tibia shaft angle was a function of knee flexion angle with an increase in patella tendon-tibia shaft angle as the knee flexion angle decreased, and that, on average, the females' patella tendon-tibia shaft angle was 4° greater than that of the males. The relationship between the patella tendon-tibia shaft angle and knee flexion angle found by Nunley *et al*[43] was consistent with those from other studies on the patella tendon-tibia shaft angle under non-weight-bearing conditions.[9 51 53]

Decreasing knee flexion angle also increases ACL loading by increasing ACL elevation angle and deviation angle, defined as the angle between the longitudinal axis of the ACL and the tibia plateau and the angle between the projection of the longitudinal axis of the ACL on the tibia plateau and the posterior direction of the tibia, respectively.[35] The resultant force along the longitudinal axis of the ACL equals the anterior shear force on the ACL divided by the cosines of the ACL elevation and deviation angles. The greater the ACL elevation and deviation angles are, the greater the ACL loading is with a given anterior shear force on the ACL. Herzog and Read[23] show that a decrease in knee flexion angle results in a more vertical line of action of the ACL, which means an increase in the ACL elevation angle. Li *et al*[35] determined the in vivo ACL elevation and deviation angles as functions of the knee flexion angle with weight bearing. Five young and healthy volunteers were recruited as subjects. The ACL elevation and deviation angles at 0°, 30°, 60° and 90° of knee flexion with weight bearing were obtained using individualised dual-orthogonal fluoroscopic images and MR image-based three-dimensional models. The results of this study showed that both ACL elevation and deviation angles increased as the knee flexion angle decreased.

Several studies show that ACL loading increases as the knee flexion angle decreases. Arms *et al*[3] studied the biomechanics of ACL rehabilitation and reconstruction and found that quadriceps muscle contraction significantly strains the ACL from 0° to 45° of knee flexion, but did not strain the ACL when knee flexion is greater than 60°. Beynnon *et al*[6] measured the in vivo ACL strain during rehabilitation exercises and found that isometric quadriceps muscle contraction resulted in a significant increase in ACL strain at 15° and 30° knee flexions, while it resulted in no change in ACL strain relative to the relaxed muscle condition at 60° and 90° of knee flexion. Li *et al*[33 34] investigated the quadriceps and hamstring muscle loading on ACL loading and showed that the in situ ACL loading increased as knee flexion angle decreased when the quadriceps muscles were loaded regardless of the hamstring muscle loading conditions.

The literature also shows that individuals at high risk of sustaining non-contact ACL injuries have a smaller knee flexion angle during athletic tasks than individuals at low risk. Epidemiological studies show that female athletes are at higher risk of sustaining non-contact ACL injuries than their male counterparts[2 18 27 36 37 45 46 46a] Recent biomechanical studies demonstrated that female recreational athletes exhibited small knee flexion angles in running, jumping and cutting tasks.[12 38] Studies also demonstrate that female adolescent athletes had a sharply increased ACL injury rate after 13 years of age.[49 54] A recent biomechanical study showed that female adolescent soccer players started decreasing their knee flexion angle during a stop-jump task after 13 years of age.[57] These results combined suggest that small knee flexion angle during landing tasks may be a risk factor for sustaining non-contact ACL injuries.

Increasing peak posterior ground reaction forces during athletic tasks increases ACL loading by inducing an increased quadriceps muscle contraction. A posterior ground reaction force creates a flexion moment relative to the knee, which needs to be balanced by a knee extension moment generated by quadriceps muscles.[58] As previously described, the quadriceps muscle contraction adds an anterior shear force on the proximal end of the tibia through the patella tendon. The greater the posterior ground reaction force is, the greater the quadriceps muscle force is, and the greater the ACL loading is (Yu *et al*, 2006). Cerulli *et al*[11] and Lamontagne *et al*[32] recently recorded in vivo ACL strain in a hop landing task. A differential variable reluctance transducer was implanted on the middle portion of the anterior-medial bundle of the ACLs of three subjects. Subjects then performed the hop landing task in a biomechanics laboratory. Force plate, EMG and in vivo ACL strain were recorded simultaneously. The results of this study showed that the peak ACL strain occurred at the impact peak vertical ground reaction force shortly after initial contact between the foot and the ground. Yu *et al*[59] demonstrated that peak impact vertical and posterior ground reaction forces occurred essentially at the same time. These results combined suggest that a hard landing with a great impact posterior ground reaction force may be a risk factor for sustaining non-contact ACL injuries.

The literature shows that individuals at high risk of sustaining non-contact ACL injuries have greater peak posterior ground reaction forces in athletic tasks. Chappell *et al*[12] studied the lower extremity kinetics and the kinematics of college recreational athletes during landings in stop-jump tasks. Their results showed that female recreational athletes had greater peak resultant proximal tibia anterior shear force and knee joint resultant extension moment during landings in stop-jump tasks than male recreational athletes. Yu *et al*[55] studied the immediate effects of a newly designed knee brace with a constraint to knee extension during a stop-jump task. Their results showed that the female college recreational athletes had greater peak posterior ground reaction force during landing in the stop-jump task than did their male counterparts. Yu *et al*[59] showed that the resultant peak proximal tibia anterior shear force was positively correlated to the peak posterior ground reaction force.

That hamstring co-contraction protects the ACL has been a long-standing clinical concept because hamstring muscles provide a posterior shear force on the tibia that is supposed to reduce the anterior shear force on the tibia from the patellar tendon and thus unload the ACL. Recent scientific studies, however, did not support this concept. Li *et al*[33] showed in a cadaver study that hamstring co-contraction did not significantly decrease tibia anterior translation when the knee flexion angle was less than 30°. Beynnon *et al*[6] found that the isometric hamstring co-contraction of the hamstring muscles did reduce in vivo ACL strain between 15° and 60° of knee flexion. Kingma *et al*[31] found that hamstring muscle activation increased only 1.3–2.0 times while knee extension moment increased 2.7–3.4 times when the knee flexion angle was between 5° and 50°, which did not suggest a hamstring recruitment pattern to reduce the ACL loading. O'Connor[60] Pandy *et al*[61] and Yu *et al*[56] studied ACL loading using a modelling and computer simulation approach and all showed that the hamstring muscles did not reduce ACL loading at all when the knee flexion angle was small.

Although biomechanical studies showed that the knee valgus moment was not a major mechanism of ACL loading, a recent epidemiological study by Hewett *et al*[24] reported that external knee valgus moment in a vertical drop landing-jump task was a predictor of ACL injuries. A total of 205 high-school soccer, basketball and volleyball players were followed for three competition seasons. Knee flexion and valgus angles at initial foot contact with the ground and the maximum knee flexion and valgus angles and maximum moments during the stance phase of the vertical drop landing-jump task were recorded prospectively for every subject. Nine subjects sustained ACL injuries after three competition seasons. The results of this study showed that knee abduction angle at landing was 8° greater in ACL-injured than in uninjured athletes, and that ACL-injured athletes had a 2.5 times greater peak external knee valgus moment and a 20% higher peak vertical ground reaction force than did uninjured athletes. The results further showed that peak external knee valgus moment predicted anterior cruciate ligament injury status with 73% specificity, 78% sensitivity and a predictive r^2 value of 0.88. The results of this study appear to suggest an association of knee valgus angle and moment with ACL injuries.

However, we may have to be cautious when interpreting the association of knee valgus angle and moment with non-contact ACL injuries observed in the study by Hewett *et al*.[24] The observed pre-injury knee valgus moments of the nine subjects who suffered ACL injuries in this study were less than 0.12 Nm/body weight/standing height. The averaged body weight and standing height of the injured subjects in this study were 62 kg and 1.68 m. This means that the pre-injury knee valgus moments of the nine subjects injured in this study were less than 12.5 Nm. These knee valgus moment loadings were similar to those in the studies by Berns *et al*,[5] Markolf *et al*[39] and Fleming *et al*[20] that demonstrated that knee valgus loading did not significantly affect ACL loading unless a significant proximal tibia anterior shear force was applied. Further, several other studies in the current literature demonstrate that knee valgus moment loading alone cannot injure the ACL when the medial collateral ligament (MCL) is intact. Bendjaballah *et al*[4] studied the effects of knee valgus-varus moment loading on cruciate and collateral ligament loadings using a finite element model. Their results suggest that cruciate ligaments are not major valgus-varus moment loading bearing structures when collateral ligaments are intact. Matsumoto *et al*[40] investigated the roles of the ACL and MCL in preventing knee valgus instability using cadaver knees. Their results demonstrate that MCL is the major structure stopping medial knee space opening. Mazzocca *et al*[41] tested the effect of knee valgus loading on MCL and ACL injuries. They found that the response of the ACL strain to knee valgus moment loading was minimal when the MCL was intact but significantly increased after the MCL rupture started due to knee valgus moment loading. Their results show that the ACL still had about 60% of its original strength after complete MCL rupture with medial knee space openings greater than 15 mm due to knee valgus moment loading. This study clearly demonstrates that there is unlikely to be a complete ACL rupture due to knee valgus moment loading without a complete MCL rupture (grade 3 injury), while clinical observation shows that most ACL non-contact ACL injuries are not accompanied by significant MCL injuries.

A recent study by Fayad *et al*[62] showed that only five out of a total of 84 contact and non-contact ACL injuries were accompanied by a complete MCL rupture. These studies combined suggest that knee valgus moment loading alone is not likely to be a major ACL loading mechanism that can result in ACL rupture or a major risk factor for sustaining non-contact ACL injuries. More scientific studies are needed before we can confidently interpret the association of knee valgus angle and moment with non-contact ACL injuries as a sole risk factor for sustaining non-contact ACL injuries.

In summary, the current literature clearly suggests that sagittal plane biomechanics are the major mechanism of ACL loading. Decreased knee flexion angle and increased quadriceps muscle force and posterior ground reaction force causing an increased knee extension moment are requirements for increased ACL loading. Although the external knee valgus moment has been demonstrated to be associated with ACL injuries, the current literature contains no evidence that knee valgus-varus and internal-external rotation moments can produce non-contact ACL injuries in and of themselves without these high sagittal plane forces.

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