STRENGTH AFFECTS SAGITTAL PLANE KNEE BIOMECHANICS THAT MAY PROTECT THE ACL DURING VARIOUS CUTTING MANOEUVRES.

by

William Davies

This Research Project is submitted as partial fulfilment of the requirements for the degree of Master of Science, St Mary's University.

1st Supervisor: Daniel Cleather

2nd Supervisor: Stephen Patterson

ABSTRACT

Strength has been associated with reduced ACL incidence in sport but the specific mechanisms are less clear. The purpose of the investigation was to examine how strength may alter biomechanics to reduce injury risk during these manoeuvres, as well as explore how knee angle varies in response to different cutting conditions. Fourteen trained subjects were split into a stronger and weaker group (28.6 vs 17.5 N/kg^{-1}), determined by relative peak isometric strength in a unilateral squat. Each participant was fitted with reflective markers, and lower limb, sagittal plane, kinetics and kinematics were evaluated using a tri axial force plate, and a 16 camera motion analysis system. Subjects performed 18 trials divided into six different cutting conditions, comprising of 2 cutting angles (45° and 90°), at 2, 4 and 6 $m.s^{-1}$. A mixed design ANOVA (2x2x3) determined that between strength groups (n=7), significant increases in hip extensor moments ($p < 0.05$), and reductions in knee extensor moment ($p < 0.05$) during weight acceptance in the strong group were observed. In addition, knee angle at initial contact was more flexed in the strong group ($p = 0.05$). When group data was collapsed, significant peak knee flexion differences existed for cut angle ($p < 0.001$), but not for velocity. Results suggest increasing lower extremity strength may reduce ACL loading by redistributing impact forces to the hip and alleviating stress at the knee, as well as reducing ACL strain directly by a more flexed knee position at ground contact. In addition, peak flexion posture is dependent on cut angle, not velocity, perhaps as a way to increase excursion to reduce ground reaction forces, and knee extensor moments, during movements with greater multi planar stresses.

Key Words: kinetics, injury prevention, approach velocity, cut angle, kinematics

Table of Contents

List of Figures

List of Tables

ACKNOWLEDGEMENTS

I would like to thank Aspire Academy for use of their biomechanics facilities, in particular Joong Hyun Ryu for his help and patience when showing me how to use Vicon.

I would also like to thank my wife Carla, and my family for their continued support.

CHAPTER 1

Introduction

CHAPTER 1: Introduction

Cutting has been cited as a leading cause of non-contact anterior cruciate ligament (ACL) injury in a range of multi directional sports (1,2,3). These injuries have important implications for the sports performer, not only in the short-term through loss of time engaged in sporting activity, but also from the long-term health risks posed by such an injury (4). Therefore, prevention strategies aimed to reduce the risk of injury are a crucial part of planning for the strength and conditioning coach. Sagittal plane biomechanics have an important role to play in the injury mechanism (5,6), and in particular, research has identified flexion angles below 30° during the first 50 milliseconds after initial ground contact as having high risk implications for injury incidence (3,7,8).

Strength has often been reported as having a preventative effect on ACL injury occurrences, often as a part of more holistic training programmes (9,10,11,12,13), however, the specific mechanisms as to how greater strength may be of benefit remain elusive. Data from a number of studies with countermovement jumps (14,15), and cutting (16) have shown that greater strength affects sagittal plane kinematics at the knee, with deeper peak knee angles of ~4-5° reported. A more flexed knee has been shown to reduce ACL strain (17,18) however, it is within the first 50 milliseconds of landing, during the impact phase, where risk of ACL rupture is highest when the knee is in a more extended position (3,7,8). Derrick (51) has reported significant correlations between peak knee angle, and contact angle and Wu et al. (37) showed that peak knee angles, and contact knee angles were significantly deeper in an experienced jumping group, who also happened to be stronger than the non-jump group, although there was no difference in range of motion at the knee. Only one study has investigated strength and knee flexion angle at initial contact in a stretch-shorten cycle movement, and report a significant correlation, suggesting stronger subjects landed in a more flexed position (19), however, more research is needed in this area.

Strength training has also been shown to increase hip contribution during the landing phase of a countermovement task (14) via larger hip extensor moments, which led to a change in the knee to hip ratio. It has been speculated that this change to a more hip dominant landing strategy could serve to alleviate extensor moments at the knee although no study has yet observed this (20,21). Stearns and Powers (14) demonstrated that knee adductor moments could be reduced, and hip moments increased when participants were subjected to only 4 weeks of hip focussed training, involving balance training, and plyometric tasks, however, they did not see significant reductions in knee extensor moments. Perhaps with greater strength changes, the subjects may have seen changes to knee extensor moments as well. Lower knee extensor values would be favourable for reducing ACL injury risk due to reduced pull on the patella tendon that would contributes to anterior tibial translation (22). Currently there is no data to suggest whether hip and knee extensor moments might be altered during higher risk manoeuvres such as cutting.

In addition to the paucity of research into strength and cutting, the current literature fails to adequately address the changing nature of the sporting environment. Many of the current studies tend to use 45° cutting protocols, with a fixed approach speed (16,23,24,25,26,27,28), which does not give an indication of how motor control strategies adapt in response to environmental changes that the performer may face, such as sharper cutting angles, or different velocities into the cut. Having an understanding of how demands change may give insight into injury mechanisms, and how these might interact with performance.

Although approach velocities between studies vary considerably between, 3.5 - 5.8m.s^{-1}, the peak knee angle is consistently measured at between 50 and 60° (16,23,24,25,26,27,28). This value is significant, as it falls within the range reported for optimal torque angle relationship of 51° in a multi joint single leg press (29). Besier et al. (30) observed that peak knee angle during linear running is significantly more extended than a cutting manoeuvre at an equivalent approach velocity demonstrating that the skills are distinct. It could be that peak knee angle is fixed at as deep an angle as possible during this high risk manoeuvre, as deeper knee creates less strain on the ACL (18). Support for this within a different movement pattern is observed by Peng et al. (31), where peak knee angle remained unchanged during incremental single leg drop jumps. In this study it was contact angle that was adapted with increasing drop height to allow greater excursion to dissipate force (32,33). No clear pattern emerges within the cutting data, however, initial contact angle shows much greater variation compared with peak angle, with values ranging from 16°- 42° (23,24). It is possible that peak knee angle is maximised to a position where optimal performance can still occur, and at the same time allow initial contact to be as deep as possible to reduce injury risk at impact. With regards to cutting angle, only 2 studies have reported sagittal plane mechanics. Besier et al. (30) reported a 2.3° increase in peak knee flexion angle at 60° cuts compared to 30° but did not statistically report this difference. Arguably changes of this magnitude would have little functional significance on the knee. On the other hand, when comparing 45° and 90° manoeuvres, Havens and Sigward (34) reported a significant increase of 8°, perhaps indicating that deeper cutting angles have larger impact on knee biomechanics. It should be noted that this study did not control for approach velocity, making results difficult to interpret, therefore more research is required in this area.

There are two main aims within the study, firstly to try to identify how strength may reduce injury risk during cutting tasks. Greater strength may allow stronger subjects to achieve deeper peak knee angles, via having a greater ability to adjust to the larger force requirements as a result of an increasing moment arm at the knee. If force is still absorbed in a similar way during the cut, a similar excursion would allow an increased in knee flexion angle at contact. Therefore the first hypothesis is that the stronger group will have a deeper knee angle at initial contact. Strength also appears to affect hip dominance during landing, as the hip would have the capacity to absorb greater forces during a cut, therefore it is hypothesised that there will be an increased hip extensor moment causing a more hip dominant energy absorption strategy during the touch down phase. The second aim is observe how knee kinematics change in response to varying approach and cut angle conditions. The data suggests that peak knee angle remains similar, perhaps in a position that optimises torque production, in a low risk knee position. It is hypothesised peak angle will remain fixed as approach velocity changes; however, increasing the cutting angle will cause an increase in peak knee angle.

CHAPTER 2

Methods

CHAPTER 2: Methods

2.1 Experimental Approach to the problem

Subjects were asked to take part in two testing sessions separated by at least 72 hrs, and no longer than 14 days. The first testing session involved the assessment of unilateral lower body strength with an isometric squatting task. The subject was then taken to the lab for familiarization of the cutting protocols which would be utilized during the second testing session, and consisted of performing a number of trials at different velocities and angles until the subject felt comfortable with the task. The second testing session measured the participant's knee angle at ground contact, and peak flexion and hip and knee extensor moment, as well as ground reaction force data during 45° and 90° cutting manoeuvre at 3 different approach velocities.

2.2 Subjects

Fourteen physically active subjects with a history of team sports participated in the study. Subjects were recruited due to having a minimum of 5 years previous experience in their respective sports and be training in a multidirectional sport for at least 2 sessions per week. Exclusion criteria for the study included any lower extremity injury that has kept the subject out of training for 3 weeks or more in the 6 months prior to testing, or a previous ACL injury. Subjects were asked to abstain from lower body resistance or vigorous activity for 48hrs prior to each testing session. Only participants who were right leg dominant participated in the study. Strength groups were selected based on their normalized relative peak force values from an isometric strength test. Subjects above the 50th percentile were assigned to the strong group (*n*=7), and those below assigned to the weak group (*n*=7). Subject characteristics for each group can be found in Table 1. The study was approved by the human research ethics committee at St

Mary's University, as well as internal review board at the Qatar anti-doping lab. Subjects were informed of the benefits and risks of the investigation prior to signing an institutionally approved informed consent document to participate in the study.

Table 1: Mean (standard deviation) characteristics of subjects by strength

	Strong (n=7)	Weak (n=7)	*p*
Strength (N/kgbw)	28.6 (3.9)	17.5 (3.7)	0.001
Age (years)	34 (3.8)	34 (5.6)	0.83
Height (cm)	175 (6.3)	178 (8.7)	0.51
Body Mass (kg)	76 (7.6)	78 (15)	0.71

2.3 Procedures

2.3.1 Isometric Strength Testing

Lower body strength was measured in the dominant limb using a single leg isometric squat, performed on a portable 0.6m x 0.4m tri axial force plate (Kistler, Winterthur, Switzerland) with a sampling rate of 1000Hz. The squat was performed using a custom made, floor bolted squat rack with height adjustable bar supports (Figure 1). The subject warmed up by performing 10 repetitions of a half squat with a 20kg barbell, followed by 2 sets at a weight they considered to be equal to 8-10 maximal repetitions (RM) for them. They were then moved to a Smith machine and asked to perform quarter squats with progressive intensity. One set at 6RM, one at 4 RM, and one at 2RM followed by 2 familiarization trials at ~90% maximal exertion. Knee and hip angles of 40° were selected to ascertain peak force in the lower limb (35). Measurement of knee and hip angle were taken with a goniometer and joint centres were established by methods described in table 2. The angle recorded is relative to the position of the proximal to distal segment i.e 40° knee angle is measured as the angle of the shank in relation to the thigh. Subjects were asked to place the heel of the dominant leg underneath the bar, with the hips back and apply as much force as possible upwards for 5 seconds on 3 occasions with 2 minutes recovery between each trial (36). Peak force was selected as the highest force achieved, but only if the second best trial was within 10% of the highest, if not a further trial was recorded. The subjects were then given a 10 minute rest during which anthropometric measurements were taken. Subjects then undertook a number of familiarization trials until they felt comfortable with the technique required.

Figure 1: Adopted subject positioning for the unilateral isometric strength test. Squat rack was bolted to the floor, and had custom made stoppers connecting the barbell to the rack.

2.3.2 Instrumentation

To record three dimensional, lower extremity kinematics during the cutting manoeuvres, a 16 camera motion system (Vicon MX, UK) was used with a sampling frequency of 250Hz. Sixteen 14mm hard markers encased in retro-reflective tape were attached to anatomical landmarks of the lower limb (Figure 2) in accordance with the Vicon clinical manager marker set (Table 2). Anthropometric measurements for leg length were taken from anterior superior iliac spine to medial malleoli, as well as and ankle and knee girth. These measurements together fully define the reference coordinate system for a three dimensional lower body coordinate system. The subject was asked to stand in the anatomical position in the centre of the force plate as a standing trial prior to data collection, to determine segment lengths. Ground reaction force data during the cutting task was collected using a 0.6 x 0.9m force plate (Type 2812A, Kistler, Winterthur, Switzerland) embedded into the floor and sampled at a frequency of 1000Hz. Cameras were synchronised to the force platforms so that joint moments could be calculated.

Table 2: Definitions of landmarks for identification of axis of rotation and segment coordinate frame

Joint Centre or Segment	Marker Site	Description
Hip	L/R Greater Trochanter	Axis of rotation at the hip. Palpate for bony prominence inferior to iliac crest.
Knee	L/R Lateral Femoral Condyle	Axis of rotation at the knee. Palpate under patella and around to feel joint line. Palpate superiorly to find bony prominence just above joint line.
Ankle	L/R Lateral Malleoli	Axis of rotation for ankle.
Pelvis	L/R ASIS L/R PSIS	Most prominent point of left or right ASIS, and left and right PSIS. Defines segment frame
Thigh	L/R thigh on lateral aspect	Midpoint between marker on lateral femoral condyle, and greater trochanter (higher on right side). To define segment length.
Shank	L/R shank on lateral aspect	Midpoint of shin in line with lateral malleolus and lateral femoral condyle (higher on the right). Define segment length.
Foot	L/R 2nd Metatarsal L/R Heel	Heel level with 2nd metatarsal placement and in line with achillies to define segment length

Figure 2: Subject positioned for standing calibration trial with retro-reflective marker placed at relevant sites (heel and PSIS are posteriorly positioned). This is to establish local coordinate system relative to global that is marked via static and dynamic calibration with the wand.

2.3.3 Change of Direction Trials

On the second day of testing, subjects reported to the biomechanics lab where they were given lycra shorts to wear and reflective markers were placed on their lower body (Figure 2). Subjects then performed a 10 minute standardized warm up including a short familiarisation at the various cutting protocols. For the trials subjects were asked to start on a marked line 15m from the centre of the force plate. To test the variations in lower body kinematics due to variations in the task, 3 different velocities (2, 4, and 6m.s^{-1}) were selected to be performed at cutting angles of 45° and 90°. The subject was asked to perform 3 trials for each of the 6 conditions, making a total of 18 manoeuvres in the session, however, if the subject failed to achieve an approach velocity within ± 5% of the target, or they were perceived to be targeting the force plate, they were asked to repeat the trial. To negate the effects of fatigue a minimum of 1 minute recovery was given between each trial. To avoid potentially harmful and unnatural cutting movement, approach velocity was measured for 3m, 8m from the force plate using light gates (Microgate Polifemo Light, Bolzano Bozen, Italy), this allowed greater realism to the task whilst providing a consistent measure for approach velocity. Subjects were asked to try to maintain their velocity through the second set of light gates and perform the cut as quickly as possible, however, they were told it was acceptable to slow down during the second approach phase in order to cut safely and allowed the movement to be pre-planned. A third and fourth light gate measured the completion time for the task but were mainly used to ensure the athlete was attempting maximal performance (Figure 3). To ensure the correct cutting angle was achieved, tape marking was applied to the floor to guide the athletes with the fourth set of photocells set up 2m from the force plate and spaced 50cm either side of the marker tape to ensure the actual cut angle for the 45° and 90° trials would be between 35° and 55°, and 80° to 100° respectively. Subjects were asked to complete the movement as fast as possible.

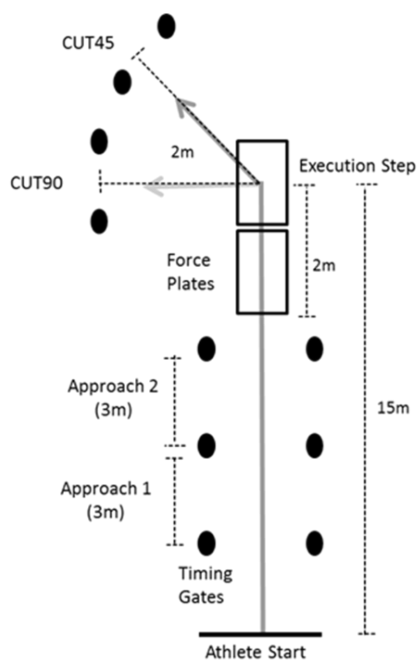

Figure 3: Experimental setup for the cutting task. Solid line arrow represents line of motion with two cutting angles at the end. The light was timer was triggered from 8m before force plate, and again 5m before the force plate.

2.3.4 Data Analysis

The force platform was used to determine heel strike events to define the stance phase of the cutting (Figure 4). Within this time frame, from initial contact to the first force peak, impact knee (KEM_1) and hip (HEM) extensor moment were recorded. Between the first force peak and when ground reaction force data reached its second peak, the highest value for knee extensor moment was reported (KEM_2). In addition, contact time, peak braking force (GRF_{br}), and ground reaction force at the first (GRF_1) and second peak (GRF_2) were also extracted. Initial contact knee angle (KA_{ic}) was reported as the knee angle at foot strike, determined as the point at which ground reaction force exceeds 15N. Peak knee angle (KA_{pe}) was taken as the deepest knee angle based on the kinematic data. Relative contribution of the knee and hip during initial contact was calculated by dividing the KEM_1 by HEM to get a knee to hip ratio. Scores larger than 1 indicate greater influence from the knee, whereas, scores below 1 indicate greater influence from the hip (20,21). To calculate joint kinetics inverse dynamics were utilised using kinematic data derived from the motion analysis, and force data and centre of pressure derived from the force plate. YXZ Cardan angles were compared using relative orientation of 2 segments, using data from previous studies (48,49). Newton Eular equations of motion were applied at the beginning of one end of the segment starting at the ground and calculating each segment up to the hip. The 'Plug-in gate' model in Vicon was used to calculate joint kinetics. Positive angles represent flexion at the hip and knee. Data was filtered using a Woltring filter quintic spline routine in mean square error mode with a smoothing factor of 10.

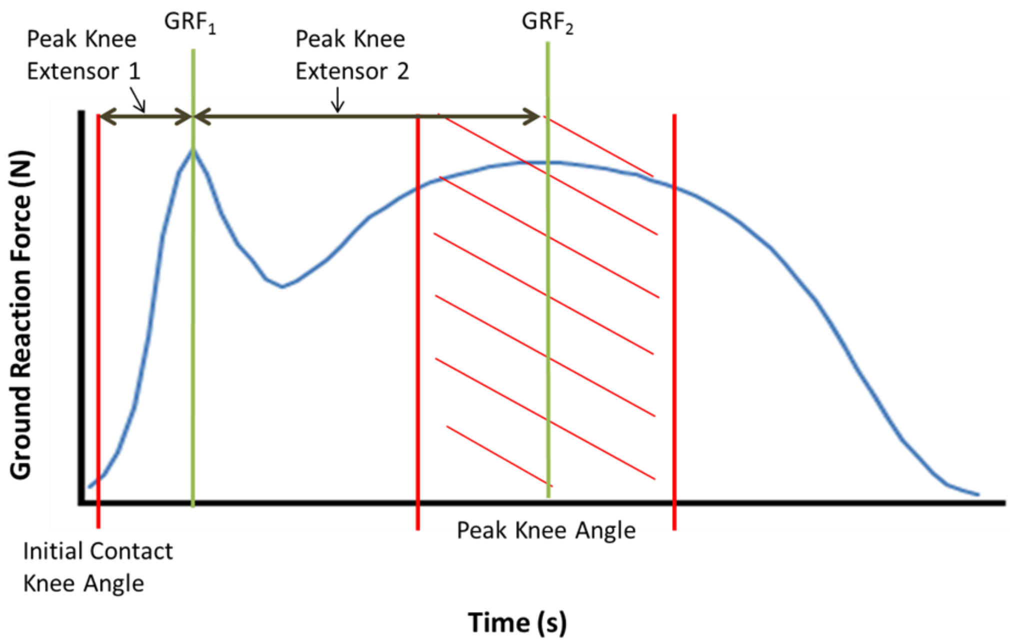

Figure 4: Diagram to show at which time the key variables were recorded in relation to the stance phase.

2.4 Statistical Analysis

A 3 way mixed design ANOVA was used with 1 between group condition (Strength) and 2 within group conditions (Cut Angle*Velocity). Statistical significance was established at an Alpha level (α) < 0.05 with all data reported as mean and standard deviation (sd). Sidak post hoc analysis was utilised to observe where differences existed between different conditions. Partial Eta squared values (n_p^2) were reported as a measure of effect size (48), where 0.01, 0.06 and 0.14 represent small, moderate, and large effects respectively. All data analysis was completed using a statistical software package (SPSS, Version 22).

CHAPTER 3

Results

CHAPTER 3: Results

The strong group had a significantly higher relative peak mean strength of 28.6 (3.9) N/kg^{-1}, compared to the weak groups 17.5 (3.7) N/Kg^{-1} ($p < 0.001$). Subject characteristics were not significantly different between strength groups.

In the final 3 meters of the approach, subjects slowed their approach run up significantly in both the 45° and 90° cutting conditions during the $6m.s^{-1}$ trial, whereas at 2 and $4m.s^{-1}$ approach velocity at this time point was still the same as the target velocity. In the 90° cut the subjects slowed to a greater extent compared to the 45° cut (5.26 vs 5.61 $m.s^{-1}$).

3.1 Strength Group Comparisons

Knee and hip extensor moment comparisons between the strong and the weak group are presented in figure 5. Stronger subjects demonstrated significantly larger peak hip extensor moments, $F(1,12) = 9.34$, $p = 0.01$, and lower knee extensor moments, $F(1,12) = 9.82$, $p = 0.009$, during the impact phase of the manoeuvre. This significantly reduced the knee to hip ratio, $F(1,12) = 11.58$, $p = 0.005$, in the stronger group. Peak knee extensor moment was lower in the strong group and approached significance, $F(1,12) = 4.30$, $p = 0.06$, $n_p^2 = 0.264$. Figure 6 presents data for knee angle comparison between strength groups. Knee flexion angle at initial contact was significantly deeper, $F(1,12) = 4.75$, $p = 0.05$, $n_p^2 = 0.28$, for the strong group. Peak flexion angle was also deeper in the strong group to a similar magnitude, however this was not significant, but did show a large effect size (66.7° vs 61°, $p = 0.16$, $n_p^2 = 0.16$). No other differences between group variables were observed.

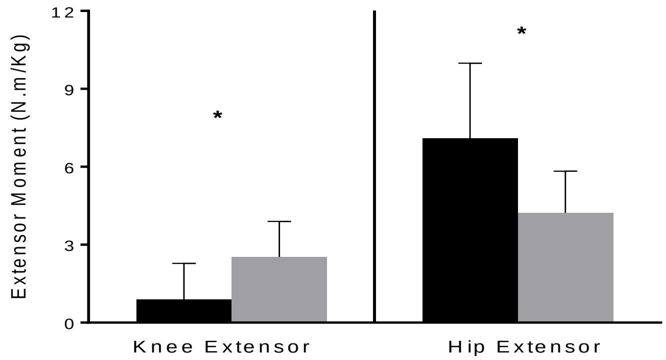

Figure 5: Demonstrates the overall mean difference of hip and knee extensor moment between the strong (black) and weak (grey) group during the cutting task. *indicates significant difference between groups (p<0.05).

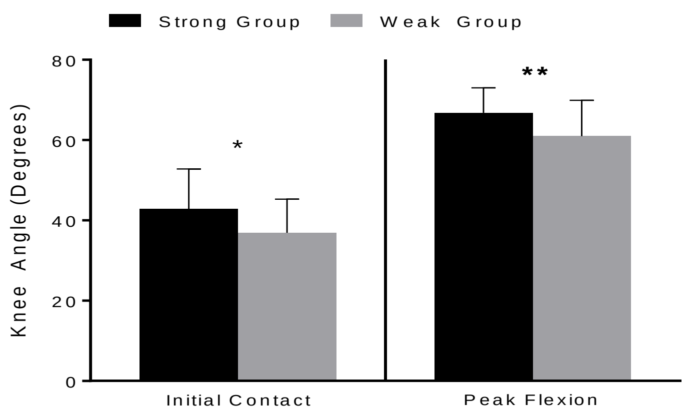

Figure 6: Demonstrates the overall mean difference of knee angle at initial contact, and at peak flexion between the strong and the weak group during the cutting task. *indicates significant difference between groups (p≤0.05). **represents large effect size between groups (ϵ>0.14).

3.2 Cutting Conditions

3.2.1 Kinematic Variables

An interaction between angle and velocity $F(2,24) = 57.49$, p = 0.001 for initial contact knee flexion angle was observed and is presented in figure 7a. Peak knee angle also observed an interaction between angle and velocity $F(2,24) = 5.92$, p = 0.008 (Figure 7b). An interaction between cut angle x velocity also existed, $F(2,24) = 12.34$, p = 0.001. There was a significant main effect for cut angle ($p = 0.001$), but this was only seen at 2 $m.s^{-1}$. Velocity demonstrated a significant main effect where 2 $m.s^{-1}$ was different from 4 and 6 $m.s^{-1}$ ($p < 0.001$).

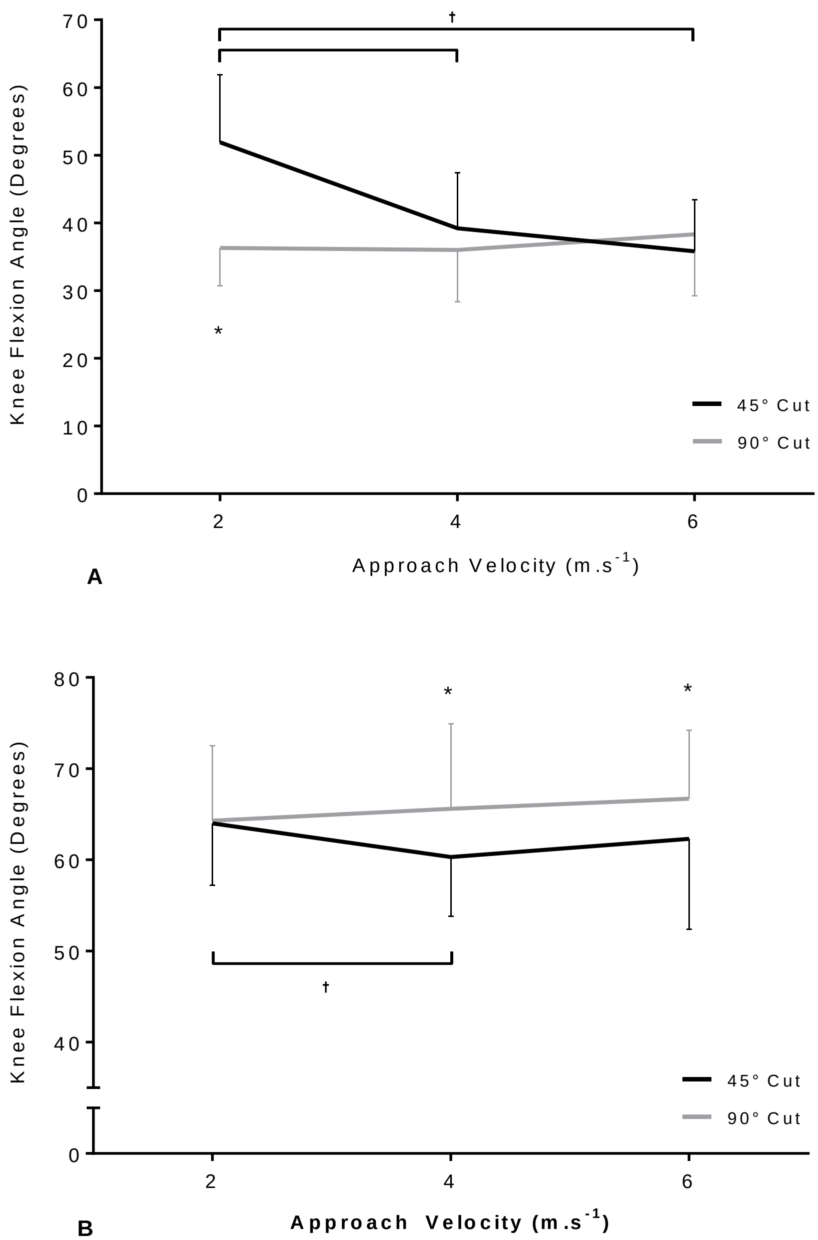

Figure 7: Collapsed group data for knee flexion angle for the 2 different cutting angles. Graph A represent initial contact and graph B represents peak knee angle. * denotes significant difference in knee flexion angle between 45 and 90 degree cut ($p<0.05$), † denotes knee angle values at initial contact were significantly different from 2$m.s^{-1}$ condition only ($p<0.05$).

3.2.2 Kinetic Variables

Data for the kinetic variables can be found in Table 3. Data recorded during the first ground reaction force peak demonstrates that both knee extensor moment, $F(2,24) = 6.93$, $p = 0.004$, and resultant ground reaction force (GRF_1), $F(2,24) = 18.77$, $p = 0.001$, show a significant interaction between cut angle and velocity. Data recorded during the second peak shows that knee extensor moment demonstrated an interaction between cut angle and velocity, $F(2,24) = 42.39$, $p = 0.001$. Ground reaction force at the second peak (GRF_2) also demonstrated an interaction between angle and velocity, $F(2,24) = 10.85$, $p = 0.001$.

Posterior ground reaction forces (GRF_{br}) displayed a velocity x cut angle interaction, $F(2,24) = 9.71$, $p = 0.001$). GRF_{br} became significantly greater with increasing velocity in the 45° cut ($p < 0.01$) whereas it did not change between approach velocities in the 90° cut. Between cut angle conditions GRF_{br} was significantly greater in the 90° cut at 2 $m.s^{-1}$ ($p < 0.001$) and 4 $m.s^{-1}$ ($p < 0.05$), whereas at 6 $m.s^{-1}$ braking was the same in both 45° and 90° manoeuvres. Contact time also showed an interaction between velocity and cut angle, $F(2,24) = 3.44$, $p = 0.049$. In the 45° condition contact time was longest at 2 $m.s^{-1}$ compared to 4 $m.s^{-1}$ ($p < 0.001$) and 6 $m.s^{-1}$ ($p \leq 0.01$). In the 90° condition, contact time remained the same at all 3 approach velocities. Between cutting angles, contact times were longer at all approach velocities in the 90° cut compared to the 45° cut condition ($p < 0.001$).

Table 3: Collapsed group data for kinetics of cutting manoeuvres.

	CUT45			CUT90			Main Effects & Interactions	*p*-value	Effect Size
	V2	V4	V6	V2	V4	V6			
Extensor Moment (Nm/kg)									
KEM$_1$	1.19 (1.09)	1.68 (1.11)	3.03 (2.14)†	1.34 (1.25)	1.58 (1.66)	1.48 (1.57)[a]	Group Velocity (2<6) Angle*velocity	=0.009 <0.001 =0.011	0.450 0.658 0.366
KEM$_2$	2.87 (0.77)	3.60 (1.09)*	5.02 (1.00)†	3.34 (0.92)[a]	3.74 (1.03)*	3.40 (0.70)[a]	Velocity (2<4<6) Angle Angle*Velocity	≤0.006 =0.19 <0.001	0.869 0.377 0.779
Ground Reaction Force (N/kg)									
GRF$_1$	205 (46)	299 (67)*	352 (83)*	283 (46)[a]	287 (75)	273 (59)[a]	Velocity (2<4,6) Angle*Velocity	≤0.005 <0.001	0.600 0.610
GRF$_2$	249 (43)	265 (37)*	261 (34)	248 (42)	244 (39)[a]	229 (30)[a]†	Angle Angle*Velocity	=0.001 <0.001	0.617 0.475

Note: Results are mean (sd). CUT45 = 45 ° cut condition, CUT90 = 90 ° cut condition, V2 = 2m.s^{-1} approach velocity, V4 = 4m.s^{-1} approach velocity , V6 = 6m.s^{-1} approach velocity, IC = Initial Contact, s = seconds

*indicates a statistically significant difference from V2 condition (p<0.05)

†indicates a statistically significant difference from V2 and V4 conditions (p<0.05)

[a] indicates a statistically significant difference from CUT45 (p<0.05)

CHAPTER 4

Discussion

CHAPTER 4: Discussion

4.1 Strength

The first aim of the study was to identify whether stronger individuals present different lower extremity mechanics during a cutting manoeuvre in their plant leg. The first hypothesis suggested the strong group would exhibit increases in knee flexion angle during the initial contact phase of stance during cutting. The results show that this hypothesis was correct, with an overall increase in knee flexion of 6.7° in the strong group compared to the weaker group. Deeper contact angles have been associated with strength in single leg jump task (19) and Wu et al. (37) observed significantly increased knee flexion at ground contact in a jump group vs non jump group, during bilateral landing tasks in which the jump group were significantly stronger. In terms of peak knee angle, Wu et al. (37) also reported increased flexion between groups, similar to Spiteri et al. (16) significant 5° increases during a 45° cutting protocol, and Stearns and Powers (14) significant 4° increases during bilateral drop jump after a 4 week plyometric training programme. The current study did not see significant changes, but the magnitude of change of 5.7° was similar to the literature and in line with the increases in knee flexion at initial contact, and carried a large effect size. The data may support a premise that stronger individuals are able to select a movement strategy that has both a deeper peak and initial contact angle. A potential mechanism that would enable a stronger athlete to achieve deeper peak knee angles might relate to their greater capacity to adjust to an increasing moment arm length. As the knee flexes, the moment arm between the axis of rotation at the knee, and the line of action of the performers mass will increase. An increase in moment arm will increase the force that is required by the muscle to maintain a deeper knee posture. This is of particular importance at peak flexion when the knee must be in a position from which it is able to extend quickly to perform the manoeuvre efficiently. Greater strength would be able to offset the greater internal force requirements imposed by the lengthening moment arm. A deeper peak knee angle would allow for a deeper initial contact angle, whilst maintaining the same displacement. In this respect, stronger athletes can maintain the performance of the skill, whilst at the same time making it safer by reducing strain on the ACL. In terms of being beneficial for injury prevention, greater knee flexion has been shown to reduce ACL load by decreasing the patella tendon tibial shaft angle, which in turn reduces tibial translation in relation to the femur (18). In addition, ACL elevation angle is decreased with greater knee flexion, causing the ACL to have less tensile strain placed upon it at any given shear force (17). This would mean that as the foot contacts the floor at the start of the plant phase, a stronger individual would have an ACL that is under lower strain which may help it to tolerate the high impact forces to a greater extent. The weaker group are still able to manage the forces that are experienced during the contact phase in a similar fashion to the strong group, as knee excursion remains similar between the groups, it is just that the knee is in a less favourable position with regards to ACL load.

The second hypothesis predicted that the strong group would have a lower knee to hip ratio than the weak group implying a greater contribution from the hip during the task. This was proved to be the case also, with a 0.55 lower knee to hip ratio in the strong group. The evidence that suggested this would arise via increased hip contribution to the movement pattern during cutting was only partly correct. In fact, the lower ratio was not only created by a significant increase in hip extensor moment, but importantly, that the stronger group also had a significantly lower knee extensor moment, and these differences were observed across all conditions.

In a 4 week, hip focused plyometric and balance training programme, Stearns and Powers (14) reported similar hip dominant changes in the knee to hip extensor ratio. However, when comparing the absolute values for hip and knee extensor moment, the 4 week intervention saw increases in hip extensor moment, but no change in knee extensor moment. However, knee adductor moments were reduced which the authors' state may mitigate injury risk in women. The current study may add to the literature as it suggests strength can alter sagittal plane knee kinetics as well by directly reducing extensor moment. This distinction is an important one as extensor moments at the knee are associated with proximal anterior shear force (22) due to the quadriceps loading the patella tendon during deceleration. These anterior forces increase the load on the ACL (38) and thus place the ligament under greater strain in the sagittal plane.

The lack of change reported in knee extensor moment within Stearns and Powers (14) work may be partly due to differences in how extensor moment was recorded. Measurement was taken as a mean for the whole of the deceleration phase, rather than just a peak value during the impact phase only, as was the case in the present study. The present study shows that values for knee extensor during the push off phase were higher than those reported at impact (Table 3), and so using a similar approach to Stearns and Powers (14) may have resulted in similar findings. Another reason for the differences may be in relation to the magnitude of strength differences between pre to post in the training study (7%), versus the strong and weak group difference of 63% in the present study. Overall, this may imply that as the individual becomes stronger, the greater loading can be transferred to the hip, but as strength continues to increase, the hip can absorb so much of the force that lower contributions can be made at the knee, and ACL strain in the sagittal plane can be moderated.

4.2 Cutting Conditions

The second part of the study aimed to establish how different conditions such as cutting angle and approach velocity into the cut would affect lower body biomechanics, specifically with regard to knee flexion angles. The first hypothesis was accurate in predicting there would be no difference in peak knee angle as a result of changing approach velocity. When cut angle data was collapsed, values were recorded at 64.1°, 63, and 64.6° at 2, 4 and 6m.s^{-1} respectively. These values are higher than those reported in the literature, however by splitting the current study's data into 45° and 90° cutting tasks more comparable figures emerge. A number of studies used approach velocities between 4 and 5m.s^{-1} with a 45° cut, and peak knee angle in these were reported within a range of 55- 63° (16,23,25,24,26). In the equivalent conditions in the present study, values of 60.3° were observed and are comparable with that of the current literature. However, in general, the data values in the present study were higher than those seen in the other literature. It is speculated that subjects in the present study may have been stronger than in the papers, and were certainly much stronger than in Spiteri et al (16) study who used a similar strength protocol. The strongest subjects in that study produced similar force to the weakest in the present study at 20.86 vs 17.5 N/Kg^{-1} respectively. This may go some way to explain the deeper differences.

The second hypothesis was also correct, peak knee angle was significantly higher in the 90° cut condition with a 3.3° increase compared to the 45° task. Whether these increases would have any functional significance is hard to tell, but it is worth noting that at 4 and 6 m.s^{-1} there were increases of 5.3° and 4.4° respectively, which may have a more practical relevance. This supports the data of Havens and Sigward (34) however their data showed a greater difference between the two cuts with a value of 8° separating the 90° and 45° cuts. The difference may be due to a difference in the approach, with subjects in that study only being given a 7.5m run up and were asked to perform the task as quickly as possible. Subjects may still have been accelerating by the time they reached the force platform which may change the mechanics of the movement. Additionally, approach speed was not controlled for in the study, and although the present study generally shows velocity to have no effect on peak knee angle, further analysis of the current studies data displays a significant difference between 2 $m.s^{-1}$ and 4 $m.s^{-1}$ in the 45° cut. It is possible that this lack of control confounded the data of Havens and Sigward (34) to some degree. Another study by Besier et al. (30) saw small increases of 2.3° in a 60° compared to 30° cut although significance was not reported, but may support a trend for increasing peak knee angle as cut angle becomes more obtuse.

The movement strategy adopted appears to try to find a balance between minimising injury risk, whilst still trying to optimise performance. As mentioned, deeper peak knee angles can reduce ACL load significantly, at a time in the movement when ground reaction forces, and extensor moments are at their highest (Table 3). Studies show that tibial displacement is reduced until at approximately 60° knee flexion, when it reaches 0mm, at which point the ACL in under very low strain (40,41). It would appear that the movement strategy selected during cutting in all conditions in the present study, enable the knee to reach this flexion angle to reduce ACL strain. Greater levels of flexion continue to reduce ACL strain up to approximately 80° (42,43) however, the deepest average knee angle in the present study occurred at 66.8°. This might suggest a balance is struck where knee angle is deemed safe, and any further flexion would impair performance.

On this note, one of the contributing factors to effective performance during a cut would be reaching optimal torque values in the quadriceps to help the athlete extend the knee efficiently during the propulsive phase. The average peak values of around 63.9° in this study are in the vicinity of the values reported for the optimal peak torques- angle relationship. Using a isometric single leg press, peak torque was shown to occur at a 51°, with a highest individual value of 62°, beyond which torque output diminishes at a steady rate, either as the length of the moment arm increases, or the force length relationship is altered negatively (29,50). Other studies have suggested the peak torque angle to be higher at between 65-70° (42,43) however the knee extensor protocol used has arguably less correspondence to the cutting manoeuvre. Taking the injury and performance data together, it would appear that injury prevention mechanisms dictate the depth of peak knee angle during cutting, but that when the movement is pre-planned and approach speed can be moderated, a depth is selected where performance can still be maximised. The disparity between the values of Hahn et al. (29) and the present study might be explained by the dynamic nature of the movement during cutting, compared to the static conditions in an isometric press. The elastic component of the passive tissues may allow a slightly deeper peak angle, which would allow the knee to return to a more efficient position supported by the stretch shorten cycle (52).

Another consideration, other than achieving a safe peak knee angle, could be related to initial contact angle, as this is the time point during the stance phase when the knee is in a more extended position and injury risk is at its highest. From an ACL injury perspective, data from observational studies (3,7), and stochastic modelling (8) report contact angles of below 30° as having high injury risk. In addition, more extended contact angles below 20° also create conditions where the hamstring is unable to apply any substantial posterior force to the tibia and would prevent it from stabilising the knee if extensor force became too large (40). A deeper peak knee angle may allow for a more flexed knee when contact with the ground is made. Initial contact knee flexion values in the present study showed no significant differences across any of the conditions except at 2m.s^{-1} during the 45° cut (51.9°), where it was significantly deeper than any other velocity condition, or equivalent velocity at 90°. All other initial contact values ranged between 35.8° and 39.2° across the 5 other conditions. These values are noteworthy in that they all appear to be markedly above the 30° value that has been reported to be high risk for incurring an ACL injury. Initial contact knee angle values above 30° have also been observed in other similar protocols involving pre-planned manoeuvres (23,30). This might suggest that deeper knee angles at peak flexion are selected to ensure a safer knee angle can be achieved at contact. A key consideration here is the level of excursion that is required for a given movement. Greater excursions at the knee, and therefore, less stiff landings, have been shown to reduce extensor moments and ground reaction forces (32,33). It could be argued that a peak knee angle is set which allows the individual enough excursion to maintain knee extensor moments and ground reaction forces within safe limits during the task, but also ensure that initial contact knee angle safely exceeds the 30° high risk value. Although deeper knee positions might be safer, a balance must found in which performance is compromised as little as possible.

When looking at the 45° condition in isolation, peak knee angles showed small differences, however, initial contact angle was significantly more flexed at the 2 m.s^{-1} velocities, with 4 and 6 m.s^{-1} showing no difference from one another. This demonstrates that under conditions where task demands are low, it is initial contact angle that is altered to reduce the range of motion at the knee, and not peak knee angle. The reduced excursion results in a stiffer knee at ground contact which has been shown to increase knee extensor moments and ground reaction forces in landing tasks (50,9). This movement pattern is utilised to try to increase the rate of loading, and as a result, the impulse required to complete the task can be generated in a shorter time period reducing time spent in contact with the floor. In addition, greater storage and return of elastic energy can be achieved which can aid performance during the propulsive phase (47).

As velocity is increased in the 45° cut, initial contact angle becomes more extended to increase the excursion range and help to moderate extensor moments. It appears a minimum limit for initial contact exists as there no difference was observed for this variable between 4 and 6 $m.s^{-1}$. With no change in initial contact or peak knee angle it seems a fixed bandwidth for excursion exists at faster velocities, and the subjects moderates approach velocity to ensure they can perform safely within these pre-set limits, evidenced by a reduction in speed in the 6 $m.s^{-1}$ condition at the second approach to 5.6$m.s^{-1}$. This may suggest subjects moderated their speed leading into the cut to ensure they could fall between the bandwidths identified so that the demands of the tasks can be met safely and without losing efficiency. Greater speed may still have been carried into the manoeuvre at 6 $m.s^{-1}$ compared to 4$m.s^{-1}$, but with similar knee excursions, and larger braking requirements, initial and peak knee extensor moment were increased significantly as a result, and may imply the maximum safe limits that the structure of the knee can endure at this specific cut angle.

As reported, 90° cutting task saw greater peak knee angles, yet the same initial contact angles compared to the 45° condition, meaning excursions were greater at 90°. Cutting tasks at 90° require larger braking forces as they subjects must come to a complete stop before redirection. To help mitigate these forces, higher joint excursions were utilised to help absorb these forces at impact and help to moderate extensor moments (32,33). However, when comparing the cutting angles at 6 $m.s^{-1}$, the 45° cut generated the same braking forces, with higher GRF_2, and much lower excursion than at 90°. This strategy resulted in significantly higher peak, and initial contact, knee extensor moments in the 45° condition. A reason knee extensor moment remained lower in the 90° cut, compared to 45° may be due to the reduction in approach velocity observed at 90°, as well as an increased excursion, but why this strategy is utilised to keep extensor moment lower is less clear. Markolf et al. (44) observed that although ACL is the primary stabiliser for anterior sheer forces, the addition of valgus and rotational moments at the knee place the ACL under greater strain. Greater cutting angles have been shown in the literature to cause greater frontal and transverse plane stresses applied at the knee (30,34,46) which when combined with extensor forces create a combined stress load on the ACL. The present study demonstrates that the subjects tried to maintain, and in 6 $m.s^{-1}$, even reduce sagittal plane extensor forces to a greater extent in the 90° condition, whereas at 45°, where less rotation and valgus movements would be experienced, extensor forces could be higher whilst applying a similar overall strain to the ACL and may result in continued performance benefits in this condition as contact times can be reduced.

The increase in peak knee angle in the 90° as opposed to selecting a more extended leg during the contact phase, may suggest a strategy that is driven more by an injury prevention mechanism at the expense of performance as task demands are increased. Selecting a straighter knee at initial contact would place the knee closer to a high risk position around 30° of flexion, thereby increasing ACL strain. By increasing peak flexion, ACL load remains lower at contact, but the deeper knee angles observed in the 90° condition would put the knee in a more unfavourable position for performance, by increasing the distance of the active mass of the subject, to the axis of rotation at the knee. This increase to the moment arm would increase the force required by the quadriceps and, alongside the reduced stiffness of the knee, may explain the significantly increased contact times at all approach velocities in the 90° condition (0.23 vs 0.32s). The greater contact time would allow greater opportunity for cross bridge formation to allow the muscle to overcome the inertia created by the deep knee position, and reduced elastic return of energy.

Further evidence of an injury prevention mechanism at work in more demanding conditions relates to changes that occur at $6m.s^{-1}$ compared to 2 and $4m.s^{-1}$ in the 90° cut condition. A reduction in peak knee extensor moment, and GRF_2 at observed which may indicate that in this high demand condition, the subject was unable to control velocity into the manoeuvre as effectively, resulting in forces that may pose risk to the integrity of the joint structures. Excessive forces are known to stimulate Golgi tendon organ into inhibiting efferent output to protect passive structures (40), and may impact performance efficiency as evidenced by lower GRF_2. In a comparable study using incremental single leg drop jumps Peng et al. (31) reported reductions in performance in the final 60cm condition, which coincided with an increased peak knee angle, suggesting when velocity into contact phase cannot be controlled, peak knee angle is affected. It is worth noting initial contact angles became more extended at this time point also, however, did not go below 30° as seen in the present study. This was not seen in the present study, possibly because the subject was able to moderate their velocity to ensure successful and safe execution of the task, whereas in a drop jump this is not possible. This may further support the notion that knee angles are primarily selected with injury prevention mechanisms as a priority, rather than performance.

In conclusion, neuromuscular programmes have demonstrated a reduction in ACL injury rates. The present study demonstrates the benefits that including strength may have in directly reducing ACL strain in dynamic movements such as cutting. Strength may have a protective effect on ACL by altering load distribution to the hip and alleviating stresses on the knee via reduced extensor moment, during initial contact with the ground. Additionally, strength may have a favourable influence on knee angle at contact. Movement strategies adopted during the deceleration phase of the manoeuvre, at greater cutting angles, appear to select deeper peak knee angles. These changes allow larger excursions for the absorption of higher forces that result from the greater braking requirements at 90°. As the demand of the cutting task increases approach velocity is moderated to ensure a balance can still be met between successful execution of the task, whilst maintaining a safe knee posture. It would appear that even in higher demand activities, a flexion angle below 30° is avoided to reduce ACL loading suggesting a movement strategy that is driven by an injury avoidance mechanism. In lower demand movements, knee excursion is reduced to help increase the rate of loading that would aid optimal task performance, unfortunately a limitation of the study is that post stride velocity was not measured which may have demonstrated this. Taken as a whole, the data may suggest that a bandwidth exists for a given cutting angle, where, in a pre-planned movement where approach velocity can be modulated, initial contact, and peak knee angles have set limits to ensure efficient and safe execution of a task.

CHAPTER 5

Practical Recommendations

CHAPTER 5: Practical Recommendations

Data from this study confirms the importance of strength as part of a neuromuscular injury prevention programme. The large strength differences between groups in the present study may indicate greater strength gains may be required to see the magnitude of changes during cutting, therefore a well-planned long term training programme may be necessary to see the magnitude of effects that were observed in the present study. Hip focussed strength programmes would help to increase the likelihood of transfer to hip moments during performance. Greater quadriceps strength may also be beneficial to enable the performer to attain deeper knee positions, without affecting performance negatively. It might therefore be efficient to use compound exercises to achieve this goal.

CHAPTER 6

References

CHAPTER 6: References

1. Olsen, O-E, Myklebust, G, Engebretsen, L, and Bahr, R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med*. 32(4): 1002-1012, 2004.
2. Kimura, Y, Ishibashi, Y, Tsuda, E, Yamamoto, Y, Tsukada, H, and Toh, S. Mechanisms for anterior cruciate ligament injuroes in badminton. *Br J Sports Med*. 44: 1124-1127, 2010.
3. Koga, H, Nakamae, A, Shima, Y, Iwasa, J, Myklebust, G, Engebretsen, L, Bahr, R, and Krosshaug, T. Mechanisms for non contact cruciate ligament injuries. *Am J Sports Med*. 38(11): 2218-2226, 2010.
4. Myklebust, G, and Bahr, R. Return to play guidelines after anterior cruciate ligament surgery. *Br J Sports Med*. 39(3): 127-131, 2005.
5. DeMorat, G, Weinhold, P, Blackburn, T, Chudik, S, & Garrett, W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med*. 32(2): 477-483, 2004.
6. Yu, B, and Garrett, WE. Mechanisms of non-contact ACL injuries. *Brit J Sports Med*. 41 (supp1): i47-i51, 2007. Doi: 10.1136/bjsm.2007.037192
7. Krosshaug, T, Nakamae, A, Boden, BP, Engebretsen, L, Smith, G, Slaughterbeck, JR, Hewett, TE, and Bahr, R. Mechanisms of anterior cruciate ligament injury in basketball. *Am J Sports Med*. 35(3): 359-367, 2007.
8. Lin, C-F, Liu, H, Gros, MT, Weinhold, P, Garrett, WE, and Yu, B. Biomechanical risk factors of non-contact injuries: a stochastic biomechanical modeling study. *J Sports Health Sci*. 16-42, 2012doi:10.1016/j.jshs.2012.01.001

9. Cahill, BR, and Griffith, EH. Effect of preseason conditioning on the incidence and severity of high school football knee injuries. *Am J Sports Med*. 6(4): 180-184, 1978.

10. Heidt, RS, Sweeterman, LM, Carlonas, RL, Traub, JA, & Tekulve, FX. Avoidance of soccer injuries with pre season conditioning. *Am J Sports Med*. 28: 659-662, 2000.

11. Myer, GD, Ford, KR, Palumbo, JP, and Hewett, TE. Neuromuscular training improves performance and lower-extremity biomechanics in female athletes. *J Strength Cond Res*. 19(1): 51-60, 2005.

12. Hewett, TE, Lindenfield, TN, Riccobene, V, & Noyes, FR. The effect of neuromuscular training on the incidence of knee injury in female athletes: a prospective study. *Am J Sports Med*. 27: 699-706, 1999.

13. Mandelbaum, BR, Silvers, HJ, Watanabe, JF, Thomas, SD, Griffin, LH, Kirkendell, DT, and Garrett, W. Effectiveness of a neuromuscular and proprioceptive training program in preventing anterior cruciate ligament injuries in female athletes: 2-year follow- up. *Am J Sports Med*. 33(7): 1003-1010, 2005.

14. Stearns, KM, and Powers, CM. Improvements in hip muscle performance result in increased use of the hip extensors during a landing task. *Am J Sports Med*. 42: 602-609, 2014.

15. McCurdy, K, Walker, J, Saxe, J, and Woods, J. The effect of short-term resistance training on hip and knee kinematics during vertical drop jumps. *J Strength Cond Res*. 26(5): 1257-1264, 2012.

16. Spiteri, T, Cochrane, JL, Hart, NH, Haff, GG, and Nimphius, S. Effect of strength on plant foot kinetics and kinematics during a change of direction task. *Eur J Sport Sci.* 13(6): 646-652, 2013.

17. Li, G, DeFrate, LE, Rubash, HE, and Gill, TJ. In vivo kinematics of the ACL during weight-bearing knee flexion. *J Orthop Res*. 23: 340-344, 2005.

18. Nunley, RM., Wright, D., Renner, JB., Yu, B. & Garrett, WE. Gender comparison of patellar tendon tibial shaft angle with eight bearing. *Res Sports Med*. 11. 173-185. 2003.

19. Nagai, T, Sell, TC, House, AJ, Abt, JP, and Lephart, SM. Knee proprioception and strength and landing kinematics during a single leg stop jump task. *J Ath Train*. 48(1): 31-38, 2013.

20. Pollard, CD, Sigward, SM, and Powers, CM. Limited hip and knee flexion during landing isd associated with increased frontal plane knee motion and moments. *Clin Biomech.* 25: 142-145, 2010.

21. Powers, CM, and Fisher, B. Mechanisms underlying ACL injury-prevention training: The brain-behaviour relationship. *J Ath Train*. 45(5): 513-515, 2010.

22. Yu, B, Lin, C-F, and Garrett, WE. Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech*. 21: 297-305, 2006.

23. Cortes, N, Morrison, S, Van Lunen, BL, & Onate, JA. Landing technique affects knee loading and position during athletic tasks. *J Sci Med Sport.* 15: 175-181, 2012.

24. Beaulieu, ML, and Lamontagne, M. Lower limb muscle activity and kinematics of an unanticipated cutting manoeuvre: a gender comparison. *Knee Surg Sports Traumatol Arthrosc*. 17: 968-976, 2009.

25. O'Connor, KM, and Bottum, M,C. Differences in cutting knee mechanics based on principal components analysis. *Med Sci Sport Exerc.* 41(4): 867-878, 2009.

26. Mclean, SG, Lipfert, SW, and van den Bogert, AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Med Sci Sport Exerc.* 36(6): 1008-1016, 2004.

27. Sanna, G, and O'Connor, KM. Fatigue-related changes in stance leg mechanics during sidestep cutting maneuvers. *Clin Biomech.* 23: 946-954, 2008.
28. Pollard, CD, Davis, IM, and Hamill, J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech*. 19:1022-1031, 2004.
29. Hahn, D, Olvermann, M, Richtberg, J, Seiberi, W, and Schwirtz, A. Knee and ankle torque-angle relationship of multi-joint leg extension. *J Biomech.* 44: 2059-2065, 2011.
30. Besier, TF, Lloyd, DG, Ackland, TR, and Cochrane, JL. Anticipatory effects on knee joint loading during running and cutting manoeuvres. *Med Sci Sports Exerc.* 40(7): 1176-1181, 2001.
31. Peng, H-T. Changes in biomechanical properties during drop jumps of incremental height. *J Strength Cond Res*. 25(9): 2510-2518, 2011.
32. Devita, P, and Skelley, WA. Effect of landing stiffness on joint kinetics and energies in the lower extremity. *Med Sci Sports Exerc*. 24: 108-115, 1992.
33. Zhang, S-N, Bates, BB, and Dufek, JS. Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc*. 32(4): 812-819, 2000.
34. Havens, KL, and Sigward, SM. Joint and segmental mechanics differ between cutting maneuvers in skilled athletes. *Gait Posture.* 41: 33-38, 2015.
35. Hart, N, Nimphius, S, Wilkie, J, and Newton, R. Reliability and validity of unilateral and bilateral isometric strength measures using a customized portable apparatus. *J Austr Str Cond*. 20(sup 1): 61-67, 2012.
36. Nuzzo, JL, Mcbride, JM, Cormie, P, and McCaulley, GO. Relationship between countermovement jump performance and multijoint isometric and dynamic tests of strength. *J Strength Cond Res.* 22(3): 699-707, 2008.

37. Wu, X, Zhang, S, Liu, Y, Zhang, D, and Xie, B. Do knee concentric and eccentric strength and sagittal-plane knee joint biomechanics differ between jumpers and non-jumpers in landing. *Human Mov Sci*. 32: 1299-1309, 2013.

38. Shelbourne, KB, Pandy, MG, and Torry, MR. Comparison of shear forces and ligament loading in the healthy and ACL-deficient knee during gait. *J Biomech*. 37: 313-319, 2004.

39. Solomonov, M, and Krogsgard, M. Sensorimotor control of knee stability. A review. *Scan J Med Sci Sports*. 11: 64-80, 2001.

40. Beynnon, BD, and Fleming, BC. Anterior cruciate ligament strain in-vivo: a review of previous work. *J Biomech*. 31(6): 519-525, 1998.

41. Markolf, KL, O'Neill, G, and Jackson, SR. Effects of applied quadriceps and hamstrings muscle loads on forces in the anterior and posterior cruciate ligaments. *Am J Sports Med.* 32(5): 1144-1149, 2004.

42. Shelbourne, KB, and Pandy, MG. A musculoskeletal model of the knee for evaluating ligament forces during isometric contractions. *J Biomech*. 30(2): 163-176, 1997.

43. Brughelli, M, Cronin, J, and Nosaka, K. Muscle architecture and optimum angle for the knee flexors and extensors: A comparison between cyclists and Australian rules football players. *J Strength Cond Res*. 24(3): 717-721, 2010.

44. Markolf, KL, Burchfield, DM, Shapiro, MM, Shepard, MF, Finerman, GAM, and Slauterbeck, JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orth Res.* 13(6): 930-935, 1995.

45. Sigward, SM, Guilherme, MC, Havens, KL. Predictors of frontal plane knee moments during side step cutting to 45 and 110 degrees in men and women: implications for anterior cruciate ligament injury. *Clin J Sports Med.* 0(0): 1-6, 2014.

46. Komi, PV. Stretch-shortening cycle: a powerful model to study normal and fatigue muscle. *J Biomech.* 33: 1197-1206, 2000.

47. Lakens, D. Calculating and reporting effect sizes to facilitate cumulative science: A practical primer for t-tests and ANOVAs. *Frontiers Psych*. 4:863, 2013. Doi:10.3389/fpsyg.2013.00863

48. Davis,R, Ounpuu, S, Tyburski, D, and Gage, JR. A gait analysis collection and reduction technique. *Human Movement Sci*. 10: 575-587, 1991.

49. Kadaba, MP, Ramakrishnan, HK, and Wootten, ME. Measurement of lower extremity kinematics during level walking. *J Ortho Res*. 8: 383-392, 1990.

50. Anderson, DE, Madigan, ML, and Nussbaum, MA. Maximum voluntary joint torque as a function of joint angle and angular velocity: model development and application to the lower limb. *J Biomech*. 40: 3105-3113, 2007.

51. Derrick, TR. The effects of knee contact angle on impact forces and accelerations. *Med Sci Sports Exerc*. 36(5): 832-837, 2004.

52. van Ingen Schenau, G, Bobbert, M,F, and de Haan, A. Mechnaics and energetics of the stretch-shortening cycle: A stimulating discussion. *J App Biomech*. 13: 484-496, 1997.

APPENDICES

Appendix 1

Appendix 1.1 Signed Ethics Form St Mary's University

St Mary's University Twickenham London

St Mary's University

Ethics Sub-Committee

Application for Ethical Approval (Research)

This form must be completed by any undergraduate or postgraduate student, or member of staff at St Mary's University, who is undertaking research involving contact with, or observation of, human participants.

Undergraduate and postgraduate students should have the form signed by their supervisor, and forwarded to the School Ethics Sub-Committee representative. Staff applications should be forwarded directly to the School Ethics Sub-Committee representative. All supporting documents should be merged into one PDF (in order of the checklist) and clearly entitled with your **Full Name, School, Supervisor.**

Please note that for all undergraduate research projects the supervisor is considered to be the Principal Investigator for the study.

If the proposal has been submitted for approval to an external, properly constituted ethics committee (e.g. NHS Ethics), then please submit a copy of the application and approval letter to the Secretary of the Ethics Sub-Committee. Please note that you will also be required to complete the St Mary's Application for Ethical Approval.

Before completing this form:
- Please refer to the **University's Ethical Guidelines**. As the researcher/ supervisor, you are responsible for exercising appropriate professional judgment in this review.
- Please refer to the Ethical Application System (Three Tiers) information sheet.
- Please refer to the Frequently Asked Questions and Commonly Made Mistakes sheet.
- If you are conducting research with children or young people, please ensure that you read the **Guidelines for Conducting Research with Children or Young People**, and answer the below questions with reference to the guidelines.

Please note:
In line with University Academic Regulations the signed completed Ethics Form must be included as an appendix to the final research project.

If you have any queries when completing this document, please consult your supervisor (for students) or School Ethics Sub-Committee representative (for staff) .

Approved by the Ethics Sub-Committee on the 30th April 2014.

Appendix 1.2 Subject information sheet

Human Subjects Recruiting & Advertising Form RO – F15

To be filled by Principal Investigator (PI)

1. Instructions

Insert information specific for your study where the form says to "insert."
Delete all parenthesis and text that does not apply to your study.

My name is Will Davies, and I am currently studying in my final year of a Masters in Strength and Conditioning at St Mary's University in Twickenham. I am inviting you to participate in my major project which I must complete as part of the qualification. The topic that I have chosen to study is related to strength and how it might affect change of direction technique. I am also looking to explore the cutting maneuver at a range of approach speeds, and cutting angles, to see how these conditions affect knee posture.

Participation in the study will involve two visits to the Aspire Academy biomechanics department facility in Doha, and will take about 60mins each. If you agree to participate, the first test session will require you to perform a maximal isometric, single leg squat movement at a 40deg knee angle. You will then be taken to the lab where you will complete a number of change of direction tasks that will be used in the next testing session to familiarize you to the approach speeds, and angles used during session 2. The second session will involve a warm up, before having retroreflective markers applied to your lower body so that your movements can be tracked using 3d motion analysis. You will then complete a total of 18 cutting trials at a range of velocities and angles, with a 1 min rest between each, and should take about 45 mins to 1 hour.

Cutting movements have been associated with anterior cruciate ligament injury and so some level of risk may be associated with the task. However, at the speeds encountered during this session this is unlikely. However, if you have had a previous injury to the knee in the previous 6 months you should make the researcher aware or are still in a rehabilitation protocol from a previous anterior cruciate ligament (ACL) injury, then you will be excluded from the study. The findings of this study may be reported in the final dissertation and may be in a scientific journal, but your name will remain confidential.

1. Benefits

Your participation in this study is likely to benefit you. We also hope that your participation in the study will provide us with information that will benefit others in the future.

1. Your Rights

- We may contact you regarding studies you may be interested in participating.
- We want to assure you that we will keep your information confidential.
- You do not have to be in this study if you do not want to participate.
- Your decision to be in any study is totally voluntary.
- Your care at Aspetar Qatar Orthopaedic and Sports Medicine Hospital will not be altered by your decision to participate or not participate.
- Your information will not be shared outside of this study team except to those groups inside and outside of Anti-Doping Lab Qatar, who are responsible for making sure studies are conducted correctly and ethically.

If you are interested in learning more about the study, please.

- Contact William Davies
- Complete the attached questions and mail it back to us using the self-addressed and stamped envelope. (***the questions (related to your study) must be submitted with the advertisement to the IRB for review***).
- Review the attached consent form and call the numbers below so that a researcher can talk with you about the study and answer your questions.

Sincerely,
(Signature of PI)

(William Davies)
Principal Investigator
IRB#

Contact No: 66158064

THIS STUDY HAS BEEN REVIEWED AND APPROVED BY ANTI-DOPING LAB QATAR INSTITUTIONAL REVIEW BOARD

Appendix 1.3 Subject consent form

St Mary's
University
Twickenham
London

School of Sport, Health
& Applied Science

Participant Consent Form

Project title:

"The Effect of Approach Velocity and Cutting angle on Knee Position during and its association with peak torque angle"

I have read the Letter of Information, have had the nature of the study explained to me, and I agree to participate. All questions have been answered to my satisfaction.

Subject Name (please print): ____

Signature:____Date: ____________________
____

Individual responsible for
obtaining consent:

Signature:____Date: ____________________
____

____

Investigator:

Signature:____Date: ______________

Appendix 1.4 Strength protocol data collection sheet

Test Session 1: Anthropometrics and Isometric Strength	Name:	Date:

Anthropometrics:

	Measure 1	Measure 2
Height		
Weight		
Leg Dom	L / R	

Right Leg	Measure 1	Measure 2
Length		
Knee width		
Ankle width		

Left Leg	Measure 1	Measure 2
Length		
Knee width		
Ankle width		

Strength:

	Joint Angle (°)		Strength Score (N)					Plate Adjustment
Target Angle	Knee angle	Hip angle	**Trial 1**	**Trial 2**	Trial 3	Trial 4	Highest Value	
40 Degrees								

*2nd highest value must be within 10% of highest

Notes:

Appendix 1.5 Cutting protocol data collection sheet

Testing Session 2: Cutting Trials		NAME:					DATE:			
45 Degree Cut		Markers visible	Hit the plate cleanly	No Targeting of plate	Cut angle achieved	Accelerated to finish	Approach velocity	Deceleration split	Approach Velocity met	Cut Split (Gate 2 - 3)
2m.s (1.42-1.57s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										
4m.s (0.71-0.79s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										
6m.s (0.47 - 0.53s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										

90 Degree Cut		Markers visible	Hit the plate cleanly	No Targeting of plate	Cut angle achieved	Accelerated to finish	Approach velocity value	Deceleration Split	Approach Velocity met	Cut Split (Gate 2 - 3)
2m.s (1.42 - 1.57s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										
4m.s (0.71 - 0.79s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										
6m.s (0.47 - 0.53s)	Trial 1									
	Trial 2									
	Trial 3									
Notes:										

Appendix 2

Appendix 2.1 Aspetar Research Approval

inspired by aspire®

ASPETAR | أسبيتار

REF: CMO/0000121/fj

Date: 24th March 2016

To: William Davies
Physical Coach,
NSMP department
Aspetar

Dear William Davies,

REVIEW OF STUDY: "The influence of strength on the biomechanics of the knee during cutting at different velocities and angles"

	Yes	No
Does this project support Aspetar's global research strategy?	✓	
Does this project have HOD approval?	✓	
Has consideration been given to the risk and benefits to both the research participants AND the organization?	✓	
Does this project have a process for dealing with possible harmful effects that may occur to research participants?	✓	
In line with the Declaration of Helsinki and the Belmont Report, does this project require participant informed consent?	✓	
Is study design robust enough to answer the proposed research question?	✓	
Is the research team suitably qualified to undertake this research project?	✓	
Are there any possible conflicts of interests for both the research team and the organization?		✓

I am satisfied that each of the above elements have been appropriately considered by the CMO Scientific Sub-Committee. Accordingly, your project proposal has been approved by the CMO's Scientific Sub-Committee on 23rd of March 2016.

Thank you and best regards,

Reviewed by:
Sebastien Racinais,
Head of Research Operations

On behalf of Prof Mathew Wilson
Director of Research

Approved by

Dr Scott Gillogly
Chief Medical Officer

Aspetar
PO Box 29222
Doha – Qatar

t +974 44132000
f +974 44132020
www.aspetar.com

عضو في مؤسسة أسباير زون
A member of Aspire Zone Foundation

Appendix 2.2 ADLQ ethics approval

Anti- Doping Lab Qatar Institutional Review Board

Tel: 44132988
Fax: 44132997
Email: ADLQ-RO@adlqatar.com

IRB SCH Registration: SCH-ADL-070
SCH Assurance: SCH-ADL-A-071

APPROVAL NOTICE

Date	28/04/2016
Lead Principal Investigator	Mr.William Thomas Davies – Aspetar
IRB Application #	F2016000126
Protocol Title	The influence of strength on the biomechanics of the knee during cutting at different velocities and angles
Submission Type	Initial Submission
Review Type	Full Board Review
Approval Period	28/04/2016- 27/04/2017

The Anti-Doping Lab Qatar Institutional Review Board has reviewed and approved the above referenced protocol.

As the Principal Investigator of this research project, you are responsible for:

- Ethical Compliance and protection of the rights, safety and welfare of human subjects involved in this research project.
- To follow the policies and procedures as set by ADLQ-IRB in any matters related to the project, following the ADLQ-IRB approval (i.e., with regards to obtaining prior approval of any deviation of protocol, reporting of unanticipated events, and submission of progress reports).
- To inform the ADLQ-RO of the date of commencement of the research*.

Director – ORS/ADLQ (Office of Research Support)
Ms. Noor AlMotawa

Anti Doping Lab Qatar

* For Commencement of Research, Protocol Deviation Reporting, Unanticipated Problem Reporting & Research Progress Annual Report, please contact - Education & Research Office, Anti-Doping Lab Qatar.

Appendix 2.3 Subject consent Form

مختبر مكافحة المنشطات قطر
Anti Doping Lab Qatar

ADLQ RESEARCH OFFICE
P.O.BOX 27775
Email: ADLQ-RO@adlqatar.com

Date Received:
For Administrative Use Only

Informed Consent Form (Adult Research Participants) RO-F05

Title of Research: "The influence of strength on the biomechanics of the knee during cutting at different velocities and angles"	
Protocol No *(if applicable)*:	
Principal Investigator: William Davies	PI Contact No: 66158064
Co-Investigators: Dan Cleather, Phil Graham-Smith	Co-PI Contact No: 55236240
Organization: Aspetar Orthopaedic and Sports Medicine Hospital	

Information Sheet

1. Introduction

We invite you to take part in a research study investigating how different cutting constraints (approach speed and cutting angle) affect the angle of the knee when it first contacts the ground, and the peak angle the knee achieves during the maneuvers. It will then look to see if there is a relationship between the two, as well as try to ascertain the influence strength has on this relationship. You were selected as a candidate for this trial because you have a history of playing team sports and fall in the category of our research. Please take your time to read this form, ask any question you may have before deciding to participate in this study. We encourage you to discuss your decision with your family or doctors.

2. Your Rights

IRB APPROVAL STAMP

ADLQ Informed Consent Form (Adult Research Participants) RO-F05

مختبر مكافحة المنشطات قطر
Anti Doping Lab Qatar