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The effects of mid-air adjustments on knee joint loading when landing from a jump

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The effects of mid-air adjustments on knee joint loading when landing from a jump

by

Guan Qiang Tan

A thesis submitted to the graduate faculty
in partial fulfillment of the requirements for the degree of

MASTER OF SCIENCE

Major: Kinesiology (Biological Basis of Physical Activity)

Program of Study Committee:
Timothy Derrick, Major Professor
Jason Gillette
Loren Zachary

Iowa State University

Ames, Iowa

2008

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ABSTRACT

Introduction: One of the most common injuries involving the knee joint is the Anterior Cruciate Ligament (ACL) tear. Female basketball and soccer players have about 3 times higher risk of ACL injury versus male athletes (Prodromos et al., 2007). Female athletes alter their motion pattern during the landing phase differently than male athletes. That is one of the potential reasons why females have a higher risk of ACL injury during sports (Chappell et al., 2002). Possible mechanisms of ACL injury are the deceleration of the knee in an extended position (Boden et al., 2000), shear forces applied to the leg during landing (Pflum et al., 2004) and large knee varus and internal rotation moments (McLean et al., 2007). The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing.

Methods: Eleven healthy college female students (age 21.8 ± 1.6 yrs, height 1.7 ± 0.5 m, mass 64.1 ± 11.7 kg) participated in the study. Three tennis balls were suspended from the ceiling. After maximum jump height was established, subjects were instructed to jump from and land on two force platforms (AMTI). Approximately 100 ms after leaving the force platform an LED positioned near one of the balls was illuminated. Subjects were asked to tap this ball using both hands. They were told to jump as high as possible for each of 21 randomly selected trials. A total of 23 retroreflective markers on the right and left lower extremities were used to determine the 3D orientation of the segments (Peak Motus). Inverse dynamics were used to calculate joint moments and reaction forces at the knee joint. Kinematics were imported to a scaled SIMM (MusculoGraphics, Inc.) model to obtain the maximal muscles forces, muscle moment arms and muscle orientations for 88 lower extremity muscles. Static optimization using a cost function that minimizes the sum of the
muscle stress squared was used to estimate the individual muscle forces. The knee joint contact forces were then calculated as the sum of the muscle forces and the joint reaction forces. The peak anterior shear force, peak varus moment, and peak internal rotation moment on the knee joint were used to assess the potential for ACL injury. Two 3 x 2 (reaching direction by right vs left leg and reaching direction by ipsilateral vs contralateral leg) repeated measures ANOVAs were used to determine statistical significance.

Results: The peak anterior shear force was significantly greater on the right knee compared to the left and the peak external rotation moment was significantly greater on the left knee compared to the right. However, data were also analyzed without reference to which leg had the highest peak values. The average peak difference between the middle ball condition and the greater value of the ipsilateral and contralateral legs from the side reaching conditions is 0.12BW, 0.03BWm, and 0.04BWm for peak anterior shear force, peak varus moment and peak external rotation moment respectively. This shows that peak anterior shear forces, peak varus moment and peak external rotation moments all increased in one of the legs when reaching to the side.

Discussion: The results suggest that reaching to the side balls had a higher risk of ACL injury than reaching to the middle ball. This result was not apparent when looking at right/left leg effects or ipsilateral/contralateral leg effects because subjects adopted different strategies to deal with the mid-air adjustments that are necessary when reaching to a side ball. Some subjects always landed on their dominant leg while others always landed on the ipsilateral leg.
CHAPTER I

INTRODUCTION

Introduction

The anterior cruciate ligament (ACL) is a complex three dimensional multifascicular structure in the knee joint. It functions to control excessive motion by preventing forward sliding of the tibia on the femur (Martini et al., 2002). It is one of the four ligaments that are critical to the stability of the knee joint. One of the most common injuries involving the knee joint is the ACL tear. It is estimated that 95,000 new ACL injuries occur every year with an annual incidence rate of 1 for every 3500 people in the general population (Miyasaka et al., 1991). Approximately 70 percent of the ACL injuries are non-contact and most of the injuries are related to sports (Boden et al., 2000). Female basketball and soccer players have about 3 times higher risk of ACL injury versus male athletes (Prodromos et al., 2007).

Much has been written about the intrinsic and extrinsic factors responsible for non-contact ACL injuries in sports. The size and shape of the intercondylar notch may contribute to stenosis. Hence the stenosis will cause ACL injury during impingement and external tibial rotation (Zhang et al., 2003). Additional mechanisms of ACL injury are deceleration of the body with the knee in an extended position (Boden et al., 2000) and increased shear forces applied to the lower leg during landing (Pflum et al., 2004). Differences in adjusting the lower extremity motion patterns during landing between males and females is one of the possible reasons why females have a higher risk of ACL injury (Chappell et al., 2002; Ford et al., 2005).
However, the effect of mid-air adjustment on the neuromuscular and biomechanical characteristics of the knee joint during landing is still unknown. In the laboratory setting, the subjects were aware of the performance before starting the experiment and could therefore preplan postural and adjust their motion accordingly. In real game situations, sporting maneuvers are not anticipated and are usually induced by a rapid decision due to some external stimulus such as how to block the opponent’s ball in a volleyball game or basketball game. Hence the simple landing often performed in the laboratory setting represents the idealized situation. All the unanticipated factors can lead to the mid-air adjustment of the upper extremity as the athletes need to decide what they will perform while they are in the air.

It is impossible to measure the force directly applied to the ACL on human subjects. Some groups have used a computer model to calculate the force (McLean et al., 2003; Pflum et al., 2004) and some groups measured the ACL force on cadaver knees (Fung and Zhang, 2003). Most research groups used peak anterior tibial force, peak varus/valgus moment, and peak external/internal rotation moment as the influential variables when analyzing the knee joint motion (Besier et al., 2001; Chappell et al., 2002; Cowling et al., 2003; Markolf et al., 1995; McLean et al., 2003; McLean et al., 2007; Pflum et al., 2004; Sell et al., 2006).

**Statement of the Problem**

The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing. This will be accomplished by comparing the peak anterior shear force, peak varus moment and peak internal rotation moment at the knee joint during a normal landing and during a landing in which the subjects must alter their body
position to tap a ball while in mid-air. We hypothesize that these mid-air adjustments will cause asymmetric landings that will increase the potential for ACL injury based on the criteria measures.

Review of Literature

1. Peak Anterior Shear Force

One of the risk factors of non-contact ACL injury is anterior shear force. Chappell et al. (2002) performed a study in stop-jump tasks from twenty healthy recreational athletes. The subjects were asked to do the forward jump, vertical jump and backward jump to simulate the motion in basketball, soccer and volleyball. Thirteen retroreflective markers were attached on the right and left acromioclavicular joints, anterosuperior iliac spines, thighs, lateral condyles, shanks, and lateral malleoli. Four video cameras were used to record subject’s performance of each task at 180 frames per second. The results of this study indicated that the proximal tibia anterior shear force plays an important role in the anterior tibial translation which may result in excessive strain on ACL that may result in a tear.

Deceleration and landing require an eccentric contraction to resist further knee flexion and this ballistic deceleration is partially due to high speed knee flexion and rapid quadriceps lengthening. This extensor mechanism placed rapid and large anterior shear loads on the proximal tibia, hence it may cause the disruption of ACL (Boden et al., 2000). Pflum et al. (2004) developed a 3-D model of the body to simulate drop-landing maneuvers. They entered the model muscle excitation patterns based on experimental electromyography (EMG). The input excitation patterns then were modified to generate a performance response in the model according to the experimental data. Joint angles, ground reaction forces and
muscle forces obtained from landing simulation then were applied to a model of the lower limb incorporating a 3-D model of the knee. They found that the pattern of ACL force in drop-landing can not be explained by the anterior pull of the quadriceps force alone.

Three factors contributed most significantly to the total shear force applied to the lower leg during landing: anterior shear force supplied by the patellar tendon, anterior shear force induced by the compressive force acting at the tibiofemoral joint and posterior shear force applied by the ground reaction. Some research groups believed that landing with an extended knee increased the contraction of the quadriceps so that it increased strain on ACL (Boden et al., 2000; Chappell et al., 2002; McLean et al., 2003). The large quadriceps force generates eccentric contraction, causing a large anterior shear force to be applied to the lower leg. The increase in the anterior shear force increases anterior translation of the tibia with respect to the femur thus causing the increased strain on ACL (Pflum et al., 2004; Boden et al., 2000).

2. Peak Varus/Valgus Moment

Valgus moment might be another factor needed to be considered for the injury of ACL (Boden et al., 2000; McLean et al., 2003). McLean et al. (2003) developed a 3-D computer model to predict knee joint loading during dynamic movement. They found it was especially noteworthy that the statistical distribution of predicted peak valgus moments was much broader than that of the anterior and internal rotation loads. It indicated that the valgus moment is more sensitive to initial limb posture and suggested that excessive valgus loading may be an important mechanism for the injury of the ACL during sidestepping (McLean et al., 2003).
Chappell et al. (2002) performed a study in stop-jump tasks, as mentioned before, and they found out that the female recreational athletes had valgus moments at the knee, whereas their male counterparts had varus moments at the knee during the landing phases of vertical and backward jumps. They also found out that there was no difference in the magnitude of the knee varus-valgus moment between the sexes. Bendjaballah et al. (1997) stated that moments caused by the collateral ligaments, medial in varus and lateral in valgus, increased the forces in the cruciate ligaments. Markolf et al. (1995) stated that the addition of varus moments to a knee loaded by anterior tibial force increased the force in extension and hyperextension, whereas the addition of valgus moment increased the force at flexed positions. Hence this indicated that the varus-valgus moment may not be responsible for the gender difference in the ACL strain during the stop-jump tasks (Chappell et al., 2002).

3. Peak External/Internal Rotation moment

The mechanisms of ACL injuries are still not very clear even though a lot of research had been done. Some of the findings conflict with each other. Markolf et al. (1995) stated that internal tibial moment is an important loading mechanism of the ACL for an extended knee and external tibial moment was a relatively unimportant mechanism for generating ACL force. They designed an in vitro study to measure the levels of ligament force under dual combinations of individual loading states and to determine which combinations applied higher load on the ACL. The anterior tibial shear force, varus/valgus moment and internal/external tibial moment were 100N, 10Nm and 10Nm respectively. Straight anterior tibial force was the most direct loading mechanism. The resultant force was recorded as the knee was extended from 90 degrees of flexion to 5 degree of hyperextension under the constant loading. They found the addition of internal tibial moment to a knee loaded by
anterior tibial force significantly increased the ACL force while the external tibial moment under the same loading condition dramatically decreased that force.

McLean et al. (2003) developed a 3-D model to predict knee joint loading during dynamic movement. They applied a forward dynamic 3-D musculoskeletal model of the lower extremity to represent a specific subject. The inputs to the model were the initial position, velocity of the skeletal elements, and the muscle stimulation patterns. Movement, ground reaction forces and resultant 3-D forces and moments acting across the knee joint were outputs. An optimization algorithm was applied to determine the muscle excitations that best reproduced the performance. Monte Carlo simulations determined that a peak internal rotation moment of 71Nm was required to cause the ACL rupture. Boden et al. (2000) demonstrated the opposite result. They reported that a slightly higher incidence of external tibial rotation injuries by using the comprehensive questionnaire and videotapes of ACL injuries.

4. Feed-forward Mechanism

Feed-forward mechanisms are a popular topic when discussing the ACL injury prevention program. They are used to anticipate external forces or loads to stabilize the joint, thus protecting the inherent structures. Cowling et al. (2003) investigated the effect of verbal instructions on muscle activity and risk of injury of the ACL during landing. Twenty-four athletes landed abruptly in single limb stance. Sagittal plane motion was recorded. Ground reaction forces and surface EMG data were recorded for the rectus femoris, vastus lateralis, biceps femoris, and semimembranosus muscles. Subjects were asked to perform 10 landings in each of 4 conditions: normal landing (N); repeat normal landing (R); landing after instruction to increase knee flexion (K); and landing after instruction to recruit hamstring
muscles earlier (M). They found that subjects can accurately respond to a simple verbal instruction, such as to increase knee flexion during landing. However they were unable to respond appropriately to a more complex instruction, such as a requirement to selectively change the way they fire the specific muscle groups. Their result also showed that simply flexing the knees while landing can decrease the risk against ACL injury and promote safer landings.

5. Proprioception

Proprioception may play a major role in injury reduction of the ACL (Cerulli et al., 2001; Mandelbaum et al., 2005). Proprioceptive signals can contribute to the generation of motor activity during ongoing movements. The primary function of proprioceptive reflexes in regulating voluntary movements is to adjust the motor output according to the biomechanical state of the body and limbs. It thus coordinates the pattern of motor activity during an evolving movement, and compensates the intrinsic variability of motor output (Kandel et al., 2000). The proprioceptive information is crucial for optimal motor performance. It is transferred to various motor control centers and is used to store information regarding joint position and kinesthesia to elicit active and reflexive movement (Mandelbaum et al., 2005). Mandelbaum et al., (2005) developed the Prevent Injury and Enhance Performance (PEP) Program. The program included 3 basic warm-up activities, 5 stretching techniques for the trunk and lower extremity, 3 strengthening exercises, 5 plyometric activities and 3 soccer specific agility drills. All these training techniques were primarily related to the feed-forward mechanism. Feed-forward mechanisms are a result of preactivated preparatory activation of muscle. The result of this study indicated that the PEP program significantly reduced the incidence of severe ACL injuries in the female athletes.
Anticipating a movement changes reflex responses and postural adjustments to minimize the forthcoming perturbation and maintain an appropriate posture. Patla et al. (1999) developed a study to examine how the coordination and control of body center of mass, head and body reorientation were sequenced when steering in planned early or initiated under time constraints. Six healthy males were involved in the study. They were asked to change their walking direction to 20, 40 and 60 degrees to the right respectively while they were walking straight ahead. The subjects had to plan and implement a direction change under time constraints, which was one stride duration. The result showed that the control of body center of mass in the new direction of travel was initiated first through appropriate foot placement and/or trunk roll motion.

Sporting maneuvers are not always anticipated during the real game situation. Sporting performance is often induced by a sudden external unanticipated stimulus. Most performances in the laboratory setting environment are anticipated, hence they may not represent the actual loading on the knee joint compared to the real game situation. Besier et al. (2001) studied the anticipatory effects on knee joint loading during running and cutting maneuvers. Eleven healthy male subjects performed running and cutting tasks under preplanned (PP) and unanticipated (UN) conditions. The tasks included a straight run (RUN), sidestep to 30 degrees (S30), and crossover cut to 30 degrees (XOV). For the PP condition, the appropriate LED on the target board was turned on at the beginning of the approach run. For the UN condition, an LED was turned on so that the subject was required to make the decision on which task to perform just before reaching the force platform. The delayed illumination of the LED on the target board was adjusted until the subjects had enough time to react to the LED and perform the tasks. The result showed that cutting maneuvers
performed without adequate planning may increase the risk of noncontact knee ligament injury due to the increased external varus/valgus and internal/external rotation moments applied to the knee. These results were probably due to the small amount of time to make appropriate postural adjustments before performance of the task, such as the position of the foot on the ground relative to the body center of mass.

Sell et al. (2006) developed another protocol to investigate knee joint loading during planned and reactive tasks. The subjects were instructed to begin each task at the designated starting point and land with one foot on each force platform. To include a reactive component to the stop-jump tasks, a laser coupled with photocell was instrumented into the experimental setup. The reactive task visual cue was controlled by a customized software program designed in LabView, which received the voltage signal from the photocell. During the initial jump, the subject passed through the laser beam and interrupted the voltage signal. This was detected by the LabView program to create the visual cue for the jumping direction. The results showed that the type of task (planned or reactive) significantly affected joint angles, ground reaction forces, and knee joint moments. The subjects used less knee flexion at peak posterior ground reaction force, a greater maximum knee flexion angle, and experienced greater deceleration forces, knee flexion moments and knee valgus moments during the reactive tasks. The difference between these 2 conditions indicated that the kinematic and kinetic characteristics are different during actual game situations. The studies performed primarily under planned (anticipated) conditions may underestimate knee joint loading characteristics that can cause the disruption of ACL.
Thesis Organization

This thesis is organized into Introduction, Manuscript and General Conclusions chapters. The manuscript is formatted according to Medicine & Science in Sports & Exercise specifications. References are included in the manuscript, with a full bibliography included at the end of the thesis. The informed consent documentation has been appended. An extended results section is also included as appendices.
CHAPTER II

The effects of mid-air adjustments on knee joint loading when landing from a jump

A paper to be submitted to Medicine & Science in Sports & Exercise

Guan Q. Tan and Timothy R. Derrick

Introduction: One of the most common injuries involving the knee joint is the Anterior Cruciate Ligament (ACL) tear. The female basketball and soccer players have about 3 times higher risk of ACL injury versus male athletes (Prodromos et al., 2007). Differences in adjusting the lower extremity motion patterns during landing between males and females is one of the possible reasons why females have a higher risk of ACL injury (Chappell et al., 2002). Possible mechanisms of ACL injury are the deceleration of the knee in an extended position (Boden et al., 2000), shear forces applied to the leg during landing (Pflum et al., 2004) and large knee varus and internal rotation moment (McLean et al., 2007). The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing.

Methods: Eleven healthy college female students (age 21.8 ± 1.6 yrs, height 1.7 ± 0.5 m, mass 64.1 ± 11.7 kg) participated in the study. Three tennis balls were suspended from the ceiling. After maximum jump height was established, subjects were instructed to jump from and land on two force platforms (AMTI). Approximately 100 ms after leaving the force platform an LED positioned near one of the balls was illuminated. Subjects were asked to tap this ball using both hands. They were told to jump as high as possible for each of 21 randomly selected trials. A total of 23 retroreflective markers on the right and left lower extremities were used to determine the 3D orientation of the segments (Peak Motus). Inverse
dynamics were used to calculate joint moments and reaction forces at the knee joint. Kinematics were imported to a scaled SIMM (MusculoGraphics, Inc.) model to obtain the maximal muscles forces, muscle moment arms and muscle orientations for 88 lower extremity muscles. Static optimization using a cost function that minimizes the sum of the muscle stress squared was used to estimate the individual muscle forces. The knee joint contact forces were then calculated as the sum of the muscle forces and the joint reaction forces. The peak anterior shear force, peak varus moment and peak internal rotation moment were used to assess the potential for ACL injury. Repeated measures ANOVAs was used to determine statistical significance.

Results: The peak anterior shear force was significantly greater on the right knee compared to the left and the external rotation moment was significantly greater on the left knee compared to the right. However, data were also analyzed without reference to which leg had the highest peak values. The average peak difference between the middle ball condition and the greater value of the ipsilateral and contralateral legs from the side reaching conditions is 0.12BW, 0.03BWm, and 0.04BWm for anterior shear force, varus and external rotation moment respectively. This shows that peak anterior shear forces, peak varus moments and peak external rotation moments all increased in one of the legs when reaching to the side.

Discussion: The results suggest that reaching to the side balls had a higher risk of ACL injury than reaching to the middle ball. This result was not apparent when looking at right/left leg effects or ipsilateral/contralateral leg effects because subjects adopted different strategies to deal with the mid-air adjustments that are necessary when reaching to a side ball.
For example, some subjects always landed on their dominant leg while some subjects always landed on the ipsilateral leg etc.

**Introduction**

The anterior cruciate ligament (ACL) is a complex three dimensional multifascicular structure in the knee joint and functions to control excessive motion by limiting joint mobility (Martini et al., 2002). It is one of the four ligaments that are critical to the stability of the knee joint. One of the most common injuries involving the knee joint is the ACL tear. It is estimated that 95,000 new ACL injuries occur every year with an annual incidence rate of 1 for every 3500 people in the general population (Miyasaka et al., 1991). There are not only obvious acute injury effects, but also an increased likelihood of significant long-term debilitation for the patients who suffered ACL injury. Approximately 70 percent of the ACL injuries are non-contact and most of the injuries are related to sports (Boden et al., 2000). Female basketball and soccer players have about 3 times higher risk of ACL injury versus male athletes (Prodromos et al., 2007).

Much has been written about the intrinsic and extrinsic factors responsible for the non-contact ACL injuries in sports. The size and shape of the intercondylar notch may contribute to stenosis. Hence the stenosis will cause the ACL injury during impingement and tibial external rotation (Zhang et al., 2003). The mechanisms of ACL injury were due to the deceleration with the knee in an extended position (Boden et al., 2000) and shear forces applied to the lower leg during landing (Pflum et al., 2004). In game situations, sporting maneuvers are not always anticipated and are often induced by a rapid decision due to some extra stimulus such as how to block the opponent’s smash in the volleyball game or how to
avoid a block during the jump shot. Hence the simple landing often performed in the laboratory setting represents the idealized situation. All the unanticipated factors can lead to postural adjustment as the athletes need to decide what they will perform while they were in the air. It is impossible to measure the force directly applied to the ACL on human subjects. Some groups have used computer models to calculate the ACL forces (McLean et al., 2003; Pflum et al., 2004) and some groups measured ACL forces on cadaver knees (Fung and Zhang, 2003).

Some research groups used peak anterior shear force, peak varus/valgus moment, and peak external/internal rotation moment as the influential variables when analyzing the knee joint motion (Besier et al., 2001; Chappell et al., 2002; Cowling et al., 2003; Markolf et al., 1995; McLean et al., 2003; McLean et al., 2007; Pflum et al., 2004; Sell et al., 2006). The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing.

Methods

Subjects

Based on Sell et al. (2006) data and a power analysis, a total of 11 healthy female college students were asked to participate in this study (Table 1). Subjects were excluded from the study if they had a history of serious musculoskeletal injury, any musculoskeletal injury within past 6 months or any disorder that affected musculoskeletal function or motor function. Before participation, all subjects were given written informed consent in accordance with the University’s Institutional Review Board.

Table 1. Subject Characteristics
### Instrumentation

Three-dimensional (3D) coordinate data of 2 cm diameter retroreflective markers were recorded using a 3-D Peak Motus motion analysis system (Vicon Peak Inc., Centennial, CO). This motion analysis system included 8 high-speed (120Hz) optical cameras synchronized using Peak Motus software. Ground reaction force data were collected at 3600 Hz by 2 strain gage force platforms (AMTI, Watertown, MA).

### Experimental Protocol

After written consent was obtained, anthropometric measurements were recorded for each subject to build up an anthropometric model. They included height and mass, segmental lengths and circumference of the thighs and shanks, widths of the ankles, knees and pelvis, feet length and width, lateral malleoli height and wing span. The subject’s actual maximal jump height and standing vertical reach height were recorded using a Vertec. The max jump height was the difference between the actual maximal jump height and standing vertical reach height. The horizontal reach distance was calculated based on 20% of their wing span (Table 1). A total of 23 retroreflective markers were used for the data collection of 3D coordinate data for each trial (Figure 1a,b).
Figure 1. Markers placement used during the collection of high speed video data.

Tennis balls were hung from the ceiling at 90% of the maximal jump height for the middle ball and 85% for the other two balls (Figure 2). A LED light was placed next to each of the targets (tennis balls). To include a reactive component to the jumping-landing task, a computer with Labview software was instrumented into the experiment set up to control 3 LEDs. The LEDs were hanging on the ceiling at the same level as the balls but 15cm behind them. The vertical force channel was connected to a computer that allowed adjustment of the time from when the subjects left the force platforms until the lights came on (around 100 ms). This allowed adjustments between subjects with various reaction times and jump heights. It was reset until the subject had just enough time to correctly perform each maneuver. The jumping-landing tasks were then demonstrated to each of the subjects. Subjects were instructed to begin each task by stepping onto and then jumping from the force platforms (1 foot on each force platform) and landing on the same force platforms. Subjects
were told to jump as high as possible for each trial.

![Diagram showing target setup with 20% wing span, 90% jump height, 85% jump height, and platforms labeled Left and Right.]

Figure 2. The target setup

Data Collection

After the retroreflective markers were attached to the subjects, they were allowed to practice a series of jumps designed to familiarize them with the reactive task. Each subject practiced 2 reactive tapping tasks in each direction (left, middle, right). The tapping order was randomly assigned with a total of 21 trials (7 trials for each of the 3 conditions). They had 60 seconds of rest between each trial. Total testing time for each subject was approximately 60 minutes.

Data Analysis

The kinematic data were filtered using a fourth-order Butterworth filter at a cutoff frequency of 12 Hz. Cardan joint angles were calculated using a flexion/extension,
abduction/adduction, internal/external rotation order. An inverse dynamics approach was used with a rigid body model to calculate the 3-D joint moments and reaction forces at the hip knee and ankle joints during the landings. Joint forces and moments were normalized to body mass. Segment mass, center of mass (COM) location, and segment moments of inertia were based on Vaughan et al. (1992). Cardan angles and linear coordinates of the hip joint were then imported into a scaled SIMM (MusculoGraphics, Inc) model to obtain the maximal muscles forces, muscle moment arms and muscle orientations for 88 lower extremity muscles. An interval of 500 ms after the impact was chosen as the landing phase. Static optimization was used to estimate the individual muscle forces using a cost function that minimizes the sum of the muscle stress squared. Muscle forces multiplied by the muscle moment arms were constrained to equal the joint moments calculated from the inverse dynamics. Knee joint contact forces were then calculated as the vector sum of the muscle forces and the joint reaction forces. Values were given in the local tibial reference frame. The peak anterior shear force, peak varus moment, and peak internal rotation moment were used to assess the potential for ACL injury.

Two 3 x 2 repeated measures ANOVAs (reaching direction by right vs left leg and reaching direction by ipsilateral vs contralateral leg) were used to determine statistical significance. Differing strategies among the subjects reduced the overall effects seen with this analysis, thus a post hoc test was also done on the average peak difference between the middle ball condition and the greater value of the ipsilateral and contralateral legs from the side reaching conditions. This allowed the detection of changes to the variables of interest regardless of which side of the body they occurred on.
Results

The treatment produced asymmetric landings according to the vertical ground reaction forces (Figure 3). On average, the peak compressive force in the knee increased by 0.87 BW \((p<0.01)\) when the subjects were asked to reach toward one of the side balls. The average peak values for anterior shear force, external rotation moment and varus moment are given in Table 2 and Figures 4, 5 and 6.

Table 2. Mean (SD) of selected ACL injury potential variables during landing from a jump with possible mid-air adjustments. Forces are in body weights (BW) and moments are in BWm.

<table>
<thead>
<tr>
<th>Reaching Direction</th>
<th>Left Leg</th>
<th>Right Leg</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right</td>
<td>1.09</td>
<td>0.16</td>
</tr>
<tr>
<td></td>
<td>(0.58)</td>
<td>(0.08)</td>
</tr>
<tr>
<td>Left</td>
<td>1.16</td>
<td>0.17</td>
</tr>
<tr>
<td></td>
<td>(0.64)</td>
<td>(0.07)</td>
</tr>
<tr>
<td>Middle</td>
<td>1.11</td>
<td>0.16</td>
</tr>
<tr>
<td></td>
<td>(0.57)</td>
<td>(0.07)</td>
</tr>
</tbody>
</table>

There were no significant differences in anterior shear force, rotation moment and varus moment between ipsilateral leg and contralateral leg (Table 3). The anterior shear force was significantly greater in the right leg compared to the left. All of the subjects were right leg dominant when asked to identify the leg that the kicked with. In addition, the right leg had a significantly reduced external rotation moment compared to the left.
Table 3. Statistic differences in anterior shear force, varus moment and external rotation moment between ipsilateral/contralateral, left/right

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>F Ratio</td>
<td>Prob&gt;F</td>
<td>F Ratio</td>
</tr>
<tr>
<td>Ipsi_Contra</td>
<td>0.58</td>
<td>0.57</td>
<td>0.59</td>
</tr>
<tr>
<td>Side</td>
<td>8.51</td>
<td>&lt;0.01</td>
<td>2.29</td>
</tr>
</tbody>
</table>

Data were also analyzed without reference to which leg had the highest peak values. The average peak difference between the middle ball condition and the greater value of the ipsilateral and contralateral legs from the side reaching conditions is given in Table 4. This shows that anterior shear forces, varus moments and external rotation moments all increased in one of the legs when reaching to the side. All variables showed some subjects with greater ipsilateral loading and some with greater contralateral loading (Table 3 and Figure 3).

Figure 3. Peak GRF compared in different situations
Table 4. Peak loading increases during the side reaching conditions when compared to the middle reaching condition.

<table>
<thead>
<tr>
<th>Variables</th>
<th>Absolute Peak Difference</th>
<th>P Value</th>
<th># of Ipsilateral Higher</th>
<th># of Contralateral Higher</th>
<th>Equal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ant. Shear Force (BW)</td>
<td>0.12</td>
<td>0.02</td>
<td>6</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td>Varus Mom. (B_WM)</td>
<td>0.03</td>
<td>0.01</td>
<td>6</td>
<td>4</td>
<td>1</td>
</tr>
<tr>
<td>Ext. Rot. Mom. (B_WM)</td>
<td>0.04</td>
<td>&lt;0.01</td>
<td>7</td>
<td>2</td>
<td>2</td>
</tr>
</tbody>
</table>
Figure 4a,b. Ensemble mean of left/right knee anterior tibia shear forces when reaching to different directions.
Figure 5a,b. Ensemble mean of left/right knee varus moment when reaching to different directions.
Figure 6a,b. Ensemble mean of left/right knee rotation moment when reaching to different directions. A negative value represents an external rotation moment.
Discussion

The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing. The result supported our hypothesis. Peak anterior shear force, peak varus moment and peak external rotation moment at the knee joint were examined to determine the potential for ACL injury. There were statistically significant leg effects for the anterior shear and the external rotation moment, but no significant effects for the direction of the mid-air movement. However, when the peak loads were examined regardless of the side of the body on which they occurred, all three variables were significantly greater when reaching toward the side compared to the middle.

The overall average anterior shear forces during the landing in this study (0.66±0.55 BW) were similar to those reported by Yu et al. (2003) females during a stop-jump task (0.79±0.34 BW). Sell et al (2006) reported values of 1.00±0.18 BW during a 40% body length broad jump onto the force platform followed by a maximal vertical jump. These studies both incorporated greater anterior movement than the current study and therefore slightly greater anterior shear forces could be expected.

Both valgus and varus moments have been associated with ACL injury (Bendjaballah et al., 1997; Boden et al., 2000). In our study of two-legged landing the subjects had varus moments throughout the period analyzed. Sell et al (2003) also found out the varus moment of 0.041±0.018 BWm during their reactive stop-jump task. Richards et al (1996) had the similar result (0.041±0.009 BWm) when testing the block jump of the elites volleyball players. However, Richards found out the valgus moment (0.088±0.013 BWm) was about two times larger compared to the varus moment. Chappell et al (2002) found that a valgus
moment in the forward stop-jump task and a varus moment in vertical stop-jump and backward stop-jump task.

The proximal anterior shear force is an important contributor to ACL injury because it causes direct strain on the ACL (Boden et al., 2000). Many researchers have used the posterior ground reaction force as a surrogate measure of the actual shear force. (Chappell et al., 2002, Sell et al., 2006, Yu et al., 2006). Shin et al (2007) reported that the ACL strain was reduced when applying the posterior force to their model that replicated in vivo conditions. However, Yu and Garrett (2007) reported that a great posterior ground reaction force may be associated with a great quadriceps muscle force; hence it will cause a great anterior draw force. The knee joint contact forces are the sum of the muscle forces and the joint reaction forces (McLean et al., 2003). Since it is impossible to measure the muscle force in vivo, static optimization using a cost function was used to estimate the individual muscles forces. Our results show that to some extent the increased quadriceps forces indicate the increased proximal anterior shear force. These results show that muscle forces need to be accounted for when analyzing the anterior shear force in landings.

It is still under debate whether the internal or external rotation moment is a contributor to the ACL injury. We expected to have internal rotation moments before the experiment based on the anatomical structure of the ACL. However, our result also showed a significant increase in external rotation moment when reaching to the sides compared with reaching to the middle. Fung et al (2007) reported that tibial external rotation and tibial abduction will cause the ACL to impinge against the lateral condylar notch and wrap around the notch wall hence cause elongation of the ACL.
Neuromuscular motor control is an important factor affecting lower extremity performance during landing. Anticipating a movement may change reflex responses and postural adjustments to minimize the forthcoming perturbation and maintain an appropriate posture. In our experimental setting, we tried to avoid anticipation by randomizing the trials so that they could not know which direction they would be required to reach. The subjects adopted different landing strategies to deal with the mid-air adjustment that are necessary when reaching to a side ball. Some subjects preferred landing on their dominant leg, some subjects preferred landed on the ipsilateral leg. Some subjects tried to return to the neutral position after tapping the ball when they were still in the air, thus they tended to land more symmetrical. Some subjects just reached toward the ball and landed without adjusting their posture. Some subjects just did not have enough flight time to adjust posture when they were in the air.

We conclude that initiating mid-air adjustments does increase the potential for ACL injuries but individual strategies used by participants in this study varied greatly. These individual strategies make it difficult to predict which limb will be most affected by the adjustments. It is likely that training to land symmetrically would reduce single leg loading and therefore reduce the risk of ACL injury. The hamstring muscles which act as dynamic stabilizers of the knee joint, protect the ACL. Weight-training programs should address strengthening the hamstring muscles.

References


CHAPTER III

GENERAL CONCLUSIONS

Every subject had her own landing motor control strategy. Some subjects always landed on their dominant leg, some subjects always landed on the ipsilateral leg ect. Some subjects tried to return to the neutral position after tapping the ball when they were still in the air, so that they can have the symmetrical landing. Some subjects just reached to the ball and landed without adjusting their posture. Although the increased anterior shear forces, varus moment, and rotation moment were not statistically significant in this study when looking at the right/left or ipsilateral/contralateral leg effects, the result suggests that reaching to the side balls had a higher risk of ACL injury than reaching to the middle ball. The valgus/varus and internal rotation moment may not be responsible for ACL injury without the anterior shear force however they contribute to the increasing of the ACL strain when the anterior shear force is applied. The technical training to avoid landing asymmetrically to reduce single leg loading is necessary for the athletes training program. It will help reducing the risk of ACL injury.
Appendix A

INFORMED CONSENT DOCUMENT

Title of Study: The effects of mid-air adjustments on knee joint loading when landing from a jump
Investigators: Timothy Derrick PhD, Guan Tan, Jonathan Lara, Brent Edwards MS, Stacey Meardon PT, ATC, CSCS

This is a research study. Please take your time in deciding if you would like to participate. Please feel free to ask questions at any time.

INTRODUCTION

The purpose of this study was to determine if mid-air postural adjustments affect the potential for ACL injury during the landing. You are being invited to participate in this study because you are a student in ISU without any ACL (anterior cruciate ligament) injury history.

DESCRIPTION OF PROCEDURES

If you agree to participate in this study, your participation will last for about two hours and a half and involve one visit the department. During the study you may expect the following study procedures to be followed: after written consent was obtained, anthropometric measurements will be recorded. They included height and mass, segmental lengths and circumference of the thighs and shanks, diameters of the ankles and knees, feet length and width, lateral malleoli height, and pelvic width. A total of 23 retroreflective markers will be used for the data collection of 3D coordination data for each trial. The markers will be placed bilaterally over the foot, calf, thigh, lower and upper torso, arm, forearm, and hand. The markers will be secured to the subjects with double-sided tape. The jumping-landing tasks will be then demonstrated to you. Limited instructions will be given to not affect the performance of the task. You will be instructed to begin each task by jumping from the force platform (1 foot on each force platform) and landing to the same force platform. When you are in the air, you need to tap the tennis ball (left, middle, or right) which had been turned on. You need to jump as high as possible for each trial. You are allowed to practice each of the jumps until you are comfortable with the task. Three-dimension (3D) coordinate data of the retroreflective markers (d=2cm) during the jump-landing were recorded using a 3-D Peak Motus motion analysis system (Vicon Peak Inc., Centennial, CO). This motion analysis system included 8 high-speed (120Hz) optical cameras instrumented and synchronized using Peak Motus software. Ground reaction force data during the jumping and landing tasks were collected at 3600 Hz by 2 strain gauge force platforms (Advanced Mechanical Technology Inc, Watertown, MA) that were located within a custom-built flooring system in which the force platforms were flush with the surrounding surface. A computer was connected with the force platform to control the sequence of the light emitting diodes (LEDs).
RISKS

While participating in this study you may experience the following risks: slight to medium fatigue during the jumping and landing performance. You are able to stop the experiment at any time if you are feeling uncomfortable. No long-term muscle or joint soreness is expected to result from the jumping and landing activity.

BENEFITS

If you decide to participate in this study there may be no direct benefit to you. It is hoped that the information gained in this study will benefit society by helping find out the effect of multitasks on the neuromuscular and biomechanical characteristics of the knee joint during landing to minimize the probability of ACL injury.

COSTS AND COMPENSATION

You will not have any costs from participating in this study. You will not be compensated for participating in this study.

PARTICIPANT RIGHTS

Your participation in this study is completely voluntary and you may refuse to participate or leave the study at any time. If you decide to not participate in the study or leave the study early, it will not result in any penalty or loss of benefits to which you are otherwise entitled.

RESEARCH INJURY

Emergency treatment of any injuries that may occur as a direct result of participation in this research is available at the Iowa State University Thomas B. Thielen Student Health Center, and/or referred to Mary Greeley Medical Center or another physician or medical facility at the location of the research activity. Compensation for any injuries will be paid if it is determined under the Iowa Tort Claims Act, Chapter 669 Iowa Code. Claims for compensation should be submitted on approved forms to the State Appeals Board and are available from the Iowa State University Office of Risk Management and Insurance.

CONFIDENTIALITY

Records identifying participants will be kept confidential to the extent permitted by applicable laws and regulations and will not be made publicly available. However, federal government regulatory agencies and the Institutional Review Board (a committee that reviews and approves human subject research studies) may inspect and/or copy your records for quality assurance and data analysis. These records may contain private information.

To ensure confidentiality to the extent permitted by law, the following measures will be taken subjects will be assigned a unique code and letter and will be used on forms instead of
their name; these identifiers will be kept with the data. Only the stated investigators will have access to study records. Additionally, records will be kept confidential, in a locked filing cabinet and/or password protected computer file for the duration of the study. If the results are published, your identity will remain confidential.

QUESTIONS OR PROBLEMS

You are encouraged to ask questions at any time during this study.

- For further information about the study contact Timothy R. Derrick, PhD at 515-294-8438, tderrick@iastate.edu
- If you have any questions about the rights of research subjects or research-related injury, please contact the IRB Administrator, (515) 294-4566, IRB@iastate.edu, or Director, (515) 294-3115, Office of Research Assurances, Iowa State University, Ames, Iowa 50011.

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PARTICIPANT SIGNATURE

Your signature indicates that you voluntarily agree to participate in this study, that the study has been explained to you, that you have been given the time to read the document and that your questions have been satisfactorily answered. You will receive a copy of the written informed consent prior to your participation in the study.

Participant’s Name (printed)  

______________________________

(Participant’s Signature)          (Date)
Figure 7a,b. Ensemble mean of left/right gluteus maximus muscle forces when reaching to different directions
Figure 8a,b. Ensemble mean of left/right vastus lateralis, vastus medialis and vastus lateralis muscle forces when reaching to different directions.
Figure 9a,b. Ensemble mean of left/right rectus femoris muscle force when reaching to different directions
Figure 10a,b. Ensemble mean of left/right tibialis anterior muscle forces when reaching to different directions
Figure 11a,b. Ensemble mean of left/right hamstring forces when reaching to different directions
Figure 12a,b. Ensemble mean of left/right gastrocnemius muscle forces when reaching to different directions
Figure 13a,b. Ensemble mean of left/right soleus muscle forces when reaching to different directions
REFERENCES


