THE RELATIONSHIPS BETWEEN PERFORMANCE AND ACL LOADING DURING ATHLETIC TASKS

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ABSTRACT

BOYI DAI: The Relationships between Performance and ACL Loading during Athletic Tasks
(Under the direction of Bing Yu, PhD)

Anterior cruciate ligament (ACL) injuries are common sports related knee injuries. While increasing performance and decreasing ACL injury risks are both important for athletes, the underlying relationships between performance and ACL loading are still unknown. Studying the relationships between performance and ACL loading can provide important information in understanding injury mechanism as well as developing injury prevention strategies.

In the current study, eighteen male and eighteen female collegiate aged recreational athletes conducted stop-jump and side-cutting tasks with different performance demands and techniques. Performance including jump height, approach speed, take-off speed, stance time, and mechanical work were evaluated among different jumping and cutting conditions. Peak ACL forces were estimated from an ACL loading model. ACL loading variables and peak ACL force variables were compared among different jumping and cutting conditions. The acute effects of performance demands on ACL loading were evaluated. The acute effects of movement patterns that should decrease ACL loading on performance outcomes were determined. Gender differences in lower extremity biomechanics were evaluated as a secondary purpose.

ACL loading increased when the movement speed increased. Soft landing and landing with increased knee flexion decreased ACL loading but also decreased jump height and movement speed and increased mechanical work, which indicated decreased performance. Males and females demonstrated different knee sagittal plane motion.

For individuals whose priority is injury prevention but not performance, adapting a slow movement pattern or soft landing pattern might decrease ACL injury risks. However, fast movements might not be avoidable during real sports competitions. The results suggest the importance of considering performance and ACL loading as a combined unit during injury risk evaluation and injury prevention. It is necessary to completely report the changes in performance in order to have a thorough understanding of training effects. The gender differences might provide information in developing gender specific ACL injury prevention programs. However, the current study only evaluated the acute relationships between performance and ACL loading. Long-term training effects on the relationships between performance and ACL loading need further investigations.

DEDICATION

To my grandma Chunhua Wang, my parents Yuewen Zhao and Qiji Dai, and my wife Lihan Deng

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LIST OF ABBREVIATIONS AND SYMBOLS

ACL	= Anterior cruciate ligament
A_A	= Ankle plantarflexion (+) - dorsiflexion (-) angle
A_H	= Hip flexion (+) - extension (-) angle
A_K	= Knee flexion (+) - extension (-) angle
ASIS	= Anterior superior iliac spines
F_{ACL}	= ACL force
F_{AS}	= Tibial anterior shear force
F_{GAS}	= Gastrocnemius force
F_{GM}	= Gluteus maximus force
F_{HAM}	= Hamstring force
$F_{K,AP}$	= Knee resultant force in the anterior-posterior direction
$F_{K,SI}$	= Knee resultant force in the superior-inferior direction
F_{PT}	= Patella tendon force
F_{SOF}	= Knee soft tissue force
F_{SOL}	= Soleus force
F_{TF}	= Tibiofemoral contact force
F_{100}	= ACL force with a 100 N tibial anterior tibia force
F_{10_ER}	= ACL force with a 10Nm knee external rotation moment
F_{10_IR}	= ACL force with a 10Nm knee internal rotation moment
F_{10_VAL}	= ACL force with a 10Nm knee valgus moment
F_{10_VAR}	= ACL force with a 10Nm knee varus moment

GRF	= Ground reaction force
$M_{A,P}$	= Ankle plantarflexion moment
M_{ER}	= Knee external rotation moment
$M_{H,E}$	= Hip extension moment
M_{IR}	= Knee internal rotation moment
$M_{K,EF}$	= Knee extension - flexion moment
M_{VAL}	= Knee valgus moment
M_{VAR}	= Knee varus moment
PSIS	= Posterior superior iliac spines
r_{AT}	= Achilles tendon moment arm for ankle joint
r_{GAS}	= Gastrocnemius moment arm for knee joint
r_{GM}	= Gluteus maximus moment arm for hip joint
$r_{HAM,H}$	= Hamstring moment arm for hip joint
$r_{HAM,K}$	= Hamstring moment arm for knee joint
r_{PT}	= Patellar tendon moment arm for knee joint
δ	= Posterior tibial plateau slope angle
α	= Patellar tendon - tibial shaft angle
β	= Hamstring-tibial shaft angle
γ	= Gastrocnemius - tibia shaft angle

CHAPTER I

INTRODUCTION

Anterior cruciate ligament (ACL) injuries are common sports related knee injuries. The estimated annual incidence rate of ACL injury is 1 in every 3,000 citizens in the United States and Norway (Miyasaka et al., 1991; Granan et al., 2008). A majority of ACL injuries occur during the landing phase of sudden deceleration maneuvers with a non-contact mechanism (Boden et al., 2000; Krosshaug et al., 2007).

Female athletes have greater ACL injury rates per sports exposure than male athletes in most sporting events (Agel et al., 2005; Hootman et al., 2007). Previous researchers have attempted to identify risk factors for non-contact ACL injury by comparing movement patterns between males and females (Malinzak et al., 2001; Chappell et al., 2002; Yu et al., 2006; Chappell et al., 2007). These studies have shown that females have restricted sagittal plane joint motion and increased joint motion in the frontal and coronal planes compared to males when performing jump landing and cutting tasks (Malinzak et al., 2001; Chappell et al., 2002; Yu et al., 2006; Chappell et al., 2007). The findings of motion analysis studies are consistent with in *vitro* and in *vivo* studies which have shown that the biomechanical movement patterns demonstrated by females induce greater ACL loading compared to males (Berns et al., 1992; Markolf et al., 1995; Weinhold et al., 2007).

In the studies which compared gender differences, as well as other studies that investigated ACL injury risk factors (Hewett et al., 2005) and evaluated training effects (Myer et al., 2005), jump landing and cutting tasks were commonly used to evaluate movement patterns. For example, strength and technique training decreased knee anterior shear forces and increased knee flexion angles during a stop-jump task in female recreational athletes (Herman et al., 2009). Modifications in foot placement and body position reduced peak valgus moments during a side-cutting task in male team sport athletes (Dempsey et al., 2007). However, while many investigators have focused on the effects of interventions on ACL loading factors, reports of changes in performance were lacking (Prapavessis and McNair, 1999; Onate et al., 2005; Onate et al., 2005; Dempsey et al., 2007). Most investigators only used jump height and running speed as the performance variables when evaluating jump landing and cutting tasks (Hewett et al., 1996; Yu et al., 2004; Myer et al., 2005; Sigward and Powers, 2006b; Vescovi et al., 2008; Dempsey et al., 2009; Herman et al., 2009). Only a few investigators reported stance time as a performance variable (Chappell and Limpisvasti, 2008; Myers and Hawkins, 2010; Wannop et al., 2010).

During real competition, achieving optimal performance is important for athletes. For example, a basketball player with a higher jump height and a shorter take-off time will have advantages in scoring, rebounding, and blocking. In addition, reducing energy expenditure during each movement should allow athletes to play longer with a greater intensity. From the injury prevention perspective, reducing ACL loading or loading factors such as peak ground reaction forces and small knee

flexion angles are important (Meyer and Haut, 2005; Jordan et al., 2007; Yu and Garrett, 2007; Taylor et al., 2011). On the other hand, from the performance perspective, faster running speed and greater jump height, short take-off time, and low energy expenditure are desirable in most sports events. However, while increasing performance and decreasing ACL loading are both important for athletes, the underlying relationships between them are unknown.

The relationships between performance and ACL loading can be studied between individuals or within individuals. Comparing ACL loading between individuals with different performance levels might introduce other factors other than performance. For example, male athletes usually perform at a higher level than female athletes (Ziv and Lidor, 2009). Because of the differences in movement patterns (Malinzak et al., 2001; Chappell et al., 2002) and internal ACL risk factors (Nunley et al., 2003; Chandrashekar et al., 2005; Chandrashekar et al., 2006) between genders, males have a lower ACL injury rate than females (Agel et al., 2005; Hootman et al., 2007). Indeed, ACL injuries are not likely related to skill levels (Harmon and Dick, 1998). No difference in ACL injury rate was found among different NCAA division levels in men’s or women's basketball or soccer (Harmon and Dick, 1998). However, studying the relationships between performance and ACL loading within individuals can partition out factors other than performance.

To determine the relationships between performance and ACL loading within individuals, at least two questions need to be answered. The first question is how changes in performance demands affect ACL loading. The second question is how changes in movement patterns that should decrease ACL loading on performance

outcomes. decrease ACL loading affect performance outcomes. Previous investigators who studied the effects of performance demands on lower extremity biomechanics focused on the effects of drop height on landing biomechanics. Generally, impact ground reaction forces increased as drop heights increased (Dufek and Bates, 1990; Zhang et al., 2000; Elvin et al., 2007; Yeow et al., 2010). In addition, the impact ground reaction forces increased during a drop jump task when subjects jumped at a faster speed (Walsh et al., 2004). However, the focus of previous studies was not ACL loading. It was unknown how ACL loading changed when the drop height increased or the support phase of a jump decreased. In addition, drop landing or drop vertical jump from a box were rarely performed during real competitions. Compared to drop landing and drop vertical jump from a box, stop jump and cutting tasks were commonly performed in sports such as basketball and soccer. In addition, because of a fast initial approach speed, the landing phase of stop-jump and cutting tasks involve sharp decelerations in the anterior-posterior direction which might increase the ACL loading (Yu et al., 2006). However, it is unknown how different performance demands such as jump height and stance time affect ACL loading during stop-jump and cutting tasks.

In terms of the effects of changes in movement patterns that should decrease ACL loading on the performance outcomes, different researchers have investigated the effects of landing techniques (Dufek and Bates, 1990; Zhang et al., 2000; Elvin et al., 2007; Yeow et al., 2010) on stance time (Walsh et al., 2007), jump height (Walsh et al., 2007), and mechanical work (Devita and Skelly, 1992; Zhang et al., 2000) during drop landing and drop vertical jump tasks. Walsh et al. (2007) found

that soft landing decreased the impact force without changing jump height during a drop vertical jump task in females. However, the stance time increased. The authors stated that the increase in stance time would put athletes at a competitive disadvantage. In biomechanical studies, energy expenditure was usually quantified using mechanical work (McCaulley et al., 2007). Previous investigators have demonstrated that soft landing reduced peak ground reaction forces and peak joint moments but increased lower extremity mechanical work when compared to stiff landing (Devita and Skelly, 1992; Zhang et al., 2000). A similar trade-off between injury risk and energy expenditure was suggested in running. Running with increased knee flexion angles can decrease impact forces, but will increase energy expenditure that might reduce performance (Derrick, 2004). Previous researchers have demonstrated that changes in technique such as soft landing and landing with increased knee flexion are effective in decreasing ACL loading factors. However, the changes in performance outcomes were largely unknown in each study. The combined results of previous studies suggest that decreases in ACL loading induced by soft landing are likely to increase stance time and mechanical work during drop landing and drop vertical jump tasks. However, no study has comprehensively examined how changes in ACL loading induced by soft landing affect performance during athletic tasks with great decelerations such as stop-jump and cutting tasks.

The previously mentioned studies suggest potential trade-off relationships between performance and ACL loading during athletic tasks. However, because those relationships are not completely understood, many researchers have treated performance and ACL loading as two independent factors during movement

evaluations. For example, investigators have identified that neuromuscular training decreased ACL loading factors during a jump landing test and increased vertical jump height during a maximum vertical jump test (Hewett et al., 1996; Myer et al., 2005; Myer et al., 2006a; Myer et al., 2006b; Chappell and Limpisvasti, 2008). Evaluating ACL loading and maximum jump height using two different tasks might favor one aspect without considering the other. Actually, by using a single jump landing task to evaluate both ACL loading factors and jump height simultaneously, several investigators did not find training improved lower extremity biomechanics or jump height (Grandstrand et al., 2006; Vescovi et al., 2008; Lim et al., 2009). The discrepancies among previous studies in training effects on performance and ACL loading might be caused by different training programs and characteristics of subjects. However, the differences in testing protocols should also be noticed. Many studies considered performance and ACL loading as two independent factors and had them tested during two different tasks. A lack of consideration of the relationship between performance and loading could also contribute to the discrepancies among previous studies.

Although previous investigators have documented some relationships between performance and ACL loading factors, no study has systematically studied their relationships simultaneously. A lack of comprehensive consideration of the relationships between performance and ACL loading might mislead the actual training effects on ACL loading in real practice and competitions. Previous investigators have documented that the ACL injury rate for NCAA women's soccer, men's basketball, and women's basketball remained unchanged from 1990 to 2002

(Agel et al., 2005). ACL injury rate for 15 sports was increasing by an average of 1.3% each year 1988 to 2004 (Hootman et al., 2007). These findings suggest that either the implementation of prevention programs is ineffective or lacking, or improvements are needed for the current prevention programs for ACL injury. Studying the underlying relationships between performance and ACL loading can provide important information in understanding injury mechanism as well as developing effective prevention strategies.

Statement of Purposes and Hypotheses

The purpose of the current study was to determine the relationships between the performance of recreational athletes in selected athletic tasks and their ACL loading while performing the athletic tasks. Performance was quantified using jump height, approach speed, take-off speed, stance time, and lower extremity mechanical work. ACL loading was quantified using biomechanical ACL loading factors including GRF, 3 dimensional knee angles, and knee moments. ACL loading was also quantified using peak ACL force estimated from a musculoskeletal model. This purpose was achieved by addressing the following two specific aims:

Specific Aim 1: to determine the effects of changes in performance demands on ACL loading in recreational athletes while performing stop-jump and side-cutting tasks.

Hypothesis 1: ACL loading would increase when the athletes jumped with a higher jump height and a shorter stance time during a stop-jump task. ACL loading would increase when the athletes cut with a faster speed and a shorter stance time during a side-cutting task.

Specific Aim 2: to determine the effects of changes in movement patterns that should decrease ACL loading on the performance outcomes.

Hypothesis 2: Soft landing and landing with increased knee flexion at initial contact would decrease ACL loading, but also decrease jump height and cutting speed and increase stance time and mechanical work compared to regular landing during stop-jump and side-cutting tasks.

Significances of the Study

1. Understanding the effects of performance demands on ACL loading could provide insight of ACL injury mechanisms, and set the targets for injury risk screening and injury prevention. ACL injuries usually occur during sharp deceleration athletic tasks, but how performance demands during these athletic tasks affect ACL loading and which aspect of performance demands has the greatest influence to ACL loading are unknown. Previous investigators generally screened and trained athletes with maximum jump height as the performance demand with the assumption that jump height was the factor mostly associated with ACL loading. However, it was still unknown whether jump height was a sensitive performance demand to ACL loading. In addition, movement speed, which was more likely to be associated with a sharp deceleration, had received little attention during injury risk screening. Screening and training athletes without knowing whether the task demands really representing the ACL injury scenario might set the wrong target in injury risk screening and injury prevention. The findings of the current study would provide information in ACL injury mechanism in terms of whether performance demands was associated with ACL loading and which performance demand had the greatest effect on ACL loading. The goal of training is to decrease injury risk factor during high risk tasks. Performance demands might be an impotent factor in determining where risk level of a task.

Injury risk screening and prevention should be conducted with a comprehensive consideration of performance demands.

2. Knowing the effects of changes in ACL loading induced by changes in techniques on performance would give us a better understand of the generalizations of certain training methods to the real world. Changes in techniques such as soft landing and landing with increased knee flexion were effective in decreasing ACL loading factors and had been an important component in many ACL prevention programs. However, many investigators only focused on the decreases in ACL loading caused by changes in techniques without a full understanding of the changes in performance. No study had comprehensively and consistently investigated the effects of soft landing and landing with increase knee flexion on performance. In addition, previous studies only studied drop landing and drop vertical jump from a box which were rarely performed during real competitions. Different from a lab setting, performance was very important during real competitions for athletes to achieve their sports goals. The improvements in movement patterns induced by changes in techniques could vanish during real competitions if the new movement patter causes a decrease in performance. The findings of the current study might reveal the limitations of certain training methods and explain why it might not be generalized to the real competition.

3. Examining the relationships between performance and ACL loading would provide implications to the future development of movement evaluation tests and criteria for reporting intervention effects on movement patterns in ACL injury prevention programs. Many previous investigators had evaluated performance and ACL loading as two independent factors. In addition, reports in changes in performance were usually incomplete. Evaluating performance and ACL loading individually might favor one aspect without considering the other during different tests. The findings of the current study would demonstrate whether we need to consider performance and ACL loading as a single unit during movement evaluations and injury prevention training. In addition, the relationships between performance and ACL loading would suggest the importance of presenting a complete report of interventions effects on both performance and ACL loading.

CHAPTER II

REVIEW OF LITERATURE

The incidence rates of ACL injuries and consequences of ACL injuries have been reviewed to justify the importance of studying and preventing ACL injuries. The characteristics of ACL injuries have been reviewed to justify the strategies that are chosen to evaluate ACL injury risks. ACL loading mechanisms have been reviewed to justify the biomechanical ACL loading factors. Previous studies that investigated performance and lower extremity biomechanics during athletic tasks have been critically reviewed to justify the novelty of the current study and provide background information to build the hypotheses for the current study.

2.1. Incidence Rates of ACL Injuries

Previous researchers have demonstrated that the annual incidence rate of ACL injury is approximately 1 in every 3,000 citizens. A majority of ACL injuries occur during sports activities. In Norway, 2714 primary ACL reconstructions were performed from 2004 to 2006 with an annual incidence rate of 34 per 100, 000 citizens (Granan et al., 2008). One thousand and seven hundred of the 2714 (65%) injuries occurred in soccer, team handball, and alpine skiing. In New Zealand, 7375 ACL injuries have been claimed from 2000 to 2005 resulting in an annual

incidence rate of 37 per 100,000 citizens (Gianotti et al., 2009). In addition, 3997 of the 7375 (54%) injuries were sports related injuries. In the United States, no national registry exists for ACL injury, so the ACL injury rate in general population is estimated from insurance data within each medical center. Three hundred and two ACL injuries were reported at Kaiser-Permanente Medical Center in San Diego from 1985 to 1988 resulting in an estimated annual incidence rate of 36 per 100,000 citizens (Miyasaka et al., 1991). 204 of the 302 (68%) injuries were sports related. 4485 ACL reconstructions were performed within Kaiser-Permanente Southern California between 2001 and 2005 corresponding to an estimated annual incidence rate of 29 per 100000 citizens (Csintalan et al., 2008).

The ACL injury rates in sporting events are usually reported as the number of injuries per 1000 exposures to normalize the sports exposure effects. Prodromos et al. (Prodromos et al., 2007) conducted a Meta-analysis study and reviewed recent literature of ACL injury rates during different sports. The ACL injury incidence rates (injuries / 1000 exposures) are 2.78 in general population indoor soccer, 0.49 in general population alpine skiing, 0.33 in elite handball, 0.25 in collegiate wrestling, 0.22 in collegiate rugby, 0.21 in collegiate soccer, 0.18 in collegiate lacrosse, 0.17 in collegiate basketball, and 0.07 in adult recreational soccer.

Previous investigators have documented that the ACL injury rate for NCAA women's soccer, men's basketball, and women's basketball remained unchanged from 1990 to 2002 (Agel et al., 2005). ACL injury rate for 15 NCAA sports was increasing by an average of 1.3% each year 1988 to 2004 (Hootman et al., 2007).

2.2. Consequences of ACL Injuries

Approximately 200,000 ACL reconstructions are performed annually in the United States at an average cost of $20,000 per surgery. This translates to an estimated annual cost of $4 billion for surgical costs alone (Brophy et al., 2009). ACL injuries not only bring financial burden to the health service and society, but also have devastating consequences on patients' quality of life and result in secondary injuries and disorders (Ingersoll et al., 2008).

2.2.1. Physical and Psychological Consequences

ACL injuries have long-term hazardous effects on patients' physical capabilities. Patients following ACL injuries usually demonstrate decreases in quadriceps strength, knee proprioception, and physical activity levels compared to pre-injury (Keays et al., 2001; de Jong et al., 2007; Ingersoll et al., 2008; Lautamies et al., 2008). Keays et al. (2001) measured quadriceps and hamstring strength in ACL injured patients 1 week before and 6 month after their ACL reconstructions. Before the surgery, 9-12% deficits of quadriceps strength were found on the surgical knee compared to non-surgical knee. However, after the surgery and rehabilitation training, the deficits of quadriceps increased to 22-28%. de Jong et al. (2007) measured the quadriceps and hamstring strength in ACL injured patients from pre-surgery to 12 month post-surgery. Quadriceps strength asymmetries were 10-20% at pre-surgery, 20%-40% at 6 month post-surgery, and 10-20% at 12 month post-surgery. The asymmetries in hamstring strength were within 10% from pre-surgery

to 12 month post-surgery. Lautamies et al. (2008) assessed quadriceps and hamstring strength and knee function in patients 5 years after ACL reconstruction. The strength deficits for quadriceps were 5-10% between surgical and non-surgical knees. More than 30% of the patients had a greater than 10% asymmetry in a single leg hop test. Fischer-Rasmussen and Jensen (2000) compared the proprioception and performance between an ACL injured group and a control group. The performance was tested using one-leg leap test and triple jump test. In a one-leg triple test, the subjects jumped up and down a step until exhausted and the number of jumps was recorded. During a triple jump test, subjects performed three continuous jumps and the total jump distance was measured. The ACL injured group had less performance in one-step leap test and triple jump test on the surgical leg compared to the control group. Proprioception was tested by reproducing 60 degrees of knee flexion and detecting passive movements. The ACL injured group had less scores on both proprioceptive tests than the control group. Ingersoll et al. (2008) summarized evidences that ACL injuries had hazardous effects on somatosensory, muscle activation, muscles strength, atrophy, balance, biomechanics, and patient-oriented outcomes.

ACL injured patients also show a fear of ACL re-injury. Kvist et al. (2005) followed ACL injured patients for 3-4 years post-surgery. Sixty-two patients completed a questionnaire including the Tampa Scale of Kinesiophobia, the Knee Injury and Osteoarthritis Outcome Score, and general questions. Only 53% of the patient returned to pre-injury activity level. In addition, fear of re-injury was

associated with low rate of activity level and low knee-related quality of life (Kvist et al., 2005).

2.2.2. Neuromuscular Control

One goal in rehabilitation following ACL injuries is to restore patients' neuromuscular control patterns with an aim to return patients to pre-injury activity level. However, previous studies have demonstrated that ACL injured patients had abnormal neuromuscular control patterns during athletic tasks even after rehabilitation training.

Bush-Joseph et al. (2001) compared the physical functions and lower extremity biomechanics between ACL injured patients with an average of 22 month following surgery and a healthy control group. The patients' surgical knees had good range of motion, strength, and stability compared to nonsurgical knees and the control group. No difference was observed in knee extension moments between two groups during light and moderate tasks including walking and stair climbing. However, decreased knee extension moments on the patients' surgical side were found during great demanding tasks including jogging and cutting. Decker et al. (2002) compared the landing strategies between ACL injured patients more than 1 year following surgery and a healthy control group. The ACL injured group had increased ankle range of motion but decreased hip flexion. The time to peak vertical ground reaction force was delayed in the ACL injured group. The ACL injured group also had 37% more ankle plantarflexor work and 39% less hip extensor work compared to the control group. Vairo et al. (2008) studied the lower extremity

biomechanics during a single-leg drop landing task in ACL injured patients 21 month following surgery. No significant differences were observed in summated lower extremity extension moments. However, the surgical side had less peak vertical ground reaction forces and increased peak hip flexion angles compared to the surgical side. Paterno et al. (2007) studied the peak ground reaction forces and loading rates during a drop vertical jump task in patients 27 month following ACL surgery. The surgical side had less peak ground reaction forces during both landing and take-off phases and less loading rate during the landing phase compared to the nonsurgical side.

Paterno et al. (2010) conducted a prospective study to predict ACL re-injuries from landing biomechanics and postural stability in young athletes. Although the comparisons between surgical and nonsurgical sides were not presented in detail, the investigators found that increased hip internal rotation moments during early landing phase, increased valgus movements, increased asymmetries in knee extension moments at initial contact, and poor postural stability of the involved limb predicted the ACL re-injuries with great sensitivity and specificity.

In summary, individuals with ACL injuries demonstrated asymmetries in lower extremity kinematics and kinetics during athletics tasks. In addition, the limb asymmetries became more pronounced when the task demands increase. The asymmetries in neuromuscular control could be caused by decreases in strength, proprioception, and a fear of ACL re-injury. The asymmetries might contribute to the greater ACL re-injury rate.

2.2.3. Osteoarthritis

Osteoarthritis is a common age-related disorder which is characterized by loss of articular cartilage in synovial joints. However, ACL injuries can cause the early onset and great prevalence of knee osteoarthritis (Lohmander et al., 2004; Lohmander et al., 2007). The loss of cartilage at the knee joint can expose and damage the bone and lead to tremendous pain and functional impairments (Lohmander et al., 2007). von Porat et al. (2004) followed 219 ACL injured male soccer players for 14 years following surgery. Among the 122 patients who had radiography 14 years following surgery, 41% of the injured knees reached the criterion of radiographic knee osteoarthritis compared to only 4% of the contralateral knee. Lohmander et al. (2004) followed 84 ACL injured soccer players 12 years following surgery. Among the 67 patients who received a knee radiograph 12 years following surgery, 82% had radiographic changes in the surgical knee and 51% reached the criterion of radiographic knee osteoarthritis. Lohmander et al. (2007) showed that an average of 50% patients with ACL or meniscus tear had osteoarthritis with associated pain and functional impairments 10-20 years following injury.

2.2.4. ACL Re-injuries

The ACL re-injury rate was much greater than the primary ACL injury rate, especially in young and adolescent athletes (Shelbourne et al., 2009). In addition,

the risk of injuring the contralateral ACL is as great as reinjuring the ACL graft (Shelbourne et al., 2009).

Two studies with large sample sizes have shown that the incident rates of ACL graft rupture and contralateral ACL rupture were both 5-6% in general patients 2-5 years following ACL reconstruction (Salmon et al., 2005; Shelbourne et al., 2009). Shelbourne et al. (2009) followed 413 ACL injured adolescent basketball and soccer players for a mean of 10 years. The ACL graft tear rate was 10% and the contralateral ACL injury rate was 16%. A study followed 56 young athletes with ACL injuries for one year after they returned to sports (Paterno et al., 2010). The investigators showed a re-injury rate of 5% for the graft rupture and 18% for the contralateral ACL.

2.3. Characteristics of ACL Injuries

2.3.1. Non-contact Mechanism

Previous researchers used video analysis and questionnaire surveys to understand the nature of ACL injury mechanisms. The investigators have found that the majority of ACL injuries occur with no direct contact with other players or external objects. Boden et al. (2000) reviewed 27 videos of ACL injuries and surveyed 89 ACL injured athletes. More than 70% of the injuries occurred with a non-contact mechanism. Krosshaug et al. (2007) analyzed 39 videos of ACL injuries during basketball games. More than 70% of the injuries happened with a non-contact mechanism. Agel et al. (2005; 2007) reviewed ACL injuries in collegiate soccer and

basketball. 65% of the injuries were non-contact injuries. Fauno and Wulff Jakobsen (2006) performed a retrospective survey study and found that more than 80% of ACL injuries in soccer occurred with a non-contact mechanism. In addition, the New Zealand injury claim data revealed that 58% of the sports related ACL injuries were non-contact ACL injuries. The registry data for Norwegian elite handball teams showed that more than 90% of the ACL injuries were non-contact ACL injuries (Myklebust et al., 1997).

In summary, previous researchers have found that 58-90% of ACL injuries occurred with a non-contact mechanism. The non-contact nature of most ACL injuries suggested that inappropriate neuromuscular control which resulted in awkward postures and imbalance force distributions were likely the major causes of most ACL injuries. ACL injuries might be preventable through appropriate neuromuscular and technique training.

2.3.2. Tasks and Timing

Studying the tasks and timing when ACL injuries occur is important to understand the mechanism of ACL injuries. Previous researchers have demonstrated that non-contact ACL injuries usually occur during the early phase of landing, cutting, pivoting, and other athletic tasks with sudden decelerations. Boden et al. (2000) found that most non-contact ACL injuries occurred at foot strike with the injured knee close to extension during a sharp deceleration or landing maneuvers. Through video analysis, Ebstrup and Bojsen-Møller (2000) found that ACL injuries occurred during side-stepping or sudden changes in speed. Krosshaug et al. (2007)

found the estimated timing of ACL injuries was 17 to 50 milliseconds after initial foot contact with the ground. The knee flexion angles were less than 20 degrees at the time of injury. Valgus knee collapse happened more often in females compare to males. By using a model-based image matching technique for 10 ACL injury cases, Koga et al. (2010) found the timing of ACL injury was approximately 40 milliseconds after initial foot contact with the ground. During the initial 40 milliseconds of ground contact, decreased knee flexion angles and increases in knee valgus and knee internal rotation angles were observed.

The previously mentioned evidence suggests that ACL injuries usually occur during the impact phase of landing when the knee is close to full extension. Seventeen to 50 milliseconds typically correspond to the first 20% of the landing phase of jumping and cutting tasks. Studies with an aim to understand ACL injury mechanism and prevent ACL injuries should focus on the early phase of landing, cutting, and other maneuvers with sharp decelerations.

2.3.3. Gender

Gender disparity in ACL injury rates have been observed by previous investigators. Females are considered more likely to sustain non-contact ACL injuries compared to males (Prodromos et al., 2007). However, it should be noticed that males actually have greater absolute incidence rate of ACL injury than females (Granan et al., 2008; Gianotti et al., 2009). When the injury rates are normalized by sports exposures, females have greater ACL injury rates than males in most sporting events (Prodromos et al., 2007).

The Norwegian registry data showed that 57% of ACL injured patients were males (Granan et al., 2008). The New Zealand injury compensation data demonstrated that 60% of ACL injured patients were males (Gianotti et al., 2009). The Kaiser-Permanente Southern California data showed that 69% of ACL injured patients were males (Csintalan et al., 2008). Another study that analyzed 8215 cases of ACL injuries from insurance data revealed that 59% of injured patients were males (Shea et al., 2004). The higher absolute incidence rate of ACL injury in males could be caused by more males participating in sports activities.

On the other hand, when the ACL injury rates are normalized to sports exposures, the female-male ratios of ACL injuries are 5:1 for elite team handball, 4.5:1 for high school basketball, 4:1 for collegiate wrestling, 4:1 for collegiate softball / baseball, 3.6:1 for collegiate basketball, 2.8:1 for general population indoor soccer, 2.7:1 for collegiate soccer, 1.9:1 for collegiate rugby, and 1.2:1 for collegiate lacrosse (Agel et al., 2005; Hootman et al., 2007).

2.3.4. Age

ACL injuries are more like to happen at younger age, especially for females. The Norwegian Registry data showed that the number of ACL reconstructions for males were similar from 15 to 34 years of age, while females had the most ACL reconstructions in the 15 to19 years old age group (Granan et al., 2008). The New Zealand injury compensation data demonstrated males had the greatest number of ACL injuries from 20 to 30 years old, while female had the greatest number of ACL injuries from 15 to 30 years old (Gianotti et al., 2009). The Kaiser-Permanente

Southern California database demonstrated that the greatest ACL injury reconstruction number for females occurred in the 14 to17 years of age group, while the reconstruction number for males were similar from 18 to 34 years of age (Csintalan et al., 2008). Shea et al. (2004) analyzed ACL injury claim data for 553 pediatric and adolescent ACL injured patients. ACL injury claim frequency increased at age of 11 years for both genders and reached a peak at 16 to17 years of age.

2.4. ACL Loading Mechanism

An ACL injury occurred when the forces applied on the ACL exceed its maximum loading (Chandrashekar et al., 2006). Previous investigators have demonstrated that proximal tibial anterior shear force, knee valgus / varus moments, knee internal rotation moment, compressive force along the tibia, knee flexion angles, and hamstring force are important ACL loading mechanisms (Berns et al., 1992; Markolf et al., 1995; Markolf et al., 1996; Li et al., 1999; Meyer and Haut, 2005; Meyer and Haut, 2008).

2.4.1. Anterior Shear Force

Anterior shear force applied on the proximal tibia is the major loading mechanism of ACL. Durselen et al. (1995) found that an application of a 140 N quadriceps force significantly increased the ACL strain from 20 to 60 degrees of knee flexion. DeMorat et al. (2004) demonstrated that a 4500 N quadriceps muscle force caused ACL injuries at 20 degrees of knee flexion in *vitro*. Markolf et al. (1995)

recorded the ACL resultant forces when a 100 N of tibial anterior shear force was applied to the cadaver knees from 90 degree to 5 degrees of flexion. The investigators found that the anterior shear force was the most direct loading mechanism of ACL. The ACL resultant force was equal to the anterior shear force when the knee flexion was 30 degrees and increased to 150% of the anterior shear force when the knee was fully extended. Berns et al. (1992) measured the ACL strain when a 200 N tibial anterior shear force was applied to the tibia in *vitro*. The ACL strain was primarily caused by the anterior shear force. ACL strain was positively correlated with anterior shear force at 0 degree and 30 degrees of knee flexions. The ACL strain under a 200 N anterior shear force was 2% at 0 degree of knee flexion and 4.7% at 30 degree of knee flexion. Fleming et al. (2001) tested the effects of tibial anterior shear force on ACL strain in *vivo*. A 130N anterior shear force was applied to the subjects' tibia when the knee was flexed at 20 degrees with and without weight bearing. For both weight bearing and non-weight bearing conditions, ACL strain increased as the anterior shear force increased.

During athletic tasks, the tibial anterior shear force is mainly generated by the quadriceps muscle force and is a result of muscle forces and joint reaction forces. During sudden deceleration tasks, quadriceps can be highly activated to generate landing and braking forces and the large quadriceps force can significantly load the ACL.

2.4.2. Valgus / Varus and Internal / External Rotation Moments

Previous researchers have found that knee valgus, varus, and internal rotation moments significantly contributed to ACL loading when the tibial anterior shear force was applied (Berns et al., 1992; Markolf et al., 1995). In the study of Markolf et al. (1995), an additional load of 10Nm valgus, varus, internal rotation, or external rotation moment was combined with the 100N anterior shear force load. The addition of internal rotation moment to the anterior shear force produced the greatest ACL force, while the addition of external rotation moment actually decreased the ACL force. The addition of varus moments to the anterior shear force increased the ACL force when the knee flexion angle was less than 30 degrees and more than 50 degrees, while the addition of valgus moment to the anterior shear force increased the ACL force when the knee flexion angle was more than 5 degrees. In the study of Bern et al. (1992), valgus / varus moments which range from -20 Nm to 20 Nm and internal / external rotation moments which ranged from -10 Nm to 10 Nm were applied to the cadaver knee. However, neither pure valgus / varus moments nor internal / external rotation moment had significant effects on the ACL strain. In the study of Fleming et al. (2001), valgus / varus moments which ranged from -10 Nm to 10 Nm and internal / external rotation moments which ranged from -9 Nm to 9 Nm were applied to the knee in *vivo*. As the knee internal rotation moments increased, ACL strain increased. However, valgus / varus moments and external rotation moments had very small effects on ACL strain.

A recent study applied dynamic loading during a single leg landing to a knee model to predict ACL strain (Shin et al., 2011). The simulation showed that the peak

ACL strain increased when valgus moments or internal rotation moments increased. In addition, combined knee valgus and internal rotation moments generated greater ACL strain than either alone. However, it should be noticed that during a single leg landing task, the quadriceps is activated and generate an anterior shear force to the tibia. In addition, previous investigators have demonstrated that medial collateral ligament is the primary structure resisting knee valgus moment, and valgus moment is not likely to significantly load the ACL or rupture the ACL until the medial collateral ligament is completely ruptured (Matsumoto et al., 2001; Mazzocca et al., 2003; Shin et al., 2009). Actually, only 6% patients who had ACL injuries completely ruptured their medial collateral ligaments (Fayad et al., 2003). This evidence suggests that the anterior shear force is the major loading mechanism of ACL, while valgus moments should be considered a secondary loading mechanism.

2.4.3. Knee Flexion Angle

The effect of anterior shear force on ACL loading is largely dependent on knee flexion angles. The major mechanics that cause great ACL loading at small knee flexion angle are patella tendon-tibial shaft angle (Nunley et al., 2003), hamstring tendon-tibia shaft angle (Nunley et al., 2003; Lin et al., 2009), and ACL elevation angle (Li et al., 2005) (These angles have been defined in Figure 3.2).

In the study of Markolf et al. (1995), the ACL forces decreased when the knee flexion angles increased under the 100 N anterior shear force load. In the study of Jordan et al. (2007), ACL length was measured during a single-legged lunge task in

vivo. The length of the ACL decreased when the knee flexion angles increased from 30 to 135 degrees.

Markolf et al. (1996) demonstrated that solely extending the knee without other external loading could load the ACL. However, the ACL forces caused by pure knee extension were only approximately 50 N at 0 degree of knee flexion and less than 30 N after 10 degrees of knee flexion. Nunley et al. (2003) showed that the patella tendon-tibial shaft angle increased when the knee flexion angles decreased. With a certain quadriceps force, an increase in patella tendon-tibia shaft angle will increase the anterior shear force. In addition, hamstring tendon-tibia shaft angles decrease when the knee flexion angles decrease (Nunley et al., 2003; Lin et al., 2009). With a certain hamstring force, a decrease in hamstring tendon-tibia shaft angle will decrease the posterior shear force. Li et al. (2005) found that the ACL elevation angles increased as the knee flexion angles decreased. To generate a certain posterior force, a greater ACL elevation angle will require a greater ACL force.

These in *vitro* and in *vivo* studies suggest that knee flexion angle is a key component of ACL loading mechanism. During athletic tasks, small knee flexion can cause great ACL loading. Increasing knee flexion might be an effective technique to reduce ACL loading and prevent ACL injuries.

2.4.4. Compressive Force

Compressive force along the tibia longitudinal axis has been shown to load the ACL through a posterior tibial plateau slope. Because of the posterior tibial

plateau slope, a compressive force along the tibia axis could not be completely counterbalanced by tibiofemoral contact force. The resultant of compressive force and tibiofemoral contact force generates an anterior shear force and load the ACL.

Meyer and Haut (2005) applied repetitive compressive forces to 16 cadaver knees which were flexed at 60, 90, or 120 degrees. ACL ruptures were observed in 14 knees. The peak compressive forces to cause ACL rupture at 60, 90, and 120 degrees of knee flexion were 6 kN, 5 kN, and 4.5 kN. Meyer and Haut (2008) also demonstrated that a mean of 5.4 kN peak compressive force caused ACL ruptures at 30 degrees of knee flexion. By increasing the posterior tibial plateau slope from 8.8 to 13.2 degrees, Giffin et al. (2004) demonstrated that the osteotomy caused a significant anterior tibial translation under a 200 N compressive force loading. Hohmann et al. (2011) showed that female patients with ACL injuries (6.7 degrees) had significantly greater posterior tibia plateau slopes than the uninjured individuals (5 degrees).

The previously mentioned studies suggest that compressive force with a posterior tibia plateau slope should be considered an important loading mechanism for ACL. A large posterior tibia plateau slope might be a risk factor for ACL injuries. During athletic tasks, the large impact ground reaction force can generate a great compressive force and load the ACL. Soft landing, which dissipates the impulse in a long period of time, can decrease the impact ground reaction force and might protect the ACL.

2.4.5. Hamstring force

The effects of hamstring force on ACL loading have been investigated in *vitro* (Li et al., 1999), in *vivo* (Beynnon et al., 1995), and by using computer simulation (O'Connor, 1993). Li et al. (1999) investigated the effects of hamstring co-contraction with quadriceps on ACL force in *vitro*. An isolated 200 N quadriceps load was applied to the cadaveric knee at different knee flexion angles. An addition of an 80 N hamstring load was then applied with the quadriceps load. The investigators found that the addition of hamstring load decreased the ACL force by approximately 15, 30, 43, and 44% at 0, 15, 30, and 60 degrees of knee flexion. However, the co-contraction of hamstring did not reduce ACL loading at 90 and 120 degrees of knee flexion. In a later study, Li et al. (2004) measured the ACL force of cadaveric knee in response to a 400 N quadriceps force, a 200 N hamstring force, or a combined 400 N quadriceps force and a 200 N hamstring force at different knee flexion angles. The investigators found that compared to isolated quadriceps force loading, the addition of hamstring force decreased ACL force at 0 and 30 degrees of knee flexion, but did not decrease ACL force when the knee flexion was more than 60 degrees (Li et al., 2004). The findings of these two studies suggest that with a constant quadriceps force, the application of a hamstring force to the knee could decrease the ACL force when the knee flexions are less than 60 degrees. However, simply adding a hamstring force that produces flexion moment without increasing the quadriceps force to obtain the resultant moment needed might not represent true hamstring co-activation and should raise attention.

Beynnon et al. (1995) measured the ACL strain in *vivo* during selected exercise. The ACL strain was measured in a relax position or different isometric contractions with quadriceps and hamstring muscles. The investigators found that isometric quadriceps contraction increased ACL strain compared to relax condition at 15 and 30 degrees of knee flexion, but not at 60 and 90 degrees of knee flexion. Co-contraction of quadriceps and hamstring increased ACL strain compared to relax condition at 15 degrees but not at 30, 60, and 90 degrees of knee flexion. This study suggests that the effects of hamstring force on ACL loading is largely dependent on knee flexion angles.

O'Connor et al. (1993) used a computer-based model to study the effects of hamstring co-contraction on ACL force. Previous in *vitro* study usually applied the hamstring force to a constant quadriceps force to study the co-contraction (Li et al., 1999). However, in the simulation done by O'Connor et al. (1993), the quadriceps force was changing according to changes in hamstring force to generate an extension moment to counterbalance the flexion moment of the hamstring. The simulation showed that the co-contraction of quadriceps and hamstring actually increased the ACL force when the knee flexion angle was less than 22 degrees. The protective effect of the hamstring to ACL only occurred when the knee flexion angle was more than 22 degrees. Yu and Garrett (2005) also used computer simulation to investigate the effects of hamstring co-contraction on ACL loading during dynamic movement. In the model, quadriceps and hamstring muscle forces were varied to balance the peak knee extension moment during a stop-jump task. The authors found that the hamstring co-contraction increased ACL loading when the knee

flexion angle were less than 15° for males and 20° for females. The discrepancies between in *vitro* studies and simulation studies suggested that different definitions of quadriceps and hamstring co-contraction might lead to different outcomes.

In summary, the addition of a hamstring force to a constant quadriceps force is likely to decrease ACL loading because of a decrease in anterior shear force. However, if the quadriceps is changing according to the hamstring force to maintain the joint resultant moments, the ACL loading might increase at low knee flexion angles because of an increase in anterior shear force.

2.5. ACL Injury Risk Factors

Knowledge regarding risk factors is crucial for the prevention of ACL injuries. A large number of studies have been conducted to identify these risk factors and qualify their roles in ACL injuries. A variety of external and internal factors have been investigated.

2.5.1. External Factors

A number of external factors have been investigated including type of competition (Myklebust et al., 1998), shoe / surface interface (Lambson et al., 1996; Olsen et al., 2003; Drakos et al., 2010), weather (Orchard and Powell, 2003), and knee bracing (Rishiraj et al., 2009).

Myklebust et al. (1998) followed 24 elite Norwegian team handball teams for three years. The injury rate was greater during competition (0.91 injuries / 1000 hours) than during practice (0.03 injuries / 1000 hours).

Olsen et al. (2003) recorded 53 ACL injuries during Norwegian team handball regular league games from 1989 to 2000. Among the 44 injuries in females, 8 of the injuries happened on wooden floors and 33 of the injuries occurred on artificial floors. Lambson et al. (1996) demonstrated that high school football players wearing Edge cleat design which had the greatest torsion resistance had a greater injury rate than the players wearing non-Edge designs. The mechanics of high injury rate on artificial floors could be caused by the great shoe / surface friction. Biomechanical studies also found that high shoe / surface frictions were associated with greater knee loading. During a side-cutting task, subjects demonstrated less knee flexion angles, less knee extension moments, and greater knee valgus moments on the high friction surface compared to the low friction surface (Dowling et al., 2010). It is postulated that a great shoe / surface friction is associated with a great posterior ground reaction force which can load the ACL (Yu et al., 2006; Yu and Garrett, 2007).

Orchard et al. (2003) found that in professional football teams, ACL injuries were more likely to occur in cold weather. The investigators suggested that the cold weather might be associated with low shoe / surface friction. However, other studies generally suggest ACL injuries are more likely to happen at high shoe / surface friction condition. Therefore, the cold weather might contribute to a high ACL injury rate with other mechanics.

The effect of using a brace on ACL injury rate is not conclusive. A recent review has summarized the literature of bracing effects on knee mechanics and injuries (Rishiraj et al., 2009). The investigators suggested that functional knee brace could provide external protections to the knee; however, compliance is low in non-injured athletes because of the fear of performance hindrance.

2.5.2. Anatomical and Hormonal Factors

The anatomical and hormonal factors are not likely to be changed without medical interventions. The anatomical and hormonal factors for ACL injuries include lower extremity alignment (Moul, 1998), femoral intercondylar notch width (Shelbourne et al., 1998; Uhorchak et al., 2003; Simon et al., 2010), joint laxity (Huston and Wojtys, 1996; Uhorchak et al., 2003), posterior tibial plateau slope (Stijak et al., 2008; Simon et al., 2010; Hohmann et al., 2011), intrinsic ACL material properties (Chandrashekar et al., 2005; Chandrashekar et al., 2006), patella tendon-tibia shaft angle (Nunley et al., 2003), and hormonal variation (Wojtys et al., 1998; Zazulak et al., 2006).

Moul (1998) found less Q-angle in male collegiate basketball players when compared to female players. Huston and Wojtys (1996) found that female athletes had greater anterior tibia laxity compared to male athletes.

Shelbourne et al. (1998) compared the intercondylar notch width between male and female ACL injured patients. Males had greater notch width than females after normalized for body height. In addition, the patients who had a less than 15 mm notch width were more likely to injury their contralateral side compared to the

patients who had a greater than 16mm notch width. In a prospective cohort study (Uhorchak et al., 2003), 859 West Point cadets were tested and followed for 4 years. Compared to the non-injured cadets, the injured cadets had less intercondylar notch width and greater generalized joint laxity. A less intercondylar notch width could put the ACL at greater risk of impingement which elongates the ACL as it wraps around the notch surface (Park et al., 2010).

Hohmann et al. (2011) examined the posterior tibial plateau slope in 272 patients with ACL injuries and a control group using radiograph. The injured females had greater posterior tibia plateau slope than the females in the control group. By using MRI technique, Stijak et al. (2008) found that patients with ACL rupture had greater posterior tibia plateau slope of the lateral condyle compared to ACL intact individuals. Simon et al. (2010) compared the medial and lateral tibial slopes as well as intercondylar notch width between an ACL injured group and a control group. The ACL injured group had greater lateral tibial plateau slope and less notch width than the control group. Therefore, the tibial plateau slope of lateral condyle and medial condyle might need to be considered separately.

Chandrashekar et al. (2005) investigated the gender differences in the anthropometric characteristics of ACL in *vitro*. Females' ACL were less in cross-sectional area, length, volume, and mass compared to males'. No gender differences were found in intercondylar notch geometry or ACL mass density. Chandrasheka et al. (2006) also compared the mechanical properties of ACL between genders. Females' ACL had lower elongation, strain, load, stress, energy

absorbed, and strain energy density at failure compared to males'. Females' ACL also had lower stiffness and modulus of elasticity compared to males'.

Nunley et al. (2003) compared the patellar tendon tibial shaft angles as a function of knee flexion angles between genders. The authors found that females' average patellar tendon tibial shaft angle was 3.7° greater than males'. The gender differences in patellar tendon tibial shaft angle can increase the anterior shear force on the tibia by approximately 13%.

Wojtys et al. (1998) observed the relationships between menstrual circle and ACL injuries in 28 females. More ACL injuries occurred during the ovulatory phase than expected. In addition, there were fewer ACL injuries during the follicular phase than expected. A Meta-analysis study (Zazulak et al., 2006) demonstrated that although the menstrual cycle was associated with anterior-posterior laxity of the knee; however, it was unknown if ACL injuries were more likely to happen during certain phase of a menstrual cycle.

2.5.3. Neuromuscular Control

Neuromuscular control is another internal risk factor for ACL injury. However, neuromuscular control can be modified through appropriate training program. Considering the gender bias in ACL injury rates, many studies have attempted to identify ACL injury risk factors by comparing movement patterns between male and female athletes during athletic tasks. In those studies that compared gender differences, jump landing and cutting tasks were commonly used to evaluate neuromuscular control patterns.

Jump landing tasks are the most commonly used tasks to evaluate neuromuscular control patterns. Decker et al. (Decker et al., 2003) showed that female recreational athletes had less knee flexion at initial contact but increased knee and ankle range of motions during a drop landing task compared to males. Females also utilized knee joint motion to absorb more energy than males. Nagano et al. (2007) found that female basketball and tennis players had greater tibial internal rotations compared to males during a drop landing task. In addition, quadriceps / hamstring EMG ratio for the 50ms time period before foot contact was greater in females than males. No gender difference was observed in knee flexion, varus, valgus, and tibial anterior translation. However, it should be noticed that in these two studies, the subjects simply landed from a drop height without a consecutive jump. The loading imposed on the lower extremities might not be as great as a landing with a consecutive jump.

Salci et al. (2004) compared the landing patterns between male and female volleyball players during spike and block landing. Female volleyball players had less knee and hip flexion angles during the 40cm landing. Ford et al. (2003) showed that female basketball players landed with greater total valgus motions and maximum valgus angles than male players during a drop vertical jump task. Yu et al. (2006) showed that female recreational athletes demonstrated less knee flexion angles and angular velocities, less hip flexion angles and angular velocities, increased impact ground reaction forces, and increased tibial anterior shear forces during a stop-jump task compared to males. Yu et al. (2005) also demonstrated that female adolescent soccer players had decreased knee and hip flexion angles at initial contact and

decreased knee and hip flexion range of motions during a stop-jump task compared to male players. Chappell et al. (2002) found that female recreational athletes had greater tibial anterior shear forces, greater knee extension moments, and greater knee varus moments during stop-jump tasks than males. Chappell et al. (2007) also found that female recreational athletes had decreased knee and hip flexions, increased quadriceps activations, and decreased hamstring activations before the landing of a stop-jump task compared to males. As shown in the previously mentioned studies, the gender differences become more pronounced during the landing with a consecutive jump compared to a simple drop landing.

Cutting tasks were also commonly studied in previous studies. Malinzak et al. (2001) showed that female recreational athletes had less knee flexion angles, more knee valgus angles, greater quadriceps activations, and less hamstring activations during running and cutting tasks compared to males. McLean et al. (2004) found that females had decreased hip and knee flexions, hip and knee internal rotations, and hip abductions but increased knee valgus during a site-cutting task with defensive players when compared to males. Mclean et al. (2005) also demonstrated that female collegiate basketball players had greater peak knee valgus and less peak hip and knee flexions during side-step, side-jump and shuttle-run tasks when compared to males. Landry et al. (2007) found that female adolescent players had decreased hip flexion angles and hip extension moments as well as increased gastrocnemius and rectus femoris activations during an un-anticipated side-cutting task compared to males. Pollard et al. (2007) found that female collegiate soccer players had grater hip internal rotations and less hip flexions during the early phase of a cutting task

compared to males. Sigward and Powers (2006b) found that female collegiate soccer players had greater external knee adduction moments and small external knee flexion moments as well as greater quadriceps activations during the early phase of a side-cutting task compared to males.

Hewett et al. (2005) conducted a prospective cohort study to identify the risk factors of ACL injury for female athletes. 205 young female athletes participated in a drop vertical jump task and were followed for 13 month. Nine athletes had non-contact ACL injuries during follow up period. The injured athletes demonstrated greater initial and maximum knee valgus angles, less maximum knee flexion angles, greater peak external hip flexion moments, and greater peak ground reaction forces during the landing phase as well as greater peak external knee valgus moments and less stance time during the stance phase compared to the non-injury subjects. These findings are similar to those reported in the many previous studies on gender differences in lower extremity movement patterns.

In summary, previous investigators have demonstrated that females have a restricted sagittal plane motion and an increased motion in the frontal and coronal planes when performing athletic tasks. The findings of motion analysis studies are consistent with ACL loading mechanism which suggests the biomechanical movement patterns demonstrated by females generate a greater ACL loading than males. Movement training programs that increase lower extremity sagittal plane motion and decrease non-sagittal plane motion might decrease ACL loading and prevent ACL injuries.

2.6. Performance and ACL Loading

During actual competition, achieving great performance is important for athletes. From the injury prevention perspective, reducing ACL loading or loading factors are important. On the other hand, from the performance perspective, fast running speed, high jump height, short take-off time, and low energy expenditure are desirable in most sporting events. However, while increasing performance and decreasing ACL loading are both important for athletes, the underlying relationships between them are largely unknown.

2.6.1. The Effects of Performance Demands on Lower Extremity Biomechanics

Previous studies have focused on the effects of drop heights on lower extremity biomechanics during drop landing and drop vertical jump tasks. Although ACL loading was not directly measured or estimated, the findings of previous studies provided some insight into the changes in ACL loading caused by different task demands.

McNitt-Gray (1993) studied lower extremity kinetics during landing from three drop heights (0.32, 0.72, 1.28m). The peak vertical ground reaction force increased as the drop heights increased. In addition, subjects landed with less initial knee, and hip flexion, but increased flexion velocities and range of motions as the drop heights increased. The joint moments and mechanical work also increased as the drop heights increased. In the study by Zhang et al. (2000), subjects landed from three drop heights (0.3, 0.6, 1m). The vertical ground reaction forces associated with toe-

touch and heel contact increased as the drop heights increased. In addition, the joint range of motions, power and work increased as the drop height increased. Dufek and Bates (1990) studied the effects of drop heights (0.4, 0.6, 1m) and drop distances (0.4, 0.7, 1m) on ground reaction forces during drop landing tasks. As the drop heights and distances increased, the ground reaction forces associated with toe and heel contact increased. Yeow et al. (2009) used regression analysis to study the effects of drop heights (0.15, 0.30, 0.45, 0.60, 0.75, 0.90 and 1.05 m) on peak ground reaction forces during drop landing tasks. The regression equation indicated that peak ground reaction forces had a positive exponential relationship with drop heights.

In the study by Bobbert et al. (1987), subjects conducted drop vertical jumps with maximum jump height from three drop heights (0.2, 0.4, 0.6m). As the drop height increased, the peak ground reaction forces increased. In addition, during landing phase, the ankle and hip range of motions, joint moments, power, and work increased as the drop heights increased. In the study of Ball et al. (2010), subjects conducted drop vertical jumps from three drop heights (0.2, 0.4, and 0.6m). As the drop heights increased, the peak ground reaction forces and stance time increased, and the time to peak ground reaction force decreased. In the study of Peng (2011), subjects performed drop vertical jumps from five drop heights (0.2, 0.3, 0.4, 0.5, 0.6m). As the drop heights increased, the peak vertical ground reaction forces, landing impulses, and landing time increased, but the time to peak ground reaction force decreased. The initial knee flexion angles and knee stiffness decreased and knee range of motions increased as the drop heights increased. The negative work

done by knee and ankle increased as the drop heights increased, but the active work stayed similar as the drop heights increased.

Walsh et al. (2004) studied the effects of drop heights as well as jump speeds on lower extremity biomechanics. The subjects were instructed to jump as high as possible or jump with a shorter stance time with three drop heights (0.2, 0.4, and 0.6m). As the drop heights increased, the peak vertical ground reaction forces increased. When the stance time decreased, the take-off velocities and joint work decreased and the maximum vertical ground reaction forces increased.

The previously mentioned studies suggest that during drop landing and drop vertical jump tasks, increases in drop heights and distances will increase vertical ground reaction forces and decrease initial knee flexion angles. In addition, the vertical ground reaction force will increase when the individuals jump with a shorter stance time. Although the focus of previous studies was not ACL loading, these findings suggest that ACL loading is likely to increase if the height of drop before a landing is increased or the support phase of a jump is decreased. However, simply landing from a drop might not necessarily represent a hazardous scenario for ACL injuries. Compared to drop landing and drop vertical jump, the landing phase of a stop-jump and a side-cutting task involve sudden deceleration in the anterior and posterior direction and are more likely to generate a large loading on the ACL. However, it is unknown how different performance demands such as jump heights and stance time affect the ACL loading during these athletic tasks.

2.6.2. The Effects of Changes in ACL Loading on Performance

The changes in ACL loading are generally induced by changes in subjects' techniques when performing athletic tasks. A commonly used strategy to reduce ACL loading is a soft landing with increased knee flexion. However, while many previous studies have focused on the effects of interventions on ACL loading, reports of the effects of interventions on performance are lacking.

Cronin et al. (2008) studied the effects of technique instructions on ground reaction forces during the landing of a volleyball spike jump. The technique instructions included landing on the forefoot, knee over toes, flexing knees before the landing, and increased knee flexion during the landing. The authors found that the changes in techniques decreased peak vertical ground reaction force by 24%. Onate et al. (2001) evaluated the effects of augmented feedback on ground reaction forces during the landing following a maximum vertical jump. The augmented feedback was given through video analysis of landing techniques. Landing with forefoot, normal varus / valgus, and increased flexion were confirmed during the analysis. The augmented feedback group decreased their peak vertical GRF by 18% immediately following the feedback and decreased their peak vertical GRF by16%1 week following the feedback. McNair et al. (2000) showed that landing techniques with landing on forefoot and increasing knee flexion before landing decreased peak vertical ground reaction forces during a drop landing task. Prapavessis et al. (1999; 2003) found that instructions for a soft landing decreased peak ground reaction forces during a drop landing task in children and high school students. Cowling et al. (2003) showed that increasing knee flexion during landing decreased peak vertical

and posterior ground reaction forces during a single leg landing task. Podraza and White (2010) studied the effects of initial knee flexion angles on lower extremity biomechanics during single leg drop landing tasks. When the initial knee flexion increased during landing, the peak vertical and posterior ground reaction forces decreased. The previously mentioned researchers have identified that soft landing with increased knee flexions was effective in decrease peak ground reaction forces which might be associated with decreased ACL loading. However, as we noticed, no performance variable such as landing time or mechanical work was reported.

Some studies have documented some changes in performance as a result of changes in landing techniques. However, most studies only used jump height as a performance variable. Zhang et al. (2000) studied the effects of landing techniques (soft, normal, and stiff) on lower extremity biomechanics when subjects landed from different drop heights. The soft landing decreased peak ground reaction forces and increased knee and hip joint range of motion. The soft landings also increased the eccentric work at knee and hip. However, the change in landing time was not reported. Onate et al. (2005) found self or combination video feedback increased knee range of motion and decreased peak vertical ground reaction forces. The jump heights were used as a covariate during the analysis. However, the changes in contact time and mechanical work were not reported.

A few investigators reported stance time as a performance parameter. Devita and Skelly (1992) studied the effects of landing stiffness on lower extremity kinetics. The soft landing had a maximum of 117 degrees of knee flexion and stiff landing had a maximum of 77 degrees of knee flexion. The soft landing decreased

the peak vertical ground reaction force. However, the soft landing increased the knee and hip work as well as the total lower extremity work during the impact phase. In addition, the time to second peak vertical ground reaction force was longer for the soft landing compared to the stiff landing. Mizner et al. (2008) found that soft landing decreased peak vertical ground reaction forces, peak knee abduction angles, peak external knee abduction moments, and increased peak knee flexion angles. No changes were observed in jump heights, but the landing time increased. The change in mechanical work was not reported. Myers and Hawkins (2010) investigated the effects of alterations to techniques on ACL loading and performance during a stop-jump task. The verbal instructions included increasing the amplitude of the jump prior to landing, increasing the amount of knee flexion at landing, and striking the ground with the toes first. The anterior tibial shear forces were estimated through an EMG driven model. The changes in technique increased knee flexion angles and decreased anterior tibial shear force. In addition, the subjects increased their jump heights and maintained their approach speeds and contact time after the modifications in techniques. However, the change in mechanical work was not reported. In addition, a drawback of this study was that the testing order was not randomized.

Walsh et al. (2007) studied the effects of instructions of soft landing on peak ground reaction forces, contact time, and flight time during a drop vertical jump task in basketball players. Different from previous studies, the investigators found that the instructions of soft landing had no effect on peak ground reaction force, contact time and flight time for males. However, instructions of soft landing decreased the peak

ground reaction forces and increased the stance time but had no effects on flight time for females. Although the number of subjects in the instruction group was only 6 and an increase in sample size might demonstrate significant differences, this study suggests the effects of instruction on performance could be inconsistent across genders.

In summary, previous studies have demonstrated that changes in techniques such as soft landing with increased knee flexion are effective in decreasing ACL loading. However, the changes in the performance were largely unknown in each study. The combined results of previous studies suggest that decreases in ACL loading induced by soft landing are likely to increase stance time and mechanical work and these changes are associated with a decrease in performance.

2.6.3. Training Effects on Performance and ACL Loading

Training programs have been developed based on existing knowledge in an attempt to alter lower extremity biomechanics and reduce ACL loading. Previous investigators have found that training could improve lower extremity biomechanics as well as improve performance. However, in many studies, the evaluations of lower extremity biomechanics and performance were conducted during two different tasks without considering the potential relationships between them. In addition, usually jump height was the only performance variable that was reported.

Myer et al. (2005) showed that a neuromuscular training program increased knee range of motion and decreased varus and valgus moments during a drop vertical jump tasks in female athletes. In addition, increases in jump heights were

observed during a maximum vertical jump task. It should be noticed that the drop vertical jump test and the maximum vertical jump test were conducted separately. It was unknown if the training had effects on jump heights during the drop vertical jump or if the training had effects on lower extremity biomechanics during the maximum vertical jump task. Myer et al. (2006a; 2006b) compared the effects of plyometric training with balance training on athletes' lower extremity biomechanics and performance. Both training methods reduced hip adduction angles and maximum ankle eversion angles during a drop vertical jump. Both groups decreased knee abduction angles during a drop landing. Plyometric training increased knee flexion during the drop vertical jump, while the balance training increased knee flexion during the drop landing. The plyometric group increased the peak vertical ground reaction forces during the drop landing, while the balance training group decreased the peak vertical ground reaction force during the drop landing. Increases in jump heights were observed during a maximum vertical jump. However, the changes in jump heights, stance time, and mechanical work during the drop vertical jump or drop landing were not reported. Chappell and Limpisvasti (2008) found that a neuromuscular training program decreased dynamic knee valgus during a stop-jump task in collegiate basketball and soccer players. The training also increased initial knee flexion and maximum knee flexion during a drop jump task. The peak vertical ground reaction forces and contact time were not different between pre-training and post-training during both jump tasks. Increases in vertical jump heights were observed during a maximum vertical jump. Similar to previous mentioned study, one limitation of this study was that the lower extremity biomechanics and jump height

were tested using two different tasks. Testing biomechanical ACL loading factors and maximum jump height using two different tasks might favor one aspect without considering the other.

Actually, by using a single jump landing task to evaluate both lower extremity biomechanics and jump height simultaneously, several investigators did not find improvements in both lower extremity biomechanics and jump height. Grandstrand et al. (2006) investigated the effects of a warm-up program on knee separation and jump height during a drop vertical jump task. Knee separation was defined as the linear distance between left and right patellae. Different from previously mentioned studies, the knee separation and jump height were assessed during the same task. No significant difference was found in knee separation or jump height before and after the training. Vescovi et al. (2008) evaluated the effects of a plyometric program on both ground reaction forces and jump height during a maximum vertical jump task. The training had a tendency to decrease the peak vertical ground reaction forces, however, no changes were observed in jump heights. Lim et al. (2009) investigated an injury prevention training program on lower extremity biomechanics as well as jump height during a rebound jump tasks. Training increased the knee flexion angles, inter-knee distances, and decreased maximum knee extension moments in the training group. However, no improvement was observed in jump height.

Hewett et al. (1996) studied the effects of a plyometric training program on lower extremity biomechanics and jump height during a volleyball block jump task. After the training, the peak landing forces and peak adduction and abduction

moments decreased during the block jump task. The jump height during the block jump tasks did not change, but the jump height during a maximum jump test increased. The results of this study support the notion that if we consider jump height during the block jump task as the performance parameter, there was no improvement in jump height. However, if we consider jump height during the maximum vertical jump test as the performance parameter, there was an improvement. It is possible that during the block jump task, subjects landed in a way to decrease knee loading with a compromise in jump height and stance time. Therefore, an improvement was observed in lower extremity biomechanics but not in jump height. However, during the maximum vertical jump test, subjects landed in a way to maximize the jump height and therefore an improvement in jump height was observed. However, the changes in lower extremity biomechanics during the maximum vertical jump test were not reported.

Discrepancies have been observed among previous studies in training effects on performance and ACL loading. The discrepancies might be caused by different training programs and characteristics of subjects. However, the differences in testing protocols should also be noticed. Many studies considered performance and ACL loading as two independent factors and have them tested during two different tasks. A lack of consideration of the relationships between performance and loading could contribute to the discrepancies among previous studies. Therefore, examining the relationships between performance and ACL loading will provide important implications to future development of movement evaluation tests and criteria for

reporting intervention effects on movement patterns in ACL injury prevention programs.

Summary

ACL injuries are common sports related injuries that have long-term hazardous effects on people's quality of life (Ingersoll et al., 2008). It is important for us to understand ACL injury mechanisms as well as develop effective and efficient training programs to prevent ACL injuries. Previous motion analysis and cadaver simulation studies have well documented the biomechanical risk factors for ACL injuries and ACL loading mechanism. Small knee flexion angles with a large quadriceps force and a large impact ground reaction forces are the major loading mechanisms for ACL injury. Although previous studies have documented some relationships between performance and ACL loading, no study has systematically studied their relationships simultaneously. There is a need to understand the relationships between performance and ACL loading for understanding injury mechanisms and developing injury prevention strategies.

CHAPTER III

METHODS

3.1. Subjects

A minimum of 34 recreational athletes (18-35 years) was needed for the current study. The subjects needed to have experience in playing sports that involve jump landing and cutting tasks (for example: basketball, volleyball, soccer, team handball, rugby). The subjects needed to be physically active and participated sports / exercise at least two times per week for a total of 2-3 hours. Sports experience was defined as currently playing sports at least 1 time per week or having previously played in high school / college / club levels. A subject was excluded if he / she (1) had no experience in playing sports that involve jump landing and cutting tasks. (2) was not physically active. (3) had an ACL injury or other major lower extremity injuries. (4) had a lower extremity injury that prevented participation in physical activity for more than 2 weeks over the previous 6 months, (5) possessed cardiovascular, respiratory, neurologic, or other conditions that prevent him / her from participating at maximal effort in sporting activities, (6) was pregnant. Subjects were recruited using a variety of techniques including word of mouth, recruitment fliers, mass recruitment email, and class recruitment.

3.2. Instrumentation

Eight video cameras (Peak Performance Technology, CO, USA) were used to collect three dimensional coordinates of reflective markers placed on the subjects at a sampling rate of 120 frames / s. Two Bertec 4060A force plates (Bertec Corporation, OH, USA) were used to collect ground reaction forces and moments at a sampling rate of 1200 samples / s. The maximum dynamic load capacities of the force plates were 20 kN in the vertical direction and 10 kN in the anterior-posterior and medial-lateral directions.

3.3. Experimental Procedure

The experiment was conducted in the Motion Analysis Lab of the Center for Human Movement Science located in the basement of Bondurant Hall at the University of North Carolina at Chapel Hill. The subjects came to the lab once for data collection. The duration for one data collection was approximately 1.5 - 2 hours.

After the subject came to the lab, informed consent forms were described to the subject. If the subject agreed to participate in the study, the subject signed the informed consent forms. The subject's date of birth, sports experience, and injury history were recorded. The subjects completed the International Physical Activity Questionnaire (Craig et al., 2003) and answered additional questions to record the type, frequency, and duration of their current sports / exercise activities. The subject's height and weight were measured. The subject's dominant leg was

determined based on preferred leg for push-off to jump for a further distance. The attire for the subjects included spandex shorts, spandex shirts (provided by the lab), and athletic shoes (provided by each subject). The subject conducted a warm-up protocol including stretching and self-selected over ground running for 5 minutes.

FIGURE 3.1 Anterior and posterior views of marker placements

For the static standing trial (Figure 3.1), retroreflective markers were attached bilaterally on the spinous process of acromioclavicular joints, posterior superior iliac spines (PSIS), anterior superior iliac spines (ASIS), greater trochanters, medial and lateral femoral condyles, tibial tuberosity, lower shank (approximately 2/3 between knee and ankle centers), medial and lateral malleolus, first and fifth metatarsal heads, first toes, and heels. During the static standing trial, the subject was instructed to raise their arms and stand with feet shoulder width apart and pointing forward.

For the stop-jump and cutting trials, six markers including bilateral medial femoral condyles, medial malleolus, and first metatarsal heads markers were removed. The subject conducted a vertical stop-jump in five conditions. A vertical stop-jump task consisted of an approach run followed by a 1-footed take-off, a 2-footed landing, and a 2-footed take-off (Chappell et al., 2002). During all the stop-jump conditions, the subjects were instructed to approach as fast as possible.

During the first condition, the subject was instructed to jump as high as possible following the 2-footed take-off. During the second condition, the subject was instructed to jump as fast as possible during the 2-footed landing while still trying to jump as high as possible. During the third condition, the subject was instructed to land softly during the 2-footed landing while still trying to jump as high as possible. During the fourth condition, the subject was instructed to increase the knee flexion at the initial contact of the 2-footed landing while still trying to jump as high as possible. During the fifth condition, the subject jumped with 60% of maximum jump height. For the fifth condition, before the data collection, the subject practiced the 60% of

maximum jump height using a Vertec (Sports Imports, Columbus, OH) until they reported feeling comfortable to jump to the targeted jump height. However, the Vertec instrument was be used during the testing to ensure consistency among conditions. We chose 60% of maximum jump height to study the effects of decease in jump height on ACL loading. A plot study with 6 subjects had demonstrated that subjects usually took 3-5 practice trials to be comfortable to reproduce the target jump height. The pilot study had shown that subjects were able to reproduce the jump height with a mean of 60% and a standard deviation of 4%of maximum jump height during this condition.

The subject also conducted a side-cutting task with the dominant leg for four conditions. A side-cutting task included an approach run followed by a 1-foot landing and a lateral cut at 45 degrees from the running direction. Pieces of adhesive tape were placed on the ground as a visual target for cutting.

During the first condition, the subject was instructed to run as fast as possible and cut as fast as possible. During the second condition, the subject was instructed to land softly during the 1-foot landing while still trying to run as fast as possible and cut as fast as possible. During the third condition, the subject was instructed to increase the knee flexion at the initial contact of the 1-footed landing while still trying to run as fast as possible and cut fast as possible. During the fourth condition, the subject was instructed to cut with 60% of maximum running and cutting speed. For the fourth condition, before the data collection, a timer was used to help the subject find the pace of 60% of maximum running and cutting speed from the start to the end of the running and cutting. The reason to choose of 60% of maximum cutting

speed was to select a speed that was significantly less than the maximum cutting speed to study the effects of decease in cutting speed on ACL loading. A plot study had demonstrated that subjects usually took 3-5 trials to be comfortable to reproduce the target cutting speed. The plot study had demonstrated that the subjects were able to reproduce the approach speed with a mean of 66% and a standard deviation of 4% of maximum approach cutting speed during this condition. The subjects were also able to reproduce the take-off speed with a mean of 64% and a standard deviation of 7% of maximum approach cutting speed during this condition.

The subject had 5 practice trials for each stop-jump and cutting condition. Five official trials were collected for each stop-jump and cutting condition to ensure that at least three good trials were collected. A trial was excluded if the subject did not meet the demands of the task, the subject's feet were not on the correct force plate, or markers were not properly tracked during data collection. For each subject, the testing order of stop-jump task and cutting task was randomized. The testing order of different conditions within stop-jump and cutting task was also randomized. Subjects were given a 3 minute break between each stop-jump and cutting condition and 30 seconds between trials to avoid fatigue.

For both stop-jump and side-cutting, soft landing and landing with increased knee flexion at initial contact were separated as two testing conditions because a pilot study with 5 subjects had demonstrated that subjects showed different movement patterns during these two conditions (Table A.1; A.2). Subjects had more

knee flexion during the landing with increased knee flexion condition than the landing softly condition.

A pilot study with 6 subjects had shown that subjects usually took 2-3 practice trials for jumping for maximum jump height, jumping fast, and cutting with maximum speed conditions and 3-5 practice trials for the other conditions to consistently reproduce movement patterns. In this pilot study, because the landing softly with increased knee flexion condition had not been separated as two conditions, only four conditions had been tested for the stop-jump and three conditions had been tested for the side-cutting. The consistency of movement patterns were assessed using coefficient of multiple correlation (CMC) (Kadaba et al., 1989) and averaged standard deviation. The pilot study had demonstrated that after practice subjects were able to produce consistent movement patterns during relatively novel tasks including soft landing, jumping for 60% of maximum height, and cutting with 60% of maximum speed conditions as compared to familiar tasks including jumping for maximum height and cutting for maximum speed conditions (Table A.3; A.4; A.5; A.6). The results of the pilot study suggested that although some tasks such as landing with increased knee flexion and jump with 60% of maximum jump height might be relatively novel for subjects, after practice subjects were able to reach consistent movement patterns during these relatively novel tasks compared to less novel tasks such as jump as high as possible and jump as fast as possible.

To randomize the testing order, each testing condition was corresponding to a drawing number. For each drawing number, a random numbers between 0 and 1 was generated. After the random numbers had been generated, the rank of the

random numbers with a descending order was calculated. The testing order was determined as a combination of drawing numbers and the rank of the random numbers. For example, as shown in Table 3.1 and 3.2, the conditions of jumping for maximum height, jumping fast, jump with increased knee flexion, and jumping for 60% of maximum height were corresponding to drawing numbers of 1, 2, 3, and 4. One random number between 0 and 1 was generated for drawing number. The rank of the random number with a descending order was 2 – 3 – 4 – 1. Therefore, the condition corresponding to drawing number 1 which was jumping for maximum height was tested as second condition. The condition corresponding to drawing number 2 which was jumping fast was tested as third condition. The condition corresponding to drawing number 3 which was jump with increased knee flexion was tested as forth condition. The condition corresponding to drawing number 4 which was jumping with 60% of maximum height was tested as first condition.

Table 3.1. Drawing number of testing conditions

Testing conditions	Drawing number
Jump Max Height	1
Jump Fast	2
Increased Flexion	3
Jump 60% Max Height	4

Table 3.2. Generating random number and testing order

Drawing number	Random number	Rank of random number	Testing order
1	0.842	2	1. Jump 60% Max Height
2	0.823	3	2. Jump Max Height
3	0.179	4	3. Jump Fast
4	0.857	1	4. Increased Flexion

3.4. Data Reduction

3.4.1. Kinematics and Kinetics

The coordinates data and force plate data were time-synchronized to 1200 Hz using a linear interpolation method. The coordinate data were filtered using a fourth–order, zero-phase-shift Butterworth filter at an estimated optimum cut-off frequency of 10 Hz (Yu et al., 1999). The force plate data were filtered using a fourth–order, zero-phase-shift low pass Butterworth filter at a frequency of 200 Hz. The toe-touch event was defined as the first frame the vertical ground reaction forces exceeding 20 N. The toe-off event was defined as the first frame the vertical ground reaction forces was less than 20 N. The stance phase was defined as the phase between the toe-touch and toe-off events. The landing phase was defined as the phase between toe-touch and maximum knee flexion events.

Lower extremity joints kinematics and kinetics were calculated during the stance phase of stop-jump and cutting tasks. The hip joint center was defined by Bell's methods (Bell et al., 1990). Bell's methods defined the pelvis coordinate using bilateral ASIS and the middle point of PSIS. The hip center was located 14% of the inter-ASIS distance medially, 30% distally, and 22% posterior to the ASIS. The knee joint center was defined as the midpoint between the medial and lateral femoral condyles. The ankle joint center was defined as the midpoint between the medial and lateral malleoli. The medial femoral condyles and medial malleolus markers were defined during the static trial and rebuilt from transformation matrix of tibial tuberosities, inferior tibia, and lateral malleolus markers during dynamic trials.

To be consistent with Bell's methods (1990), the pelvis local coordinate was defined using bilateral ASIS and the middle point of PSIS makers. The thigh local coordinate was defined using hip joint center, knee joint center, and lateral femoral condyle markers. The shank local coordinate was defined using knee joint center, ankle joint center, and lateral femoral condyle markers. The foot local coordinate was defined using first toe, heel, and fifth metatarsal head makers. Cardan joint angles between adjacent segment local coordinates were calculated in an order of flexion–extension, valgus–varus, and internal–external rotation (Chao, 1980; Grood and Suntay, 1983). Angular velocities and angular accelerations were determined using segment local coordinates (Chao, 1980; Hong and Bartlett, 2008).

Segment masses, center of mass locations, and segment moments of inertia were based on modified Clauser's methods. An inverse dynamics approach was used to calculate the lower extremity joint resultant forces and resultant moments (Greenwood, 1987). Joint resultant forces and moments were transferred to the local reference frames and expressed as internal forces and moments. Joint mechanical work was calculated as the time integration of joint power which was the product of joint resultant moments and joint angular velocities. Force data were normalized to body weight. Moment data were normalized to the product and body weight and body height. Anterior, medial, and superior forces were defined as positive forces. Knee flexion, internal rotation, and varus angles and moments were defined as positive angles and moments. All data calculations were performed in a MS3D70 computer program package (MotionSoft, Chapel Hill, NC).

3.4.2. Musculoskeletal Model

A musculoskeletal model (Figure 3.2) was used to estimate ACL force during the landing phase of stop-jump and cutting tasks. The components that contributed to ACL loading included tibial anterior shear force, tibial internal - external rotation moment, and tibial varus - valgus moment. Muscle moment arms, muscle peak forces, directions of muscle forces, and joint geometry were obtained from the literature. Gastrocnemius force, hamstring force, and patellar tendon force were estimated from a modified torque driven model using lower extremity joint angles and joint resultant moments (DeVita and Hortobagyi, 2001; Kulas et al., 2010). The estimated muscle forces and knee joint resultant moment and force were used to calculate tibiofemoral contact force, tibial anterior shear force, and ACL force (Kernozek and Ragan, 2008; Lin et al., 2009). All data calculations were performed using a customized computer program written in Matlab 7.4.0 (MathWorks Inc., PA, USA).

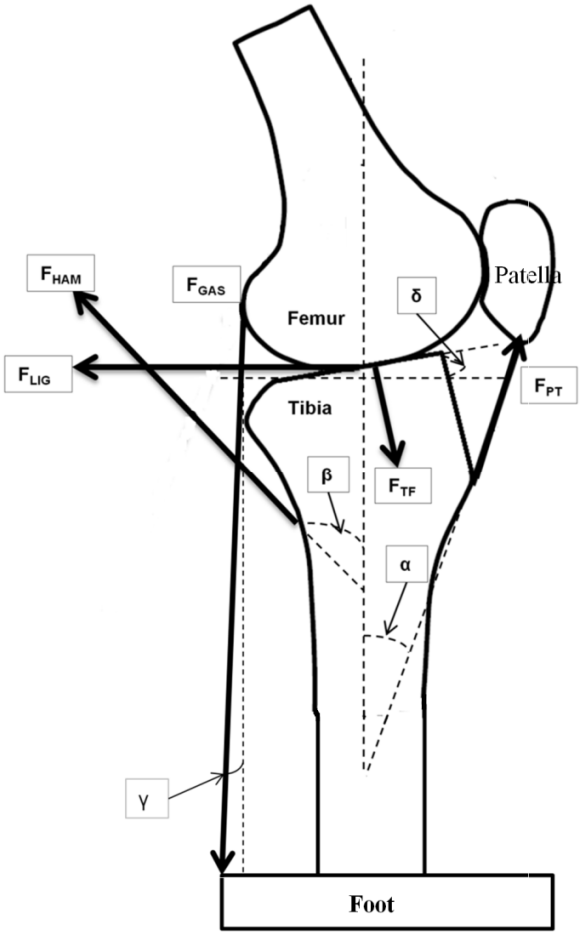

FIGURE 3.2 The knee geometry and muscle forces. F_{GAS}: gastrocnemius force; F_{HAM}: hamstring force; F_{PT}: patella tendons force; F_{TF}: tibiofemoral contact force; F_{SOF}: Soft tissue force; δ: posterior tibial plateau slope angle; α: patellar tendon-tibia shaft angle; β: hamstring-tibia shaft angle; γ: gastrocnemius-tibia shaft angle.

Determination of Gastrocnemius Force

The Achilles tendon moment arms for ankle joint were determined by digitizing the data graphically reported by Maganaris et al. (2000) from 22.5° of ankle dorsiflexion to 37.5° of ankle plantarflexion. The moment arms for more than 22.5° of dorsiflexion were assumed to be equal to the moment arm for 22.5° of dorsiflexion. The moment arms for more than 37.5° of plantarflexion were assumed to be equal to the moment arm for 37.5° of plantarflexion. The calculated moment arms were then adjusted with body height.

For $-22.5^\circ \leq A_A \leq 37.5^\circ$

$$r_{AT} = 4.79\times10^{-2} + 3.99\times10^{-4}\times A_A + 3\times10^{-6}\times A_A^{\,2} - 2.12\times10^{-7}\times A_A^{\,3}$$

($R^2 > 0.99$) 3.1

For $A_A < -22.5^\circ$

$$r_{AT} = 4.3\times10^{-2} \quad 3.2$$

For $A_A > 37.5^\circ$

$$r_{AT} = 5.6\times10^{-2} \quad 3.3$$

where r_{AT} was Achilles tendon moment arm for ankle joint and A_A was ankle plantarflexion (+) - dorsiflexion (-) angle.

The sum of gastrocnemius and soleus forces were determined by dividing the ankle plantarflexion moments (calculated from inverse dynamics) by Achilles tendon moment arms for ankle joint, assuming no co-contraction by the ankle dorsifiexors. The ratio between gastrocnemius force and soleus force was assumed to be the same as the ratio between their peak forces (gastrocnemius: 1914.4 N, soleus: 3585.9 N) (Arnold et al., 2010). Previous investigator have observed non-significant differences in EMG patterns between gastrocnemius and soleus muscles during the landing phase of drop landing tasks (Iida et al., 2011). These findings suggested that the assumption of equal muscle activation level between gastrocnemius and soleus might be reasonable. The gastrocnemius and soleus forces were zero if the ankle resultant moments were dorsiflexion moments.

$$\frac{F_{GAS}}{F_{SOL}} = \frac{1914.4}{3585.9} \qquad 3.4$$

$$F_{GAS} + F_{SOL} = \frac{M_{A,P}}{r_{AT}} \qquad 3.5$$

where F_{GAS} was gastrocnemius force; F_{SOL} was soleus force; $M_{A,P}$ was ankle plantarflexion moment; r_{AT} was Achilles tendon moment arm for ankle joint. F_{GAS} and F_{SOL} were the two unknowns in equations 3.4 and 3.5.

Determination of Hamstring Force

The hamstring and gluteus maximus moment arms for hip joint were determined by data reported by Nemeth and Ohlsen (1985) from 5° to 90° of hip flexion for males and females respectively. The moment arms for less than 5° of flexion were assumed to be equal to the moment arm for 5° of flexion. The moment arms for more than 90° of flexion were assumed to be equal to the moment arm for 90° of flexion. The calculated moment arms were then adjusted with body height.

For males

For $5^\circ \leq A_H \leq 90^\circ$

$$r_{GM} = 8.23\times10^{-2} - 2.43\times10^{-4}\times A_H - 4.52\times10^{-6}\times A_H^{\ 2} + 1.3\times10^{-8}\times A_H^{\ 3}$$

(R^2>0.99) 3.6

$$r_{HAM,H} = 6.36\times10^{-2} + 8.98\times10^{-4}\times A_H - 1.25\times10^{-5}\times A_H^{\ 2} + 6.65\times10^{-9}\times A_H^{\ 3}$$

(R^2>0.99) 3.7

For $A_H < 5^\circ$

$$r_{GM} = 8.1\times10^{-2}$$ 3.8

$$r_{HAM,H} = 6.8\times10^{-2}$$ 3.9

For $A_H > 90^\circ$

$$r_{GM} = 3.3 \times 10^{-2} \quad 3.10$$

$$r_{HAM,H} = 4.8 \times 10^{-2} \quad 3.11$$

where r_{GM} was gluteus maximus moment arm for hip joint; $r_{HAM,H}$ was hamstring moment arm for hip joint; A_H was hip flexion (+) - extension (-) angle.

For females:

For $5^{\circ} \leq A_H \leq 90^{\circ}$

$$r_{GM} = 7.63 \times 10^{-2} - 2.21 \times 10^{-4} \times A_H - 3.95 \times 10^{-6} \times A_H^{2} + 8.71 \times 10^{-9} \times A_H^{3}$$

(R^2>0.99) 3.12

$$r_{HAM,H} = 5.75 \times 10^{-2} + 7.42 \times 10^{-4} \times A_H - 1.11 \times 10^{-5} \times A_H^{2} + 7.68 \times 10^{-9} \times A_H^{3}$$

(R^2>0.99) 3.13

For $A_H < 5^{\circ}$

$$r_{GM} = 7.5 \times 10^{-2} \quad 3.14$$

$$r_{HAM,H} = 6.1 \times 10^{-2} \quad 3.15$$

For $A_H > 90^{\circ}$

$$r_{GM} = 3.1 \times 10^{-2} \quad 3.16$$

$$r_{HAM,H} = 4.0 \times 10^{-2} \quad 3.17$$

where r_{GM} was gluteus maximus moment arm for hip joint; $r_{HAM,H}$ was hamstring moment arm for hip joint; A_H was hip flexion (+) - extension (-) angle.

The sum of hamstring and gluteus maximus moments for hip joint was modeled to be hip extension moments (calculated from inverse dynamic), assuming no co-contraction by the hip flexors. The ratio between hamstring moment and gluteus maximus moment was assumed to be the same as the ratio between peak hamstring muscle force (2169.8 N) multiplying hamstring moment arm and peak gluteus maximus (1852.7 N) muscle force multiplying gluteus maximus moment arms (Arnold et al., 2010). Previous investigators have observed non-significant differences in EMG patterns between biceps femoris and gluteus maximus muscles during the landing phase of drop landing tasks (Iida et al., 2011). These findings suggested that the assumption of equal muscle activation level between hamstring and gluteus maximus might be reasonable. The hamstrings and gluteus maximus forces were zero if the hip resultant moments were flexion moments.

$$F_{Ham} \times r_{HAM,H} + F_{GM} \times r_{GM} = M_{H,E} \quad 3.18$$

$$\frac{F_{HAM} \times r_{HAM,H}}{F_{GM} \times r_{GM}} = \frac{2169.8 \times r_{HAM,H}}{1852.7 \times r_{GM}} \qquad 3.19$$

where F_{HAM} was hamstring force; $r_{HA,H}$ was hamstring moment arm for hip joint; F_{GM} was gluteus maximus force; r_{GM} was gluteus maximus moment arm for hip joint; $M_{H,E}$ was hip extension moment; F_{HAM} and F_{GM} were the two unknowns in equations 3.18 and 3.19.

Determination of Patella Tendon Force

The moment arms of patella tendon and hamstring for knee joint were determined by data reported by Smidt et al. (1973) from 0° to 90° of knee flexion. The moment arms for more than 90° of flexion were assumed to be equal to the moment arm for 90° of flexion. The calculated moment arms were then adjusted with body height.

For $0^\circ \leq A_K \leq 90^\circ$

$$r_{PT} = 4.39 \times 10^{-2} + 2.86 \times 10^{-4} \times A_K - 3.92 \times 10^{-6} \times A_K^{\ 2} - 2.03 \times 10^{-23} \times A_K^{\ 3}$$

(R^2>0.99) 3.20

$$r_{HAM,K} = 2.51 \times 10^{-2} + 6.72 \times 10^{-4} \times A_K - 6.72 \times 10^{-6} \times A_K^{\ 2} - 8.23 \times 10^{-9} \times A_K^{\ 3}$$

(R^2>0.99) 3.21

For $A_K > 90^\circ$

$$r_{PT} = 3.8 \times 10^{-2} \quad 3.22$$

$$r_{HAM,K} = 2.5 \times 10^{-2} \quad 3.23$$

where r_{PT} was patellar tendon moment arm for knee joint; $r_{HAM,K}$ was hamstring moment arm for knee joint; A_K was knee flexion (+) - extension (-) angle.

The gastrocnemius moment arms normalized by tibia length for knee joint were determined by digitizing the data graphically reported by Visser et al. (1990) from 0° to 100° of knee flexion. The moment arms for more than 100° of flexion were assumed to be equal to the moment arm for 100° of flexion. The tibia length for each subject was determined as the distance between knee center and ankle center visual markers during the static trial. The calculated moment arms were then adjusted with tibia length.

For $0^\circ \leq A_K \leq 100^\circ$

$$r_{GAS} = \text{Tibia_Length} \times (4.57 \times 10^{-2} - 1.34 \times 10^{-4} \times A_K) \quad 3.24$$

For $A_K > 100^\circ$

$$r_{GAS} = 3.23 \times 10^{-2} \quad 3.25$$

where r_{Ga} is gastrocnemius moment arm for knee joint; A_K is knee flexion (+) - extension (-) angles.

Patella tendon forces were calculated as a function of knee resultant moments (calculated from inverse dynamic), gastrocnemius moments for knee joint, hamstring moments for knee joint, and patellar tendon moment arms for knee joint.

$$F_{PT} \times r_{PT} + F_{HAM} \times r_{HAM,K} + F_{GAS} \times r_{GAS} = M_{K,EF} \qquad 3.26$$

where F_{PT} was patellar tendon force; r_{PT} was patellar tendon moment arm for knee joint; F_{HAM} was hamstring force; $r_{HAM,K}$ was hamstring moment arm for knee joint; F_{GAS} was gastrocnemius force; r_{GAS} was gastrocnemius moment arm for knee joint; $M_{K,EF}$ was knee extension - flexion moment. F_{PT} was the only unknown in equation 3.26.

Determination of Tibiofemoral Contact Force

A posterior tibial plateau slope was modeled as 5.6 degrees for males and 5.0 degrees for females (Hohmann et al., 2011). The gastrocnemius-tibia shaft angle was model as 3° (DeVita and Hortobagyi, 2001). The patellar tendon-tibia shaft angles and hamstring-tibia shaft angles were modeled as a function of knee flexion-extension angles for males and females respectively (Nunley et al., 2003; Lin et al., 2009).

For males:

$$\alpha = 22.03 - 0.3 \times A_K$$

(R^2=0.83) 3.27

$$\beta = 0.001 + 0.89 \times A_K$$

(R^2=0.96) 3.28

For females:

$$\alpha = 25.7 - 0.3 \times A_K$$

(R^2=0.83) 3.29

$$\beta = 0.001 + 0.89 \times A_K$$

(R^2=0.96) 3.30

where α was patellar tendon-tibia shaft angle; β was hamstring-tibia shaft angle; A_K was knee flexion (+) - extension (-) angle.

Tibiofemoral contact forces were calculated as a function of knee resultant forces in the superior- inferior direction (calculated from inverse dynamic), gastrocnemius forces, hamstring forces, patellar tendon forces, the posterior tibial plateau slope angle, gastrocnemius-tibia shaft angles, patellar tendon-tibia shaft angles, and hamstring- tibia shaft angles.

$$F_{TF} \times \cos(\delta) + F_{PT} \times \cos(\alpha) + F_{HAM} \times \cos(\beta) + F_{GAS} \times \cos(\gamma) = F_{K,SI} \quad 3.31$$

where F_{TF} was tibiofemoral contact force; δ was posterior tibial plateau slope angle; F_{PT} was patellar tendon forces; α was patellar tendon-tibia shaft angle; F_{HAM} was hamstring force; β was hamstring-tibia shaft angle; F_{GAS} was gastrocnemius force; γ was gastrocnemius-tibia shaft angle; $F_{K,SI}$ was knee joint resultant force in the superior-inferior direction. F_{TF} was the only unknown in equation 3.31.

Determination of Knee Ligament Force and Tibial Anterior Shear Force

Knee ligament forces were calculated as a function of knee joint resultant forces in the anterior-posterior direction, gastrocnemius forces, hamstring forces, patellar tendon forces, tibiofemoral contact forces, posterior tibial plateau slope angle, gastrocnemius-tibia shaft angles, patellar tendon-tibia shaft angles, and hamstring-tibia shaft angles. Tibial anterior shear forces had the same magnitude but different direction as the ligament forces.

$$F_{SOF} + F_{TF} \times \sin(\delta) + F_{PT} \times \sin(\alpha) + F_{HAM} \times \sin(\beta) + F_{GAS} \times \sin(\gamma) = F_{K,AP} \qquad 3.32$$

$$F_{SOF} = -F_{AS} \qquad 3.33$$

where F_{SOF} was knee soft tissue force; F_{TF} was tibiofemoral contact force; δ was posterior tibial plateau slope angle; F_{PT} was patellar tendon force; α was patellar tendon-tibia shaft angle; F_{HAM} was hamstring force; β was hamstring- tibia shaft angle; F_{GAS} was gastrocnemius force; γ was gastrocnemius-tibia shaft angle; $F_{K,AP}$

was knee resultant force in the anterior-posterior direction; F_{AS} was the tibial anterior shear force. F_{SOF} and F_{AS} were the two unknowns in 3.32 and 3.33.

Determination of ACL Force

The first component of ACL force was caused by the tibial anterior shear force. The ACL forces caused by a 100 N anterior shear force at different knee angles were determined by digitizing the data graphically reported by Markolf et al. (1995). Markolf et al. (1995) recorded in *vitro* ACL forces when a 100 N of tibial anterior shear force was applied to the cadaver knees from 90 degree to 5 degrees of flexion. ACL forces were also measured when an additional load of 10 Nm valgus, varus, internal rotation, or external rotation moment was combined with the 100N anterior shear force load.

For $A_K < 0^\circ$

$$F_{100} = 160 \qquad 3.34$$

For $0^\circ \leq A_K \leq 90^\circ$

$$F_{100} = 160.29 - 2.00 \times A_K + 1.15 \times 10^{-2} \times A_K^{\ 2}$$

$(R^2 > 0.99)$ 3.35

For $A_K > 90^\circ$

$$F_{100} = 75 \qquad 3.36$$

where F_{100} was ACL force with a 100 N anterior tibia force; A_K was knee flexion (+) - extension (-).

The second component of ACL force was caused by the tibial internal - external rotation moment. The ACL forces caused by a 10 Nm internal rotation moment at different knee angles with or without a tibial anterior shear force were determined by digitizing the data graphically reported by Markolf et al. (1995). Effects of internal rotation moment on ACL force were modeled differently with or without a tibial anterior shear force. Internal rotation moment caused ACL force only when there was a knee internal rotation angle. Internal rotation moment would cause 0 ACL force if there was a knee external rotation angle.

When there was a knee internal rotation angle and a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_IR} = 115 \qquad 3.37$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_IR} = 118.15 - 7.04 \times A_K + 9.47 \times 10^{-2} \times A_K^{\ 2} - 3.9 \times 10^{-4} \times A_K^{\ 3}$$

(R^2>0.99) 3.38

For $A_K > 90^\circ$

$$F_{10_IR} = -32 \qquad 3.39$$

where F_{10_IR} was ACL force with a 10 Nm internal rotation moment; A_K was knee flexion (+) - extension (-).

When there was a knee internal rotation angle but not a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_IR} = 230 \qquad 3.40$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_IR} = 231.56 - 8.57 \times A_K + 1.20 \times 10^{-1} \times A_K^{\ 2} - 5.01 \times 10^{-4} \times A_K^{\ 3}$$

(R^2>0.99) 3.41

For $A_K > 90^\circ$

$$F_{10_IR} = -62 \qquad 3.42$$

where F_{10_IR} was ACL force with a 10 Nm internal rotation moment; A_K was knee flexion (+) - extension (-).

The ACL forces caused by a 10 Nm external rotation moment at different knee angles with or without a tibial anterior shear force were determined by digitizing the data graphically reported by Markolf et al. (1995). Effects of external rotation moment on ACL force were modeled differently with or without a tibial anterior shear

force. External rotation moment caused ACL force only when there was a knee external rotation angle. External rotation moment would cause 0 ACL force if there was a knee internal rotation angle.

When there was a knee external rotation angle and a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_ER} = -35 \qquad 3.43$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_ER} = -33.53 - 8.19 \times 10^{-1} \times A_K + 5.1 \times 10^{-3} \times A_K^{\ 2}$$

(R^2=0.93) 3.44

For $A_K > 90^\circ$

$$F_{10_ER} = -68 \qquad 3.45$$

where F_{10_ER} was ACL force with a 10 Nm external rotation moment; A_K was knee flexion (+) - extension (-).

When there was a knee external rotation angle but not a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_ER} = 65 \qquad 3.46$$

For $0^{\circ} \le A_K \le 90^{\circ}$

$$F_{10_ER} = 64.33 - 2.15 \times A_K + 3.10 \times 10^{-2} \times A_K^{\ 2} - 1.59 \times 10^{-4} \times A_K^{\ 3}$$

(R^2>0.99) 3.47

For $A_K > 90^{\circ}$

$$F_{10_ER} = 7 \qquad 3.48$$

where F_{10_ER} was ACL force with a 10 Nm external rotation moment; A_K was knee flexion (+) - extension (-).

The third component of ACL force was caused by the tibial varus - valgus rotation moment. The ACL forces caused by a 10 Nm varus moment at different knee angles with or without a tibial anterior shear force were determined by digitizing the data graphically reported by Markolf et al. (1995). Effects of varus moment on ACL force were modeled differently with or without a tibial anterior shear force. Varus moment caused ACL force only when there was a knee varus angle. Varus moment would cause 0 ACL force if there was a valgus angle.

When there was a knee varus angle and a tibial anterior shear force

For $A_K < 0^{\circ}$

$$F_{10_VAR} = 42 \qquad 3.49$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_VAR} = 43.33 - 1.01 \times A_K - 1.20 \times 10^{-2} \times A_K^{\ 2} + 5.58 \times 10^{-4} \times A_K^{\ 3} - 3.80 \times 10^{-6} \times A_K^{\ 4}$$

(R^2=0.96) 3.50

For $A_K > 90^\circ$

$$F_{10_VAR} = 14 \qquad 3.51$$

where F_{10_VAR} is ACL force with a 10 Nm varus moment; A_K is knee flexion (+) - extension (-).

When there was a knee varus angle but not a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_VAR} = 100 \qquad 3.52$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_VAR} = 100.32 - 3.73 \times A_K + 5.70 \times 10^{-2} \times A_K^{\ 2} - 2.68 \times 10^{-4} \times A_K^{\ 3}$$

(R^2>0.99) 3.53

For $A_K > 90^\circ$

$$F_{10_VAR} = 32 \qquad 3.54$$

where F_{10_VAR} is ACL force with a 10 Nm varus moment; A_K is knee flexion (+) - extension (-).

The ACL forces caused by a 10 Nm valgus moment at different knee angles with or without a tibial anterior shear force were determined by digitizing the data graphically reported by Markolf et al. (1995). Effects of valgus moment on ACL force were modeled differently with or without a tibial anterior shear force. Valgus moment caused ACL force only when there was a knee valgus angle. Valgus moment would cause 0 ACL force if there was a varus angle.

When there was a knee valgus angle and a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_VAL} = 2 \qquad 3.55$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_VAL} = 3.53 + 5.63 \times A_K - 1.09 \times 10^{-1} \times A_K^{\ 2} + 5.86 \times 10^{-4} \times A_K^{\ 3}$$

(R^2>0.99) 3.56

For $A_K > 90^\circ$

$$F_{10_VAL} = 55 \qquad 3.57$$

where F_{10_VAL} is ACL force with a 10 Nm valgus moment; A_K is knee flexion (+) - extension (-).

When there was a knee valgus angle but not a tibial anterior shear force

For $A_K < 0^\circ$

$$F_{10_VAL} = 60 \qquad 3.58$$

For $0^\circ \le A_K \le 90^\circ$

$$F_{10_VAL} = 59.24 - 1.34 \times A_K + 7.72 \times 10^{-2} \times A_K^{\ 2} - 1.4 \times 10^{-3} \times A_K^{\ 3} + 7.31 \times 10^{-6} \times A_K^{\ 4}$$

(R^2=0.97) 3.59

For $A_K > 90^\circ$

$$F_{10_VAL} = 42 \qquad 3.60$$

where F_{10_VAL} is ACL force with a 10Nm valgus moment; A_K is knee flexion (+) - extension (-).

Finally, the ACL forces during dynamic tasks were determined as the sum of ACL forces caused by tibial anterior shear force, ACL forces caused by tibial internal - external rotation moment, and ACL forces caused by tibial varus - valgus moment.

$$F_{ACL} = \frac{F_{100}}{100} \times F_{AS} + \frac{F_{10_IR-ER}}{10} \times M_{IR-ER} + \frac{F_{10_VAR-VAL}}{10} \times M_{VAR-VAL} \quad 3.61$$

where F_{ACL} was ACL forces. F_{100} was ACL force with a 100 N tibial anterior tibia force; F_{AS} was tibial anterior shear force; F_{10_IR-ER} was ACL force with a 10Nm internal - external rotation moment; M_{IR-ER} was knee internal - external rotation moment; $F_{10_VAR-VAL}$ was ACL force with a 10Nm varus - valgus moment; $M_{VAR-VAL}$ was knee varus - valgus moment.

3.4.3. Assumptions and Limitations of the Musculoskeletal Model

A number of assumptions were made for the musculoskeletal model to estimate ACL force. The first major assumption of the model was that there was no muscle co-contraction at the ankle and hip joints. This assumption was made in order to calculate the gastrocnemius and hamstring muscle forces. However, previous investigators have observed significant co-contractions at the ankle and hip joints during jump landing tasks (Chappell et al., 2007; Iida et al., 2011). In the current model, co-contraction at the ankle would increase the gastrocnemius force which would affect the ACL force. Co-contraction at the hip would increase the hamstring force which would affect the ACL force. To address the influence of this assumption to the results of ACL force, a sensitivity analysis was conducted to assess the effect of different percentages of co-contraction at ankle and hip on the estimate of ACL force. The sensitivity analysis demonstrated whether a lack of consideration of co-contraction would overestimate or underestimate ACL force.

However, the specific co-contraction pattern during each jumping and cutting task was still unknown and the unknown co-contraction pattern at ankle and hip joints was the major limitation of the current model.

The second major assumption of the model was that the force distribution between gastrocnemius and soleus and the force distribution between hamstring and gluteus maximus only depended on their peak muscle forces and moment arms without a consideration of muscle activation level, muscle force-length relationship, and muscle force-velocity relationships. Previous investigator had observed non-significant differences in EMG patterns between biceps femoris and gluteus maximus muscles and between gastrocnemius and soleus muscles during the landing phase of drop landing tasks (Iida et al., 2011). These findings suggested that the assumption of equal muscle activation level between gastrocnemius and soleus and between hamstring and gluteus maximus might be reasonable. The equal muscle activation level agreed with the assumption that the muscle force distribution was only dependent on peak muscle forces and muscle moment arms without considering muscle activation levels. However, the similar muscle activation levels observed by previous investigators (Iida et al., 2011) might not translate to the tasks and subjects in the current study. The lack of input of muscle activation level was a major limitation of the current model. In addition, the muscle force-length and force-velocity relationships were not included in the model.

In addition, it was assumed that muscle moment arms obtained from the literature was proportional to subjects' body height or segment length. The effect of muscle activation on the length of muscle moment arm was small and negligible.

The lines of action of muscle force were assumed to be the same crossing different subjects. The moments generated by passive tissues including ligaments and joint capsule were small and negligible. The effects of friction force between femur and tibia were small and negligible. The ACL loadings caused by anterior shear force, internal rotation moment, and varus - valgus moment were additive.

All previously mentioned assumptions of the model limited its application and generalization to the real world. The ACL force estimated from the model should be interpreted with caution.

3.4.4. Face Validity of the Musculoskeletal Model

In the current model, the ACL forces were estimated. The golden standard for validating the model was to compare the estimated force with in *vivo* forces. However, measuring in *vivo* ACL forces involves methods and rarely is feasible (Cerulli et al., 2003). The changes in ACL forces during dynamic tasks could be indirectly validated by measuring changes in ACL length using a noninvasive fluoroscopic and magnetic resonance imaging technique (Taylor et al., 2011). However, this technique is demanding in terms of resources and was not feasible for the current study. Therefore, the face validity of the model was only assessed by comparing the results to previous literature.

First of all, previous investigators conducted video analysis and found that the timing of ACL injuries usually occurred within the first 50 milliseconds after the initial foot contact with the ground (Krosshaug et al., 2007; Koga et al., 2010). Previous

investigators also found that peak ACL strain after landing was likely to occur at the timing of peak impact GRF (Cerulli et al., 2003, Taylor et al., 2011). Peak ACL forces during the stance phase of the landing of jumping and cutting tasks were calculated in the current study. If the timing of the peak ACL forces estimated from the model was within the first 50 milliseconds after the landing and occurred at timing of peak impact GRF, the face validity of the model would be supported.

Secondly, previous investigators have employed in *vivo* measurement to evaluate ACL force / length during jump landing tasks (Cerulli et al., 2003; Pflum et al., 2004; Kernozek and Ragan, 2008; Laughlin et al., 2011; Taylor et al., 2011) (Table 3.3; 3.4). Based on ACL tensile properties (Chandrashekar et al., 2006), the ACL force could be estimated from ACL length. The peak ACL force calculated in the current study was compared with previous studies. If the peak ACL force estimated in the current study was within the range of peak ACL forces calculated in previous literature, the face validity of the model would be supported.

Thirdly, previous studies have shown that tibial anterior shear force applied was the major loading mechanism of ACL force, while knee internal - external rotation moments and valgus - varus moments had relatively small contribution to ACL force (Berns et al., 1992, Fleming et al., 2001). The composition of peak ACL force in the current study would be analyzed. If the anterior shear force was the major loading mechanism of peak ACL force, the face validity of the model would be supported.

Fourthly, Brown et al. (2012) recently demonstrated that landing with increased knee flexion angle decreased peak ACL length during the landing.

Therefore, if the peak ACL force estimated in the current study was less during landing with increased knee flexion condition than landing with regular knee flexion condition, the face validity of the model would be supported.

Table 3.3. Previous modeling studies that estimated ACL force

Studies	Methods	Tasks	Peak ACL force
Kernozek et al.	EMG driven model	Drop landing	94 N
Pflum et al.	EMG driven model	Drop landing	253 N
Laughlin et al.	Optimization model	Soft and stiff single-leg landing	449 N for soft landing; 506 N for stiff landing

Table 3.4. Previous in *vivo* studies that calculated ACL strain

Studies	Methods	Tasks	Peak ACL strain	Estimated peak ACL force
Cerulli et al.	Strain Gauge	Hop landing	5.47%	505 N
Taylor et al.	Fluoroscope with MRI	Drop vertical jump	12% for prelanding; 7% for landing	1109 N for pre-landing; 647 N for landing

3.4.5. Sensitivity Analysis

The major assumption of the musculoskeletal model to estimate ACL force was that there was no muscle co-contraction at the ankle and hip joints. Co-contraction at the ankle would increase the gastrocnemius force which would affect the ACL force. Co-contraction at the hip would increase the hamstring force which would affect the ACL force. To address the effect of this assumption on the results of ACL force, a sensitivity analysis was conducted. Different combinations of 50% and 100% of co-contraction at the ankle and hip joints were applied to the model. The effects of co-contraction on the magnitude and timing of peak ACL force were assessed.

The percentage of co-contraction was defined using the percentage of joint resultant moments generated by antagonist muscles. For example, if the joint resultant moment was a 100 Nm extension moment at the hip joint, 100% co-contraction meant that the hip flexors generated 100 Nm flexion moments at the hip joint. Therefore, the hip extensors needed to generate a 200 Nm extension moment to achieve a 100 Nm extension joint resultant moment. On the other hand, if the joint resultant moment was a 100 Nm flexion moment at the hip joint, 100% co-contraction meant that the hip extensors generated 100 Nm extension moments at the hip joint. Therefore, the hip flexors needed to generate a 200 Nm flexion moment to achieve a 100 Nm flexion joint resultant moment.

A 50% co-contraction at the ankle generally increased the gastrocnemius force to 1.5 times of the magnitude without co-contraction. A 100% co-contraction at

the ankle generally increased the gastrocnemius force to 2 times of the magnitude without co-contraction. A 50% co-contraction at the hip generally increased the hamstring force to 1.5 times of the magnitude without co-contraction. A 100% co-contraction at the hip generally increased the hamstring force to 2 times of the magnitude without co-contraction.

Another additional simulation was conducted to assess the direct effects of hamstring co-contraction on ACL force. One unit hamstring force would be applied to the knee joint at different knee flexion angles. Because the hamstring force generated a knee flexion moments, quadriceps forces were adjusted to generate a knee extension moment to maintain the equilibrium of joint resultant moments. The tibiofemoral compressive force caused by hamstring and quadriceps forces was also calculated. The sum of the posterior shear force caused by hamstring force, the anterior shear force caused by quadriceps, and the anterior shear force caused by tibiofemoral compression force were evaluated at different knee flexion angle. The knee flexion angle at which the total shear force started to become posterior shear force was determined for males and females respectively.

3.4.6. Performance Measures

Performance measures for the stop-jump task included jump height, approach speed, stance time, and sagittal plane total lower extremity mechanical work. Performance measures for the side-cutting task included approach speed, take-off speed, stance time, and total sagittal plane lower extremity mechanical work.

Jump height was calculated by subtracting vertical coordinate of the center of left and right ASIS and left and right PSIS makers during the static trial from the maximum vertical coordinate of the center of left and right ASIS and left and right PSIS makers during jump tasks.

The instant speed of the center of left and right ASIS and left and right PSIS markers at the toe-touch and toe-off was calculated to quantify approach and take-off speed.

Stance time was calculated as the time duration of the stance phase from toe-touch to toe-off events.

Lower extremity mechanical work was calculated as the sum of sagittal plane ankle, knee, and hip work during stance phase.

3.4.7. Kinematics and Kinetics that Affect ACL Force

Peak posterior ground reaction forces (PPGRF, FIGIRE 3.3) during the landing phase was calculated and used as a critical time point for knee loading (Cerulli et al., 2003; Lamontagne et al., 2005; Lin et al., 2009). For both stop-jump and side-cutting tasks, the ACL loading variables included knee flexion angles at the initial landing, knee flexion angles at PPGRF, and maximum knee flexion angles. ACL loading variables also include knee varus angles and internal rotation angles at PPGRF as well as PPGRF, peak vertical ground reaction forces, knee extension moments, varus - valgus moments, and internal rotation moments at PPGRF.

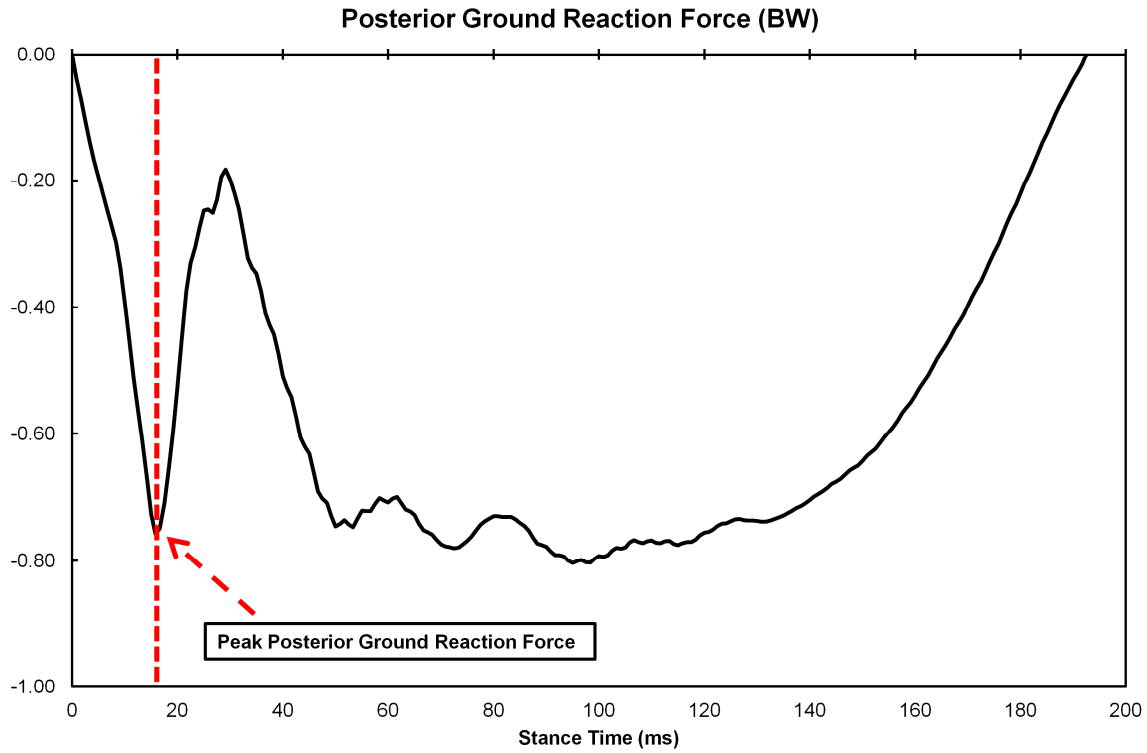

FIGURE 3.3 One representative trial of posterior ground reaction force during a stop-jump task. Peak posterior ground reaction force was identified as the first impact peak force which occurred approximate 20 ms after landing.

3.4.8. Peak ACL force

Magnitude and timing of peak ACL force estimated from the model was calculated during the stance phase to assess changes in ACL loading. The contributions of anterior shear force, internal – external rotation moment, and varus - valgus moments to peak ACL force were also calculated.

3.5. Data Analysis

All the good trials (3-5) out of five official trials for each condition were reduced for data analysis. Within-session reliability was assessed using CMC (Kadaba et al., 1989). Data for dependent variables were averaged across trials for each condition. The univariate z scores were calculated for each variable for each jumping and cutting condition. An outlier was defined as a data point with a z score greater than 3.5 (Iglewicz and Hoaglin, 1993). The univariate normality was screened using Kolmogorov-Smirnov test for each variable for each jumping and cutting condition. A p value less than 0.05 was defined as a violation of the assumption of normality. The inter-subject homoscedasticity between males and females was evaluated using Levene's test for each variable for each jumping and cutting condition. A p value less than 0.05 was defined as a violation of the assumption of homoscedasticity. If the sphericity assumption in repeated measures ANOVAs was violated as detected by Mauchly's test, the Greenhouse - Geisser correction was used (Greenhouse and Geisser, 1959). Because the data were examined by ANOVAs instead of MANOVs, multi-colinearity should not raise a concern in statistical analysis.

Hypothesis 1 involved an assessment of the effects of changes in performance demands on ACL loading. ACL loading variables were compared among conditions of jumping for maximum height, jumping fast, and jumping for 60% of maximum height. The independent variables were jump condition (jumping for

maximum height, jumping fast, and jumping for 60% of maximum height) and gender (male and female). The dependent variables included ACL loading variables and peak ACL force variables. Performance measures were also compared to confirm the differences in performance demands. The differences in dependant variables among stop-jump conditions and between genders were tested using 3 x 2 mixed design ANOVAs with jump condition as within-subject factor and gender as between-subject factor. If an interaction effect of condition and gender was significant, 95% confidence interval method was conducted for pairwise comparisons between each pair of two conditions.

The ACL loading variables and peak ACL force variables were also compared between cutting for maximum speed and cutting for 60% of maximum speed conditions. The independent variables were cutting condition (cutting for maximum speed and cutting for 60% of maximum speed) and gender (male and female). The dependent variables included ACL loading variables and peak ACL force variables. Performance measures were also compared to confirm the differences in performance demands. The differences in dependant variables between cutting conditions and between genders were tested using 2 x 2 mixed design ANOVAs with cutting condition as within-subject factor and gender as between-subject factor. If an interaction effect of condition and gender was significant, 95% confidence interval method was conducted for pairwise comparisons between each pair of two conditions.

Hypothesis 2 involved an assessment of the effects of changes in ACL loading on performance. Performance outcomes were compared among jumping for

maximum height, soft landing, and increased knee flexion landing conditions. The independent variables were jump condition (jumping for maximum height, soft landing, and increased knee flexion landing conditions) and gender (male and female). The dependent variables included performance measures. ACL loading variables and peak ACL force variables were also compared to confirm the differences in ACL loading. The differences in dependant variables among stop-jump conditions and between genders were tested using 3 x 2 mixed design ANOVAs with cutting condition as within-subject factor and gender as between-subject factor. If an interaction effect of condition and gender was significant, 95% confidence interval method was conducted for pairwise comparisons between each pair of two conditions.

The performance outcomes were also compared among cutting with maximum speed, soft landing, and increased knee flexion landing conditions. The independent variables were cutting condition (cutting with maximum speed, soft landing, and increased knee flexion landing conditions) and gender (male and female). The dependent variables included performance measures. ACL loading variables and peak ACL force variables were also compared to confirm the differences in ACL loading. The differences in dependant variables among cutting conditions and between genders were tested using 3 x 2 mixed design ANOVAs with cutting condition as within-subject factor and gender as between-subject factor. If an interaction effect of condition and gender was significant, 95% confidence interval method was conducted for pairwise comparisons between each pair of two conditions.

A Type I error rate of 0.05 was selected as an indication of statistical significance. Statistical analysis was conducted in SPSS 16.0 (SPSS, IL, USA)

3.6. Family-Wise Type I Error Rate

For hypothesis 1, it was hypothesized that ACL loading would increase when the athletes jumped with a higher jump height and a shorter stance time during a stop-jump task. ACL loading would increase when the athletes cut with a faster speed and a shorter stance time during a side-cutting task.

To test hypothesis 1, the peak ACL force would be compared between jumping fast and jumping for maximum height conditions, between jumping fast and jumping for 60% of maximum height conditions, and between jumping for maximum height condition and jumping for 60% of maximum height conditions. The peak ACL force would also be compared between cutting with maximum speed and cutting with 60% of maximum speed condition. A total of 4 pair-wise comparisons would be conducted to test hypothesis 1.

For hypothesis 2, it was hypothesized that soft landing and landing with increased knee flexion at initial contact would decrease ACL loading, but also decrease jump height and cutting speed and increase stance time and mechanical work compared to regular landing during stop-jump and side-cutting tasks.

To test hypothesis 2, the peak ACL force, jump height, approach speed, stance time, and total mechanical work would be compared between jumping for maximum height and jumping with increased knee flexion landing conditions and

between jumping for maximum height and jumping with soft landing conditions. The peak ACL force, approach speed, take-off speed, stance time, and total mechanical work would be compared between cutting for maximum speed and cutting with increased knee flexion landing conditions and between cutting with maximum speed and cutting with soft landing conditions. A total of 20 pair-wise comparisons would be conducted to test hypothesis 2.

The family-wise Type I Error rate for Hypotheses would be calculated as:

$$\alpha_F = 1 - (1 - \alpha_1)(1 - \alpha_2) \ldots (1 - \alpha_n) \qquad 3.62$$

where α_F was family-wise Type I Error rate; α_1 was Type I Error rate for the first significant pair-wise comparison; α_2 was Type I Error rate for the second significant pair-wise comparison; α_n was Type I Error rate for the last significant pair-wise comparison.

3.7. Power Analysis

The current study was a mixed design with testing condition as a within-subject factor and gender as a between-subject factor. Previous studies which investigated the effects of landing conditions and landing techniques on lower extremity biomechanics generally combined males and females together for data analysis and observed medium to large effect size (McNair et al., 2000, Onate et al., 2005, Peng, 2011, Prapavessis and McNair, 1999). The focus the current study was

to compared performance and ACL loading under different conditions within individuals. Gender effects were the secondary analysis for the current study and were not included in the power analysis.

Previous studies that compared the effects of landing techniques and landing conditions on peak impact vertical ground reaction forces, knee range of motions, stance time, and lower extremity mechanical work demonstrated medium to large effect size (Devita and Skelly, 1992; McNitt-Gray, 1993; Zhang et al., 2000; Onate et al., 2001; Walsh et al., 2004; Peng, 2011). Therefore, a medium effect size was assumed for all the dependant variables in the current study. In the current study, the dependant loading variables during the stop-jump task were tested using 3 x 2 mixed design ANOVAs. Assuming the effect size was no less than 0.25 (medium effect size) and the correlation coefficients among repeated measures were no greater than 0.5 for each ANOVA, a sample size of 28 was needed for a type I error no greater than 0.05 and a power no less than 0.8. Assuming the effect size was no less than 0.5 (medium effect size) for each pairwise comparison, a sample size of 34 was needed for a type I error no greater than 0.05 and a power no less than 0.8.

The dependent variables during the cutting task will be tested using 3 x 2 or 2 x 2 mixed design ANOVAs. Assuming the effect size was no less than 0.25 (medium effect size) and the correlation coefficients among repeated measures were no greater than 0.5 for each ANOVA, a sample size of 28 or 34 was needed for a type I error no greater than 0.05 and a power no less than 0.8. Assuming the effect size was no less than 0.5 (medium effect size) for each pairwise comparison, a sample

size of 34 was needed for a type I error no greater than 0.05 and a power no less than 0.8.

In summary, a sample size of 28 subjects was needed for ANOVAs for the stop-jump tasks. A sample size of 34 subjects was needed for ANOVAs for the side-cutting tasks. A sample size of 34 subjects was need for pairwise comparisons for both stop-jump and side-cutting tasks. Therefore, 34 subjects were needed for the current study.

CHAPTER IV

RESULTS

4.1. Subjects

Eighteen male and 18 female subjects (Table 4.1) participated in the study. All subjects met the inclusion criteria. All the testing went well with no accidents or unexpected events. Male subjects were significantly taller and heavier than female subjects (Table 4.1). All subjects had experience playing basketball, soccer, volleyball, or rugby.

Table 4.1. Means (standard deviations) of subject information.

	Males (n = 18)	Females (n = 18)	P-value for Gender Effect
Age (yr)	23.06 (3.73)	21.61 (2.52)	0.19
Height (m)	1.80 (0.05)	1.68 (0.07)	< 0.001
Mass (kg)	76.88 (8.79)	64.87 (6.02)	< 0.001
Sports Experience (yr)	13.39 (4.83)	12.78 (4.74)	0.710

4.2. Reliability

For the stop-jump conditions (Table 4.2), vertical GRF, anterior-posterior GRF, ankle dorsiflexion - plantarflexion angle, ankle dorsiflexion - plantarflexion moment, hip flexion - extension angle, knee flexion - extension angle, knee flexion - extension velocity, knee flexion - extension moment, knee superior - inferior joint resultant force, and knee anterior - posterior joint resultant force had CMCs between 0.8 and 0.94. Medial - lateral GRF, hip flexion - extension moment, and knee valgus - varus angle had CMCs approximately between 0.7 and 0.8. Knee internal - external rotation angle, knee internal - external rotation moment, knee valgus - varus moment, and knee medial - lateral joint resultant force had CMCs approximately 0.6.

CMCs were compared between jumping for maximum height and jumping for 60% of maximum height conditions. CMCs for ankle dorsiflexion - plantarflexion angle ($p = 0.010$), ankle dorsiflexion - plantarflexion moment ($p = 0.040$), hip flexion - extension angle ($p = 0.010$), hip flexion - extension moment ($p = 0.010$), knee flexion - extension velocity ($p = 0.003$), and knee medial - lateral joint resultant force ($p = 0.005$) during jumping for 60% of maximum height condition were significantly less than those during jumping for maximum height condition. No significant difference was observed in CMCs for other variables ($p > 0.050$) between jumping for maximum height and jumping for 60% of maximum height conditions.

CMCs were compared between jumping for maximum height and increased knee flexion landing conditions. CMCs for knee valgus – varus moment ($p = 0.010$), knee superior - inferior joint resultant force ($p = 0.020$), and knee medial - lateral joint

resultant force ($p = 0.001$) during increased knee flexion landing condition were significantly less than those during jumping for maximum height condition. No significant difference was observed in CMCs for other variables ($p > 0.050$) between jumping for maximum height and increased knee flexion landing conditions.

For the side-cutting conditions (Table 4.3), vertical GRF, anterior - posterior GRF, medial - lateral GRF, ankle flexion - extension angle, ankle flexion - extension moment, hip flexion - extension angle, hip flexion - extension moment, knee flexion - extension angle, knee flexion - extension velocity, knee flexion - extension moment, knee superior - inferior joint resultant force, and knee anterior - posterior joint resultant force had CMCs between 0.8 and 0.96. Knee medial - lateral joint resultant force had CMCs between 0.7 and 0.8. Knee valgus - varus angle, knee internal - external rotation angle, knee internal - external rotation moment, and knee valgus - varus moment had CMCs between 0.5 - 0.7.

CMCs were compared between cutting with maximum speed and cutting with 60% of maximum speed conditions. CMCs for vertical GRF ($p = 0.040$), ankle flexion - extension moments ($p < 0.001$), knee valgus - varus angle ($p < 0.001$), knee valgus - varus moment ($p = 0.002$), knee medial - lateral joint resultant force ($p < 0.001$) during cutting with 60% of maximum speed condition were significantly greater than those during cutting with maximum speed. No significant difference was observed in CMCs for other variables ($p > 0.050$) between cutting with maximum speed and cutting with 60% of maximum speed conditions.

CMCs were compared between cutting with maximum speed and increased knee flexion landing conditions. CMC for medial - lateral GRF ($p = 0.010$) and knee

valgus - varus angle ($p < 0.001$) during increased knee flexion landing condition was significantly less than those during jumping for maximum height condition. CMCs for knee flexion - extension velocity ($p = 0.040$), knee valgus - varus angle ($p < 0.001$), knee valgus - varus moment ($p = 0.004$), and knee medial - lateral joint resultant force ($p = 0.010$) during increased knee flexion landing condition were significantly greater than those during jumping for maximum height condition. No significant difference was observed in CMCs for other variables ($p > 0.050$) between cutting with maximum speed and increased knee flexion landing conditions.

Table 4.2. Means (standard deviations) of CMCs for kinematic and kinetic variables during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft landing
Vertical GRF	0.89 (0.07)	0.88 (0.08)	0.87 (0.06)	0.86 (0.07)	0.88 (0.06)
A-P GRF	0.87 (0.08)	0.84 (0.08)	0.85 (0.07)	0.86 (0.07)	0.83 (0.08)
M-L GRF	0.71 (0.14)	0.72 (0.14)	0.74 (0.14)	0.72 (0.14)	0.72 (0.12)
Ankle F-E Angle	0.87 (0.08)	0.92 (0.04)	0.90 (0.06)	0.92 (0.05)	0.91 (0.06)
Ankle F-E Moment	0.84 (0.12)	0.89 (0.07)	0.85 (0.09)	0.86 (0.10)	0.84 (0.09)
Hip F-E Angle	0.88 (0.09)	0.93 (0.06)	0.89 (0.07)	0.94 (0.04)	0.91 (0.06)
Hip F-E Moment	0.66 (0.18)	0.75 (0.14)	0.68 (0.16)	0.79 (0.10)	0.77 (0.12)
Knee F-E Angle	0.91 (0.05)	0.96 (0.03)	0.94 (0.05)	0.94 (0.05)	0.95 (0.03)
Knee F-E Velocity	0.98 (0.01)	0.99 (0.01)	0.98 (0.01)	0.99 (0.01)	0.99 (0.01)
Knee I-E Angle	0.48 (0.26)	0.56 (0.25)	0.53 (0.26)	0.58 (0.25)	0.57 (0.26)
Knee V-V Angle	0.60 (0.22)	0.76 (0.15)	0.74 (0.15)	0.77 (0.16)	0.78 (0.18)
Knee F-E Moment	0.88 (0.08)	0.89 (0.08)	0.90 (0.05)	0.87 (0.09)	0.87 (0.10)
Knee I-E Moment	0.51 (0.18)	0.51 (0.17)	0.50 (0.20)	0.48 (0.21)	0.45 (0.23)
Knee V-V Moment	0.60 (0.16)	0.64 (0.17)	0.61 (0.17)	0.56 (0.19)	0.60 (0.21)
Knee S-I Force	0.89 (0.07)	0.88 (0.08)	0.86 (0.07)	0.84 (0.08)	0.87 (0.06)
Knee A-P Force	0.93 (0.05)	0.93 (0.05)	0.94 (0.03)	0.92 (0.07)	0.93 (0.04)
Knee M-L Force	0.61 (0.22)	0.68 (0.17)	0.60 (0.17)	0.57 (0.21)	0.63 (0.19)

Note: A-P: anterior - posterior; M-L: medial - lateral; S-I: superior - inferior; F-E: flexion - extension; I-E: internal - external rotation; V-V: valgus - varus.

Table 4.3. Means (standard deviations) of CMCs for kinematic and kinetic variables during the side-cutting conditions

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
Vertical GRF	0.95 (0.02)	0.96 (0.02)	0.94 (0.04)	0.94 (0.05)
A-P GRF	0.94 (0.03)	0.93 (0.04)	0.95 (0.03)	0.94 (0.04)
M-L GRF	0.93 (0.04)	0.92 (0.05)	0.90 (0.07)	0.92 (0.06)
Ankle F-E Angle	0.91 (0.08)	0.94 (0.04)	0.92 (0.05)	0.91 (0.08)
Ankle F-E Moment	0.89 (0.08)	0.95 (0.03)	0.88 (0.09)	0.88 (0.10)
Hip F-E Angle	0.92 (0.05)	0.90 (0.06)	0.93 (0.08)	0.90 (0.07)
Hip F-E Moment	0.85 (0.06)	0.83 (0.10)	0.87 (0.07)	0.82 (0.11)
Knee F-E Angle	0.90 (0.06)	0.91 (0.06)	0.92 (0.06)	0.92 (0.07)
Knee F-E Velocity	0.96 (0.02)	0.97 (0.02)	0.97 (0.01)	0.97 (0.02)
Knee I-E Angle	0.59 (0.23)	0.64 (0.21)	0.57 (0.22)	0.52 (0.22)
Knee V-V Angle	0.49 (0.21)	0.65 (0.19)	0.66 (0.19)	0.52 (0.25)
Knee F-E Moment	0.86 (0.08)	0.87 (0.08)	0.89 (0.07)	0.87 (0.08)
Knee I-E Moment	0.66 (0.20)	0.60 (0.24)	0.70 (0.20)	0.61 (0.20)
Knee V-V Moment	0.68 (0.14)	0.79 (0.17)	0.76 (0.174	0.69 (0.16)
Knee S-I Force	0.95 (0.02)	0.96 (0.02)	0.93 (0.05)	0.93 (0.05)
Knee A-P Force	0.95 (0.11)	0.96 (0.02)	0.96 (0.03)	0.96 (0.03)
Knee M-L Force	0.72 (0.16)	0.87 (0.12)	0.80 (0.15)	0.70 (0.21)

Note: A-P: anterior - posterior; M-L: medial - lateral; S-I: superior - inferior; F-E: flexion - extension; I-E: internal - external rotation; V-V: valgus - varus.

4.3. Data Screening

4.3.1. Outlier

An outlier was defined as a data point with a z score greater than 3.5 (Iglewicz and Hoaglin, 1993). There were 26 ACL loading and performance variables for each stop-jump and side-cutting condition. Each variable had 35 data points. For the 5 stop-jump conditions, 37 outliers were identified among all 4680 data points (Table 4.4). For the 4 side-cutting conditions, 37 outliers were identified among all 3744 data points (Table 4.5).

The same statistical tests were conducted with and without the outliers to assess the effects of outliers on statistical outcomes. Including the outliers had minimal effect on the statistical outcomes and the interpretations of the data. Therefore, all the outliers were included in the statistical analysis.

Table 4.4. Number of outliers for variables during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
PPGRF					
Time_PPGRF					
VGRF_PPGRF					
Ini_KF	1		2		
Max_KF	1	1			
ROM_KF			1		
Ini_KFV					
KFV_ PPGRF					
KF_PPGRF		1			
KIR_PPGRF	1				1
KVA_PPGRF	2	1	1	1	1
KFM_PPGRF				1	
KIRM_PPGRF	1			1	
KVAM_PPGRF		1		1	
ACL_Peak					
ACL_AS		1			
ACL_IR	1	1	2		1
ACL_VV	1				
Time_ACL		1	1	1	1
Ankle Work	1	1		1	
Knee Work					
Hip_Work				1	
Total_Work					
Approach Speed			1		
Jump Height			1		
Stance Time					

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

Table 4.5. Number of outliers for variables during the side-cutting conditions.

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
PPGRF	1	1		
Time_PPGRF		1		
VGRF_PPGRF				
Ini_KF	1			
Max_KF				
ROM_KF				
Ini_KFV				
KFV_ PPGRF	1		2	
KF_PPGRF	1			1
KIR_PPGRF	1	1		
KVA_PPGRF	1	1	1	1
KFM_PPGRF		1	1	
KIRM_PPGRF		1		
KVAM_PPGRF	1			1
ACL_Peak	1	1	1	1
ACL_AS		1		
ACL_IR				
ACL_VV	2		1	1
Time _ACL			1	
Ankle Work	1	1	1	
Knee Work				
Hip_Work	1			
Total_Work	1			
Approach Speed				
Take-off Speed				
Stance Time		1		1

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

4.3.2. Normality

There were 26 ACL loading and performance variables for each stop-jump and side-cutting condition. For the 5 stop-jump conditions, 7 variables violated the assumption of normality among all the 130 variables (Table 4.6). For the 4 side-cutting conditions, 5 variables violated the assumption of normality among all the 104 variables (Table 4.7).

A small portion of the variables violated the assumption of normality. Repeated measure ANOVA was robust against moderate violation of normality (Collier et al., 1967; Howell, 2009). Therefore, the violations of normality should have minimal effects on the statistical outcomes.

Table 4.6. P-values of Kolmogorov - Smirnov test for variables during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
PPGRF	0.956	0.685	0.929	0.744	0.374
Time_PPGRF	0.983	0.753	0.604	1.000	0.813
VGRF_PPGRF	0.863	0.889	0.903	0.848	0.442
Ini_KF	0.920	0.910	0.447	0.901	0.592
Max_KF	0.291	0.570	0.838	0.378	0.606
ROM_KF	0.995	0.817	0.641	0.127	0.665
Ini_KFV	0.696	0.658	0.351	0.667	0.618
KFV_ PPGRF	0.627	0.913	0.819	0.639	0.601
KF_PPGRF	0.426	0.840	0.864	0.902	0.806
KIR_PPGRF	0.329	0.868	0.987	0.610	0.544
KVA_PPGRF	0.438	0.550	0.687	0.738	0.390
KFM_PPGRF	0.641	0.590	0.191	0.229	0.406
KIRM_PPGRF	0.931	0.338	0.408	0.437	0.371
KVAM_PPGRF	0.617	0.269	0.168	0.368	0.654
ACL_Peak	0.456	0.393	0.446	0.204	0.644
ACL_AS	0.923	0.611	0.762	0.337	0.934
ACL_IR	0.000	0.020	0.002	0.016	0.002
ACL_VV	0.262	0.434	0.372	0.754	0.093
Time _ACL	0.919	0.065	0.378	0.027	0.388
Ankle Work	0.071	0.107	0.050	0.045	0.137
Knee Work	0.647	0.876	0.434	0.089	0.269
Hip_Work	0.154	0.587	0.188	0.642	0.115
Total_Work	0.718	0.310	0.378	0.599	0.526
Approach Speed	0.950	0.922	0.850	0.386	0.683
Jump Height	0.779	0.411	0.763	0.539	0.622
Stance Time	0.900	0.376	0.911	0.928	0.647

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

Table 4.7. P-values of Kolmogorov - Smirnov test for variables during the side-cutting conditions

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
PPGRF	0.380	0.694	0.373	0.602
Time _PPGRF	0.902	0.503	0.977	0.669
VGRF_PPGRF	0.947	0.223	0.445	0.922
Ini_KF	0.639	0.973	0.479	0.455
Max_KF	0.807	0.786	0.929	0.919
ROM_KF	0.855	0.709	0.841	0.971
Ini_KFV	0.805	0.680	1.000	0.823
KFV_ PPGRF	0.165	0.929	0.484	0.965
KF_PPGRF	0.726	0.754	0.996	0.607
KIR_PPGRF	0.550	0.851	0.706	0.552
KVA_PPGRF	0.674	0.546	0.732	0.362
KFM_PPGRF	0.801	0.002	0.132	0.074
KIRM_PPGRF	0.340	0.332	0.342	0.310
KVAM_PPGRF	0.225	0.140	0.318	0.292
ACL_Peak	0.199	0.909	0.205	0.404
ACL_AS	0.337	0.449	0.757	0.355
ACL_IR	0.275	0.034	0.144	0.315
ACL_VV	0.015	0.878	0.199	0.343
Time _ACL	0.917	0.922	0.056	0.989
Ankle Work	0.194	0.017	0.033	0.214
Knee Work	0.323	0.377	0.464	0.520
Hip_Work	0.216	0.070	0.324	0.094
Total_Work	0.653	0.804	0.663	0.607
Approach Speed	0.645	0.994	0.950	0.917
Take-off Speed	0.478	0.851	0.781	0.754
Stance Time	0.983	0.279	0.832	0.609

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

4.3.3. Homoscedasticity

There were 26 variables for each stop-jump and side-cutting condition. For the 5 stop-jump conditions, 11 variables violated the assumption of between - subject homoscedasticity among all the 130 variables (Table 4.8). For the 4 side-cutting conditions, 19 variables violated the assumption of between - subject homoscedasticity among all the 104 variables (Table 4.9).

A small portion of the variables violated this assumption and ANOVA was robust against moderate violation of between - subject homoscedasticity (Box, 1954; Howell, 2009). Therefore, the violations of between - subject homoscedasticity should have minimal effects on the statistical outcomes.

In summary, repeated measure ANOVA was considered an appropriate statistical method to analyze the data in the current study.

Table 4.8. P-values of Levene test for variables during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
PPGRF	0.464	0.876	0.490	0.085	0.835
Time_PPGRF	0.486	0.729	0.327	0.091	0.201
VGRF_PPGRF	0.448	0.251	0.749	0.115	0.246
Ini_KF	0.608	0.641	0.238	0.973	0.208
Max_KF	0.065	0.499	0.432	0.887	0.269
ROM_KF	0.764	0.688	0.946	0.882	0.720
Ini_KFV	0.084	0.269	0.598	0.586	0.122
KFV_ PPGRF	0.075	0.749	0.011	0.439	0.002
KF_PPGRF	0.094	0.966	0.595	0.573	0.603
KIR_PPGRF	0.011	0.091	0.213	0.185	0.539
KVA_PPGRF	0.625	0.316	0.090	0.110	0.106
KFM_PPGRF	0.208	0.297	0.877	0.010	0.111
KIRM_PPGRF	0.549	0.624	0.764	0.578	0.531
KVAM_PPGRF	0.118	0.236	0.225	0.172	0.110
ACL_Peak	0.336	0.997	0.147	0.613	0.133
ACL_AS	0.447	0.774	0.205	0.428	0.272
ACL_IR	0.496	0.279	0.877	0.721	0.200
ACL_VV	0.037	0.187	0.228	0.413	0.577
Time _ACL	0.665	0.027	0.639	0.053	0.016
Ankle Work	0.098	0.050	0.032	0.100	0.213
Knee Work	0.354	0.442	0.055	0.758	0.163
Hip_Work	0.151	0.362	0.659	0.113	0.150
Total_Work	0.530	0.169	0.465	0.171	0.199
Approach Speed	0.012	0.169	0.073	0.054	0.049
Jump Height	0.307	0.098	0.553	0.008	0.079
Stance Time	0.388	0.868	0.713	0.736	0.820

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

Table 4.9. P-values of Levene test for variables during the side-cutting conditions

	Cut 100% Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
PPGRF	0.912	0.204	0.002	0.398
Time _PPGRF	0.128	0.488	0.625	0.246
VGRF_PPGRF	0.830	0.021	0.051	0.028
Ini_KF	0.069	0.963	0.087	0.324
Max_KF	0.012	0.228	0.535	0.070
ROM_KF	0.077	0.915	0.314	0.347
Ini_KFV	0.688	0.201	0.112	0.349
KFV_ PPGRF	0.029	0.532	0.191	0.357
KF_PPGRF	0.107	0.368	0.966	0.160
KIR_PPGRF	0.104	0.398	0.031	0.032
KVA_PPGRF	0.218	0.073	0.307	0.081
KFM_PPGRF	0.558	0.100	0.186	0.009
KIRM_PPGRF	0.521	0.977	0.024	0.142
KVAM_PPGRF	0.854	0.037	0.706	0.688
ACL_Peak	0.111	0.159	0.646	0.854
ACL_AS	0.777	0.369	0.809	0.310
ACL_IR	0.479	0.028	0.021	0.048
ACL_VV	0.148	0.551	0.518	0.006
Time _ACL	0.722	0.039	0.385	0.454
Ankle Work	0.957	0.596	0.372	0.241
Knee Work	0.182	0.218	0.471	0.888
Hip_Work	0.019	0.081	0.082	0.075
Total_Work	0.004	0.655	0.038	0.061
Approach Speed	0.363	0.281	0.891	0.470
Take-off Speed	0.002	0.226	0.779	0.953
Stance Time	0.128	0.099	0.482	0.468

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion; Ini_KFV: Initial knee flexion velocity; KFV_ PPGRF: Knee flexion velocity at PPGRF; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF; ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external r otation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Time_ACL: Timing of peak ACL force.

4.4. Face Validity of ACL Loading Model

4.4.1. Timing of Peak ACL Force

The peak ACL force occurred within 55 ms after initial contact during all stop-jump conditions (Table 4.10). The peak posterior GRF occurred within 32 ms after initial contact during all stop-jump conditions (Table 4.10). The timing of peak ACL force occurred later than the timing of peak posterior GRF during all jumping conditions. The differences between the timing of peak ACL force and peak posterior GRF were between 6 and 23 ms during all stop-jump conditions.

The peak ACL force occurred within 40 ms after initial contact force during cutting with maximum speed condition (Table 4.11). The peak ACL forces occurred between 50 ms and 90 ms after initial contact during other side-cutting conditions (Table 4.11). The peak posterior GRF occurred within 35 ms after initial contact during all side-cutting conditions (Table 4.11). The timing of peak ACL force occurred later than the timing of peak posterior GRF during all cutting conditions. The differences between the timing of peak ACL force and peak posterior GRF were less than 13 ms during cutting with maximum speed conditions. The differences between the timing of peak ACL force and peak posterior GRF were more than 20 ms during the other side-cutting conditions.

Table 4.10. Means (standard deviations) of peak ACL force timing and peak posterior GRF timing during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
Peak ACL Force Timing (ms)	35.2 (10.6)	51.0 (32.6)	42.7 (18.5)	55.0 (42.2)	49.7 (27.5)
Posterior GRF Timing (ms)	27.6 (8.9)	30.9 (9.2)	31.1 (9.0)	30.8 (8.6)	31.1 (7.3)

Table 4.11. Means (standard deviations) of peak ACL force timing and peak posterior GRF timing during the side-cutting conditions.

	Cut 100% Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
Peak ACL Force Timing (ms)	38.7 (18.7)	88.9 (21.1)	60.0 (43.8)	56.5 (27.7)
Posterior GRF Timing (ms)	26.7 (9.9)	35.2 (12.3)	33.4 (11.2)	31.7 (11.3)

4.4.2. Magnitude of Peak ACL Force

The magnitudes peak ACL forces ranged from 0.64 to 1.34 body weights during different stop-jump and side-cutting conditions (Table 4.12, 4.13).

Table 4.12. Means (standard deviations) of peak ACL force magnitude during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
Peak ACL Force (BW)	1.34 (0.61)	0.87 (0.48)	0.96 (0.46)	0.64 (0.31)	0.71 (0.35)

Table 4.13. Means (standard deviations) of peak ACL force magnitude during the side-cutting conditions

	Cut 100% Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
Peak ACL Force (BW)	1.25 (0.73)	0.84 (0.32)	0.91 (0.39)	0.89 (0.39)

4.4.3. The Composition of Peak ACL Force

Tibial anterior shear force contributed to 43% - 83% of peak ACL force during different stop-jump and side-cutting conditions (Table 4.14; 4.15). Valgus - varus moments contributed to 23 - 50% of peak ACL force during different stop-jump and side-cutting conditions. Internal - external rotation moments contributed to -6% - 8% of peak ACL force during different stop-jump and side-cutting conditions.

Table 4.14. Means (standard deviations) of compositions of peak ACL force during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
Contribution from Anterior Shear Force (%)	63.9 (31.0)	55.5 (37.4)	61.7 (31.6)	42.7 (34.1)	57.4 (34.5)
Contribution from Internal - External Rotation Moments (%)	1.6 (18.6)	5.8 (21.8)	6.4 (15.2)	7.2 (15.0)	3.2 (17.6)
Contribution from Valgus - Varus Moments (%)	34.5 (27.2)	38.6 (28.9)	32.0 (26.9)	50.1 (31.5)	39.4 (30.3)

Table 4.15. Means (standard deviations) of compositions of peak ACL force during the side-cutting conditions

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
Contribution from Anterior Shear Force (%)	55.7 (41.9)	83.2 (29.0)	48.6 (32.1)	69.3 (37.0)
Contribution from Internal - External Rotation Moments (%)	-0.3 (30.6)	-6.0 (16.0)	8.0 (33.0)	-4.4 (27.0)
Contribution from Valgus - Varus Moments (%)	44.5 (28.2)	22.8 (21.1)	43.3 (27.0)	35.1 (29.7)

4.4.4. The Relationship between Peak ACL Force and Knee Flexion Angle

For the stop-jump tasks, the peak ACL force during the increased knee flexion landing condition was less than the peak ACL force during the jumping for maximum height condition (Table 4.12, $p < 0.001$). For the side-cutting tasks, the peak ACL force during the increased knee flexion condition was less than the peak ACL force during the cutting with maximum speed condition (Table 4.13, $p = 0.004$).

4.5. Sensitivity Analysis

Ankle co-contraction had small effects on the magnitude and timing of peak ACL forces (Table 4.16; 4.17; 4.18; 4.19). On average, a 100% ankle co-contraction increased the magnitude of peak ACL force by 7 % and decreased the timing of peak ACL force by 3 % during different stop-jump conditions. A 100% ankle co-contraction increased the magnitude of peak ACL force by 9 % and increased the timing of peak ACL force by 7 % during different side-cutting conditions.

Hip co-contraction had small effects on the magnitude and timing of peak ACL forces (Table 4.16; 4.17; 4.18; 4.19). On average, a 100% hip co-contraction decreased the magnitude of peak ACL force by 5 % and increased the timing of peak ACL force by 1 % during different stop-jump conditions. A 100% hip co-contraction decreased the magnitude of peak ACL force by 3 % and decreased the timing of peak ACL force by 2 % during different side-cutting conditions.

A combination of co-contractions at the ankle and hip joints had small effects on the magnitude and timing of peak ACL forces (Table 4.16; 4.17; 4.18; 4.19). On

average, a 100% ankle and 100% hip co-contraction increased the magnitude of peak ACL force by 2.0 % and decreased the timing of peak ACL force by 1.4 % during different stop-jump conditions. A 100% ankle and 100% hip co-contraction increased the magnitude of peak ACL force by 5 % and decreased the timing of peak ACL force by 1 % during different side-cutting conditions.

The simulation of hamstring co-contraction showed that hamstring co-contraction did not decrease ACL force until the knee flexion angle was greater than 25 degrees for males and 26 degrees for females.

Table 4.16. Mean (standard deviation) of magnitudes of peak ACL force with different percentages of ankle and hip co-contraction during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
0% ankle and 0% hip co-contraction (BW)	1.34 (0.61)	0.87 (0.48)	0.96 (0.46)	0.64 (0.31)	0.71 (0.35)
50% ankle and 0% hip co-contraction (BW)	1.39 (0.62)	0.90 (0.48)	1.00 (0.47)	0.66 (0.33)	0.73 (0.36)
100% ankle and 0% hip co-contraction (BW)	1.44 (0.63)	0.93 (0.50)	1.04 (0.48)	0.68 (0.34)	0.76 (0.37)
0% ankle and 50% hip co-contraction (BW)	1.30 (0.61)	0.83 (0.47)	0.94 (0.47)	0.62 (0.30)	0.69 (0.35)
0% ankle and 100% hip co-contraction (BW)	1.28 (0.62)	0.82 (0.48)	0.93 (0.47)	0.60 (0.30)	0.68 (0.35)
50% ankle and 50% hip co-contraction (BW)	1.35 (0.63)	0.86 (0.48)	0.97 (0.48)	0.64 (0.32)	0.72 (0.36)
100% ankle and 100% hip co-contraction (BW)	1.36 (0.64)	0.87 (0.51)	1.00 (0.50)	0.64 (0.32)	0.73 (0.37)

Table 4.17. Mean (standard deviation) of magnitudes of peak ACL force with different percentages of ankle and hip co-contraction during the side-cutting conditions

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
0% ankle and 0% hip co-contraction (BW)	1.25 (0.73)	0.84 (0.32)	0.91 (0.39)	0.89 (0.39)
50% ankle and 0% hip co-contraction (BW)	1.28 (0.74)	0.90 (0.30)	0.93 (0.39)	0.93 (0.39)
100% ankle and 0% hip co-contraction (BW)	1.33 (0.75)	0.97 (0.29)	0.96 (0.41)	0.98 (0.40)
0% ankle and 50% hip co-contraction (BW)	1.23 (0.70)	0.84 (0.32)	0.87 (0.39)	0.87 (0.40)
0% ankle and 100% hip co-contraction (BW)	1.22 (0.69)	0.84 (0.32)	0.85 (0.39)	0.86 (0.40)
50% ankle and 50% hip co-contraction (BW)	1.26 (0.71)	0.89 (0.31)	0.89 (0.39)	0.91 (0.40)
100% ankle and 100% hip co-contraction (BW)	1.28 (0.71)	0.96 (0.30)	0.89 (0.41)	0.95 (0.42)

Table 4.18. Mean (standard deviation) of timing of peak ACL forces with different percentages of ankle and hip co-contraction during the stop-jump conditions

	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion	Soft Landing
0% ankle and 0% hip co-contraction (BW)	35.20 (10.64)	51.00 (32.61)	42.68 (18.49)	54.97 (42.23)	49.69 (27.52)
50% ankle and 0% hip co-contraction (BW)	36.31 (11.74)	50.94 (32.60)	40.58 (14.58)	53.08 (40.73)	49.45 (27.42)
100% ankle and 0% hip co-contraction (BW)	36.83 (11.72)	49.18 (33.50)	40.10 (14.29)	50.38 (40.67)	48.24 (27.43)
0% ankle and 50% hip co-contraction (BW)	34.04 (8.61)	51.80 (35.23)	43.05 (18.79)	54.93 (43.02)	49.58 (26.65)
0% ankle and 100% hip co-contraction (BW)	32.95 (10.67)	55.64 (36.54)	42.02 (19.08)	57.68 (42.37)	49.95 (28.19)
50% ankle and 50% hip co-contraction (BW)	34.73 (10.94)	52.28 (33.80)	41.61 (18.48)	54.67 (42.34)	48.90 (26.83)
100% ankle and 100% hip co-contraction (BW)	34.09 (11.81)	54.43 (36.92)	41.95 (18.61)	54.74 (42.46)	48.40 (26.92)

Table 4.19. Mean (standard deviation) of timing of peak ACL forces with different percentages of ankle and hip co-contraction during the side-cutting conditions

	Cut Max Speed	Cut 60% Max Speed	Increased Flexion	Soft Landing
0% ankle and 0% hip co-contraction (BW)	38.72 (18.68)	88.89 (21.08)	60.02 (43.85)	56.51 (27.71)
50% ankle and 0% hip co-contraction (BW)	41.63 (22.84)	94.26 (20.87)	61.04 (43.78)	56.75 (26.77)
100% ankle and 0% hip co-contraction (BW)	43.71 (22.64)	96.23 (23.24)	60.92 (43.86)	58.54 (26.97)
0% ankle and 50% hip co-contraction (BW)	39.88 (19.91)	85.29 (22.82)	60.38 (44.92)	58.12 (38.67)
0% ankle and 100% hip co-contraction (BW)	38.24 (19.48)	83.23 (25.34)	60.68 (46.15)	56.63 (40.07)
50% ankle and 50% hip co-contraction (BW)	39.54 (19.44)	88.66 (19.62)	60.71 (44.75)	57.32 (33.97)
100% ankle and 100% hip co-contraction (BW)	38.99 (18.36)	87.67 (24.06)	60.19 (45.02)	55.38 (39.29)

4.6. Specific Aim 1: Effects of Performance Demands on ACL Loading

4.6.1. Stop-Jump

Performance Measures

Results were presented with p values for interaction or main effects followed by 95% confidence interval (CI) for differences of pair-wise comparisons.

An interaction effect of condition and gender was present for jump height ($p < 0.001$). Post-hoc testing showed that jump height during jumping for maximum height condition was higher than that during jumping fast (95% CI for difference: [0.08, 0.12 m]) and jumping for 60% of maximum height [0.15, 0.20 m] conditions for both males and females. Post-hoc testing also showed that jump height during jumping fast condition was higher than that during jumping for 60% of maximum height condition for males only. Males had higher jump height than females [0.09, 0.17 m] during all three jumping conditions. The actual jump height during the jumping for 60% of maximum jump height condition was 63.5% of the jump height during the jumping for maximum height condition.

No interaction effect of condition and gender was present for stance time ($p = 0.932$). A condition effect was present for stance time ($p < 0.001$). Post-hoc testing showed that stance time during jumping fast condition was shorter than that during jumping for maximum height [-114.4, -76.3 ms] and jumping for 60% of maximum height [-91.7, -57.3 ms] conditions for both males and females. No gender effect was present for stance time ($p = 0.064$, [-1.4, 47.0 ms]).

Table 4.20. Means (standard deviations) and P-values for ANOVAs for performance variables during jumping fast, jumping for maximum height, and jumping for 60% of maximum jump height conditions

	Jump Fast		Jump Max Height		Jump 60% of Max Height		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
Jump Height (m)	0.45 (0.09)	0.31 (0.07)	0.57 (0.10)	0.39 (0.05)	0.34 (0.06)	0.27 (0.06)	<0.001	<0.001	<0.001
Stance Time (ms)	233.83 (27.66)	212.09 (38.46)	331.76 (54.20)	304.86 (68.42)	307.38 (49.56)	287.53 (46.23)	0.932	<0.001	0.064

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No interaction effect of condition and gender was present for peak posterior GRF ($p = 0.104$, Table 4.21, Figure B.1, Figure B.2). A condition effect was present for peak posterior GRF ($p < 0.001$). Post-hoc testing showed that peak posterior GRF during jumping fast condition was greater than that during jumping for 60% of maximum height [0.22, 0.36 BW] and jumping for maximum height [0.14, 0.31 BW] conditions for both males and females. Peak posterior GRF during jumping for maximum height condition was greater than that during jumping for 60% of maximum height [0.00, 0.11 BW] condition for both males and females. No gender effect was present for peak posterior GRF ($p = 0.787$, [-0.13, 0.16 BW]).

No interaction effect of condition and gender was present for timing of peak posterior GRF ($p = 0.120$). A condition effect was present for timing of peak posterior GRF ($p = 0.004$). Post-hoc testing showed that timing of peak posterior GRF during jumping fast condition was earlier than that during jumping for 60% of maximum height [-6.0, -1.0 ms] and jumping for maximum height conditions [-5.5, -1.3 ms] for both males and females. No gender effect was present for timing of peak posterior GRF ($p = 0.449$, [-3.4, 7.6 ms]).

No interaction effect of condition and gender was present for vertical GRF at peak posterior GRF ($p = 0.078$, Figure B.3, Figure B.4). A condition effect was present for vertical GRF at peak posterior GRF ($p < 0.001$). Post-hoc testing showed that vertical GRF at peak posterior GRF during jumping fast condition was greater than that during jumping for 60% of maximum height [0.47, 0.78 BW] and jumping for maximum height [0.26, 0.64 BW] conditions for both males and females. Vertical

GRF at peak posterior GRF during jumping for maximum height condition was greater than that during jumping for 60% of maximum height [0.04, 0.31 BW] condition for both males and females. No gender effect was present for vertical GRF at peak posterior GRF ($p = 0.225$, [-0.13, 0.51 BW]).

No interaction effect of condition and gender was present for initial knee flexion angle ($p = 0.161$). Condition and gender effects were found for initial knee flexion angle ($p < 0.001$, $p < 0.001$, Figure B.5, Figure B.6). Post-hoc testing showed that initial knee flexion angle during jumping for 60% of maximum height condition was less than that during jumping fast [-6.5, -2.2 deg] and jumping for maximum height conditions [-7.9, -3.4 deg] for both males and females. Males had greater initial knee flexion angle than females during all three stop-jump conditions [3.7, 11.7 deg].

No interaction effect of condition and gender was present for maximum knee flexion angle ($p = 0.324$). Condition and gender effects were found for maximum knee flexion angle ($p < 0.001$, $p = 0.006$). Post-hoc testing showed that maximum knee flexion angle during jumping fast condition was less than that during jumping for 60% of maximum height [-11.7, -5.9 deg] and jumping for maximum height [-18.1, -12.1 deg] conditions for both males and females. Maximum knee flexion angle during jumping for 60% of maximum height condition was less than that during jumping for maximum height [-9.7, -3.0 deg] condition for both males and females. Males had greater maximum knee flexion angle than females during all three stop-jump conditions [2.1, 11.3 deg].

No interaction effect of condition and gender was present for range of motion of knee flexion angle ($p = 0.969$). A condition effect was present for range of motion of knee flexion angle ($p < 0.001$). Post-hoc testing showed that range of motion of knee flexion angle during jumping fast condition was less than that during jumping for 60% of maximum height [-16.4, 9.8 deg] and jumping for maximum height [-16.9, 10.7 deg] conditions for both males and females. No gender effect was present for range of motion of knee flexion angle ($p = 0.642$, [-5.4, 3.4 deg]).

No interaction effect of condition and gender was present for initial knee flexion velocity ($p = 0.759$, Figure B.7, Figure B.8). Condition and gender effects were found for initial knee flexion velocity ($p = 0.007$, $p = 0.003$). Post-hoc testing showed that initial knee flexion velocity during jumping fast condition was less than that during jumping for maximum height [-97.2, -14.7 deg/s] and jumping for 60% of maximum jump height [-96.9, -14.5 deg/s] conditions for both males and females. Females had greater initial knee flexion velocity than males during all three stop-jump conditions [35.5, 162.2 deg/s].

No interaction effect of condition and gender was present for knee flexion velocity at peak posterior GRF ($p = 0.149$). Condition and gender effects were found for knee flexion velocity at peak posterior GRF ($p = 0.001$). Post-hoc testing showed that knee flexion velocity at peak posterior GRF during jumping fast condition was less than that during jumping for maximum height [-80.1, -11.0 deg/s] and jumping for 60% of maximum jump height [-92.7, -24.9 deg/s] conditions for both males and females. No gender effect was present for knee flexion velocity at peak posterior GRF ($p = 0.064$, [-2.7, 92.3 deg/s]).

An interaction effect of condition and gender was present for knee flexion angle at peak posterior GRF (p = 0.049). Post-hoc testing showed that knee flexion angle at peak posterior GRF during jumping for 60% of maximum height condition was less than that during jumping for maximum height conditions for males. Males had greater knee flexion angle at peak posterior GRF than females during jumping fast and jumping for maximum height conditions.

No interaction effect of condition and gender was present for knee internal rotation angle at peak posterior GRF (p = 0.329, Figure B.9, Figure B.10). A condition effect was present for knee internal rotation angle at peak posterior GRF (p = 0.010). Subjects demonstrated external rotation angle at peak posterior GRF. Post hoc test showed that knee external rotation angle at peak posterior GRF during jumping for maximum height condition was less than that during jumping fast [-1.7, -0.3 deg] and jumping for 60% of maximum height [-1.4, -0.1 deg] conditions for both males and females. No gender effect was present for knee internal rotation angle at peak posterior GRF (p = 0.152, [-4.3, 0.7 deg]).

An interaction effect of condition and gender was present for knee flexion moment at peak posterior GRF (p = 0.013, Figure B.13, Figure B.14). Subjects demonstrated extension moment at peak posterior GRF. Post-hoc testing showed that knee extension moment at peak posterior GRF during jumping fast condition was greater than that during jumping for 60% of maximum height [0.03, 0.05 BW*BH] and jumping for maximum height [0.01, 0.04 BW*BH] conditions for males. No gender effect was present for knee extension moment at peak posterior GRF [-0.01, 0.03 BW*BH].

No interaction or main effect was present for knee varus angle at peak posterior GRF, knee internal rotation moment at peak posterior GRF, or knee varus moment at peak posterior GRF ($p > 0.050$, Figure B.11, Figure B.12, Figure B.15, Figure B.16, Figure B.17, Figure B.18).

Table 4.21. Means (standard deviations) and P-Values for ANOVAs for ACL loading factor variables during jumping fast, jumping for maximum height, and jumping for 60% of maximum jump height conditions

	Jump Fast		Jump Max Height		Jump 60% Max Height		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
PPGRF (BW)	-0.93 (0.29)	-0.82 (0.25)	-0.64 (0.26)	-0.65 (0.26)	-0.57 (0.21)	-0.61 (0.18)	0.104	<0.001	0.787
Time_PPGRF (ms)	29.30 (8.30)	25.84 (9.39)	30.60 (8.51)	31.27 (10.11)	32.80 (8.14)	29.35 (9.62)	0.120	0.004	0.449
VGRF_PPGRF (BW)	2.23 (0.70)	1.82 (0.52)	1.63 (0.63)	1.52 (0.46)	1.44 (0.48)	1.37 (0.44)	0.078	<0.001	0.225
Ini_KF (Deg)	27.85 (8.45)	19.24 (6.56)	29.39 (6.72)	20.36 (6.05)	21.95 (7.65)	16.48 (5.59)	0.161	<0.001	<0.001
Max_KF (Deg)	62.65 (5.36)	54.65 (9.42)	77.78 (11.28)	68.76 (9.44)	69.45 (6.67)	65.41 (8.05)	0.324	<0.001	0.006
ROM_KF (Deg)	34.80 (7.31)	35.42 (8.14)	48.40 (8.45)	49.40 (10.41)	47.50 (8.78)	48.92 (8.12)	0.969	<0.001	0.642
Ini_KFV (Deg/s)	109.55 (150.40)	209.95 (93.37)	173.92 (104.75)	257.50 (87.65)	159.17 (112.47)	271.78 (93.65)	0.759	0.007	0.003
KFV_PPGRF (Deg)	413.71 (125.58)	426.38 (77.73)	427.02 (79.37)	504.07 (80.26)	456.53 (105.31)	501.20 (53.65)	0.149	0.001	0.064
KF_PPGRF (Deg)	35.95 (5.22)	27.12 (7.67)	38.51 (7.26)	31.70 (6.74)	31.81 (6.14)	27.38 (7.16)	0.049	<0.001	0.002
KIR_PPGRF (Deg)	-1.86 (5.14)	-0.04 (2.23)	-1.11 (5.01)	1.19 (3.52)	-1.35 (3.77)	-0.04 (2.72)	0.329	0.010	0.152
KVA_PPGRF (Deg)	-1.31 (5.05)	-1.05 (6.05)	-1.00 (5.20)	0.12 (7.07)	-1.24 (4.53)	-0.12 (6.60)	0.588	0.299	0.658
KFM_PPGRF (BW*BH)	-0.11 (0.05)	-0.07 (0.03)	-0.06 (0.04)	-0.07 (0.03)	-0.06 (0.03)	-0.05 (0.03)	0.013	<0.001	0.240
KIRM_PPGRF (BW*BH)	0.00 (0.05)	-0.01 (0.06)	-0.01 (0.03)	-0.01 (0.04)	-0.01 (0.02)	-0.01 (0.03)	0.653	0.241	0.749
KVAM_PPGRF (BW*BH)	0.00 (0.06)	-0.01 (0.03)	-0.01 (0.04)	-0.01 (0.02)	-0.01 (0.04)	-0.01 (0.03)	0.962	0.479	0.951

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion: Ini_KFV: Initial knee flexion velocity; Max_KFV: Maximum knee flexion velocity; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF.

Peak ACL Force

No interaction of condition and gender was present for peak ACL force ($p = 0.272$, Table 4.22). A condition effect was present for peak ACL force ($p < 0.001$). Post-hoc testing showed that peak ACL force during jumping fast condition was greater than that during jumping for 60% of maximum height [0.24, 0.52 BW] and jumping for maximum height [0.32, 0.62 BW] conditions for both males and females. No gender effect was present for peak ACL force ($p = 0.477$, [-0.43, 0.21 BW]).

No interaction of condition and gender was present for timing of peak ACL force ($p = 0.182$). A condition effect was present for timing of peak ACL force ($p = 0.006$). Timing of peak ACL force during jumping fast condition was earlier than that during jumping for maximum height [-26.9, -4.7 ms] and jumping for 60% of maximum height [-14.3, -0.6 ms] conditions for both males and females. Timing of peak ACL force during jumping for 60% of maximum height condition was earlier than that during jumping for maximum height condition for both males and females [-15.7, -0.9 ms]. No gender effect was present for timing of peak ACL force ($p = 0.314$, [-5.7, 17.1 ms]).

No interaction of condition and gender was present for peak ACL force caused by anterior shear force ($p = 0.052$). Condition and gender effects were found for peak ACL force caused by anterior shear force ($p < 0.001$, $p = 0.025$). Post-hoc testing showed that peak ACL force caused by anterior shear force during jumping fast condition was greater than that during jumping for maximum height [0.17, 0.45 BW] and jumping for 60% of maximum jump height [0.14, 0.30 BW] conditions for

both males and females. Females had greater peak ACL force caused by anterior shear force than males during all three stop-jump conditions [0.04, 0.57 BW].

No interaction or main effect was present for peak ACL force caused by internal - external rotation moment ($p > 0.050$). The contribution of internal rotation moment to peak ACL force was small.

No interaction of condition and gender was present for peak ACL force caused by valgus - varus moment ($p = 0.173$). Condition and gender effects were found for peak ACL force caused by valgus - varus moment ($p = 0.003$, $p = 0.027$). Peak ACL force caused by valgus - varus moment during jumping fast condition was greater than that during jumping for maximum height [0.05, 0.29 BW] and jumping for 60% of maximum height [0.06, 0.28 BW] conditions for both males and females. Males had greater peak ACL force caused by valgus - varus moment than females during all three stop-jump conditions [0.03, 0.40 BW].

Table 4.22. Means (standard deviations) and P-Values for ANOVAs for ACL force variables during jumping fast, jumping for maximum height, and jumping for 60% of maximum jump height conditions

	Jump Fast		Jump Max Height		Jump 60% Max Height		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
ACL_Peak (BW)	1.29 (0.67)	1.39 (0.56)	0.78 (0.46)	0.99 (0.49)	0.96 (0.52)	0.97 (0.42)	0.272	<0.001	0.477
Timing_ACL (ms)	34.25 (9.20)	36.16 (12.11)	58.14 (40.00)	43.87 (21.95)	45.08 (21.14)	40.28 (15.64)	0.182	0.006	0.314
ACL_AS (BW)	0.70 (0.49)	0.99 (0.48)	0.31 (0.35)	0.77 (0.47)	0.54 (0.44)	0.71 (0.40)	0.052	<0.001	0.025
ACL_IR (BW)	-0.03 (0.27)	0.10 (0.51)	0.07 (0.29)	0.01 (0.13)	0.05 (0.17)	0.05 (0.15)	0.149	0.903	0.817
ACL_VV (BW)	0.62 (0.58)	0.30 (0.27)	0.38 (0.23)	0.21 (0.19)	0.36 (0.29)	0.21 (0.23)	0.173	0.003	0.027

Note: ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external rotation moment; ACL_VV: Peak ACL force caused by valgus – varus moment; Timing_ACL: Timing of peak ACL force.

4.6.2. Side-Cutting

Performance Measures

An interaction effect of condition and gender was present for approach speed ($p = 0.003$, Table 4.23). Post-hoc testing showed that approach speed during cutting with maximum speed condition was faster than that during cutting with 60% of maximum speed condition for both males and females [1.6, 1.9 m/s]. Males had faster approach speed than females during cutting with maximum speed condition only. The actual approach speed during cutting with 60% of maximum speed condition was 55 % of the approach speed during cutting with maximum speed condition.

An interaction effect of condition and gender was present for take-off speed ($p = 0.002$). Post-hoc testing showed that take-off speed during cutting with maximum speed condition was faster than that during cutting with 60% of maximum speed condition for both males and females [1.7, 2.0 m/s]. Males had faster take-off speed than females during cutting with maximum speed condition only. The actual take-off speed during cutting with 60% of maximum speed condition was 55 % of the approach speed during cutting with maximum speed condition.

No interaction effect of condition and gender was present for stance time ($p = 0.234$). A condition effect was present for stance time ($p < 0.001$). Stance time during cutting with maximum speed condition was shorter than that during cutting with 60% of maximum speed condition for both males and females [-170.7 ms, -111.0 ms]. No gender effect was present for stance time ($p = 0.918$, [-43.7, 39.5 ms]).

Table 4.23. Means (standard deviations) and P-Values for ANOVAs for performance variables during cutting with maximum speed and cutting with 60% of maximum speed conditions

	Cut Max Speed		Cut 60% Max Speed		P-Value		
Variables	Male	Female	Male	Female	Interaction	Condition	Gender
Approach speed (m/s)	4.01 (0.33)	3.68 (0.31)	2.07 (0.27)	2.14 (0.39)	0.003	<0.001	0.164
Take-off speed (m/s)	4.36 (0.43)	4.01 (0.26)	2.27 (0.32)	2.36 (0.36)	0.002	<0.001	0.173
Stance Time (ms)	290.58 (54.29)	306.26 (33.22)	449.21 (124.72)	429.29 (56.99)	0.234	<0.001	0.918

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No interaction effect of condition and gender was present for peak posterior GRF ($p = 0.134$, Table 4.24, Figure B.19, Figure B.20). A condition effect was present for peak posterior GRF ($p < 0.001$). Peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.33, 0.50 BW]. No gender effect was present for peak posterior GRF ($p = 0.905$, [-0.13, 0.15 BW]).

No interaction effect of condition and gender was present for timing of peak posterior GRF ($p = 0.292$). A condition effect was present for timing of peak posterior GRF ($p = 0.002$). Timing of peak posterior GRF during cutting with maximum speed condition was earlier than that during cutting with 60% maximum speed condition for both males and females [-13.7, -3.3 ms]. No gender effect was present for timing of peak posterior GRF ($p = 0.964$, [-5.7, 5.4 ms]).

No interaction effect of condition and gender was present for vertical GRF at peak posterior GRF ($p = 0.349$, Figure B.21, Figure B.22). A condition effect was present for vertical GRF at peak posterior GRF ($p < 0.001$). Vertical GRF at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.63, 1.20 BW]. No gender effect was present for vertical GRF at peak posterior GRF ($p = 0.400$, [-0.52, 0.21 BW]).

No interaction effect of condition and gender was present for initial knee flexion ($p = 0.051$, Figure B.23, Figure B.24). A condition effect was present for initial knee flexion ($p < 0.001$).Post-hoc testing showed that initial knee flexion during

cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [10.8, 16.0 deg]. No gender effect was present for initial knee flexion ($p = 0.654$, [-2.9, 4.5 deg]).

No interaction effect of condition and gender was present for maximum knee flexion ($p = 0.051$). A condition effect was present for maximum knee flexion ($p < 0.001$). Post-hoc testing showed that maximum knee flexion during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [3.5, 7.7 deg]. No gender effect was present for maximum knee flexion ($p = 0.097$, [-0.6, 7.0 deg]).

No interaction effect of condition and gender was present for range of motion of knee flexion ($p = 0.720$). A condition effect was present for range of motion of knee flexion ($p < 0.001$). Range of motion of knee flexion during cutting with maximum speed condition was less than that during cutting with 60% maximum speed condition for both males and females [-10.6, -5.1 deg]. No gender effect was present for range of motion of knee flexion ($p = 0.115$, [-1.0, 9.1 deg]).

No interaction effect of condition and gender was present for initial knee flexion velocity ($p = 0.34$, Figure B.25, Figure B.26). A condition effect was present for initial knee flexion velocity ($p < 0.001$). Initial knee flexion velocity during cutting with maximum speed condition was less than that during cutting with 60% of maximum speed condition for both males and females [-144.8, -67.8 deg/s]. No gender effect was present for initial knee flexion velocity ($p = 0.148$, [-18.6 118.1 deg/s]).

No interaction or main effect was present for knee flexion velocity at peak posterior GRF ($p > 0.050$).

An interaction effect of condition and gender was present for knee flexion at peak posterior GRF ($p = 0.031$). Post-hoc testing showed that knee flexion at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [8.6, 12.9 deg].

No interaction effect of condition and gender was present for knee internal rotation angle at peak posterior GRF ($p = 0.063$, Figure B.27, Figure B.28). A condition effect was present for knee internal rotation angle at peak posterior GRF ($p = 0.001$). Subjects demonstrated external rotation angle at peak posterior GRF during both cutting conditions. The external rotation angle at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.4, 1.5 deg]. No gender effect was present for knee internal rotation angle at peak posterior GRF ($p = 0.142$, [-2.9, 0.4 deg]).

No interaction effect of condition and gender was present for knee varus angle at peak posterior GRF ($p = 0.942$, Figure B.29, Figure B.30). A condition effect was present for knee varus angle at peak posterior GRF ($p < 0.001$). Subjects demonstrated valgus angle at peak posterior GRF during both cutting conditions. The valgus angle at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition

for both males and females [1.2, 2.6 deg]. No gender effect was present for knee varus angle at peak posterior GRF ($p = 0.815$, [-3.8, 3.0]).

No interaction effect of condition and gender was present for knee flexion moment at peak posterior GRF ($p = 0.183$, Figure B.31, Figure B.32). A condition effect was present for knee flexion moment at peak posterior GRF ($p < 0.001$). Subjects demonstrated extension moment at peak posterior GRF during both cutting conditions. The extension moment at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.02, 0.06 BW*BH]. No gender effect was present for knee flexion moment at peak posterior GRF ($p = 0.607$, [-0.03, 0.02 BW*BH]).

No interaction effect of condition and gender was present for knee internal rotation moment at peak posterior GRF ($p = 0.687$, Figure B.33, Figure B.34). A condition effect was present for knee internal rotation moment at peak posterior GRF ($p < 0.001$). The knee internal rotation moment at peak posterior GRF during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.03, 0.08 BW*BH]. No gender effect was present for knee internal rotation moment at peak posterior GRF ($p = 0.108$, [-0.05, 0.01 BW*BH]).

No interaction effect of condition and gender was present for knee varus moment at peak posterior GRF ($p = 0.416$, Figure B.35, Figure B.36). A condition effect was present for knee varus moment at peak posterior GRF ($p < 0.001$). Subjects demonstrated varus moment during the cutting with maximum speed

condition and valgus moment during the cutting with 60% maximum speed condition [0.01, 0.04 BW*BH]. No gender effect was present for knee varus moment at peak posterior GRF ($p = 0.054$, [0.00, 0.04 BW*BH]).

Table 4.24. Means (standard deviations) and P-Values for ANOVAs for ACL loading factor variables during cutting with maximum speed and cutting with 60% of maximum speed conditions

	Cut Max Speed		Cut 60% Max Speed		P-Value		
Variables	Male	Female	Male	Female	Interaction	Condition	Gender
PPGRF (BW)	-0.73 (0.30)	-0.67 (0.28)	-0.25 (0.14)	-0.32 (0.24)	0.134	<0.001	0.905
Time_PPGRF (ms)	25.23 (11.52)	28.10 (8.09)	36.50 (14.25)	33.88 (10.30)	0.292	0.002	0.964
VGRF_PPGRF (BW)	1.93 (0.81)	1.95 (0.88)	0.88 (0.36)	1.17 (0.58)	0.349	<0.001	0.400
Ini_KF (Deg)	25.70 (8.26)	22.30 (4.34)	9.71 (6.61)	11.47 (6.80)	0.051	<0.001	0.654
Max_KF (Deg)	49.12 (7.66)	50.25 (3.62)	41.47 (7.71)	46.76 (5.91)	0.051	<0.001	0.097
ROM_KF (Deg)	23.44 (10.12)	27.95 (6.42)	31.76 (8.72)	35.29 (8.27)	0.720	<0.001	0.115
Ini_KFV (Deg/s)	-56.68 (139.16)	-8.49 (115.50)	48.01 (112.87)	99.36 (90.38)	0.934	<0.001	0.148
KFV_PPGRF (Deg)	251.51 (180.21)	311.91 (94.89)	271.40 (91.73)	313.66 (97.23)	0.753	0.708	0.084
KF_PPGRF (Deg)	29.27 (6.09)	26.77 (3.43)	16.16 (7.76)	18.40 (5.77)	0.031	<0.001	0.939
KIR_PPGRF (Deg)	-2.91 (3.81)	-1.17 (2.10)	-1.47 (2.29)	-0.73 (1.65)	0.063	0.001	0.142
KVA_PPGRF (Deg)	-2.68 (3.93)	-2.32 (5.82)	-0.80 (3.91)	-0.38 (6.22)	0.942	<0.001	0.815
KFM_PPGRF (BW*BH)	-0.07 (0.06)	-0.05 (0.04)	-0.02 (0.02)	-0.03 (0.04)	0.183	<0.001	0.607
KIRM_PPGRF (BW*BH)	0.06 (0.07)	0.08 (0.07)	0.00 (0.03)	0.03 (0.03)	0.687	<0.001	0.108
KVAM_PPGRF (BW*BH)	0.02 (0.04)	0.00 (0.04)	-0.01 (0.01)	-0.02 (0.02)	0.416	<0.001	0.054

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion: Ini_KFV: Initial knee flexion velocity; Max_KFV: Maximum knee flexion velocity; KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF.

Peak ACL Force

No interaction effect of condition and gender was present for peak ACL force ($p = 0.241$, Table 4.25). A condition effect was present for peak ACL force ($p < 0.001$). Peak ACL force during cutting with maximum speed condition was greater than that during cutting with 60% of maximum speed condition for both males and females [0.18, 0.65 BW]. No gender effect was present for peak ACL force ($p = 0.875$, [-0.33, 0.28 BW]).

No interaction effect of condition and gender was present for timing of peak ACL force ($p = 0.064$). A condition effect was present for timing of peak ACL force ($p < 0.001$). Timing of peak ACL force during cutting with maximum speed condition was earlier than that during cutting with 60% maximum speed condition for both males and females [-59.8, 40.5 ms]. No gender effect was present for timing of peak ACL force ($p = 0.686$, [-11.0, 7.3 ms]).

No interaction effect of condition and gender was present for peak ACL force caused by varus - valgus moment ($p = 0.424$). A condition effect was present for peak ACL force caused by varus - valgus moment ($p < 0.001$). Peak ACL force caused by varus - valgus moment during cutting with maximum speed condition was greater than that during cutting with 60% maximum speed condition for both males and females [0.20, 0.63 BW]. No gender effect was present for peak ACL force caused by varus - valgus moment ($p = 0.320$, [-0.12 0.36 BW]).

No interaction or main effect was present for peak ACL force caused by anterior shear force or peak ACL force caused by internal - external rotation moments ($p > 0.050$).

Table 4.25. Means (standard deviations) and P-Values for ANOVAs for ACL force variables during cutting with maximum speed and cutting with 60% of maximum speed conditions

	Cut Max Speed		Cut 60% Max Speed		P-Value		
Variables	Male	Female	Male	Female	Interaction	Condition	Gender
ACL_Peak (BW)	1.31 (0.92)	1.19 (0.50)	0.76 (0.24)	0.92 (0.37)	0.241	0.001	0.875
Timing_ACL (ms)	33.25 (17.02)	44.18 (19.12)	92.52 (15.29)	85.26 (25.55)	0.064	<0.001	0.686
ACL_AS (BW)	0.64 (0.63)	0.76 (0.63)	0.61 (0.29)	0.81 (0.45)	0.780	0.937	0.160
ACL_IR (BW)	-0.02 (0.41)	-0.06 (0.32)	-0.04 (0.07)	-0.05 (0.19)	0.808	0.931	0.757
ACL_VV (BW)	0.69 (0.77)	0.49 (0.51)	0.19 (0.15)	0.16 (0.12)	0.424	<0.001	0.320

Note: ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external rotation moment; ACL_VV: Peak ACL force caused by valgus-varus moment; Time_ACL: Timing of peak ACL force.

4.7. Specific Aim 2: Movement Pattern Effects on Performance

4.7.1. Stop-jump

Performance Measures

No interaction or main effect was present for ankle work ($p > 0.050$, Table 4.26).

No interaction effect of condition and gender was present for knee work ($p = 0.922$). A condition effect was present for knee work ($p < 0.001$). Post-hoc testing showed that knee work during increased knee flexion landing condition was greater than that during jumping for maximum height [0.03 0.05 J/BW*BH] and soft landing condition [0.01, 0.04 J/BW*BH] for both males and females. Knee work during soft landing condition was greater than that during jumping for maximum height condition for both males and females [0.00, 0.02 J/BW*BH]. No gender effect was present for knee work ($p = 0.212$, [-0.01, 0.03 J/BW*BH]).

No interaction effect of condition and gender was present for hip work ($p = 0.978$). Condition and gender effects were found for hip work ($p < 0.001$, $p = 0.017$). Post-hoc testing showed that hip work during increased knee flexion landing condition was greater than that during jumping for maximum height [0.04, 0.07 J/BW*BH] and soft landing [0.02, 0.05 J/BW*BH] conditions for both males and females. Hip work during soft landing condition was greater than that during jumping for maximum height condition for both males and females [0.01, 0.03 J/BW*BH].

Males had greater hip work than females during all three stop-jump conditions [0.01 0.07 J/BW*BH].

No interaction effect of condition and gender was present for total work ($p = 0.921$). Condition and gender effects were found for total work ($p < 0.001$, $p = 0.018$). Post-hoc testing showed that total work during increased knee flexion landing condition was greater than that during jumping for maximum height [0.07, 0.11 J/BW*BH] and soft landing [0.03, 0.08 J/BW*BH] condition for both males and females. Total work during soft landing condition was greater than that during jumping for maximum height condition for both males and females [0.02, 0.05 J/BW*BH]. Males had greater total work than females during all three stop-jump conditions [0.01, 0.08 J/BW*BH].

No interaction effect of condition and gender was present for approach speed ($p = 0.339$). A condition effect was present for approach speed ($p < 0.001$). Post-hoc testing showed that the approach speed during soft landing was less than the approach speed during jumping for maximum height [-0.33, -0.17 m/s] and increased knee flexion landing [-0.25, -0.09] conditions for both males and females. No gender effect was present for approach speed ($p = 0.327$, [-0.10, 0.29 m/s]).

No interaction effect of condition and gender was present for jump height ($p = 0.057$). Condition and gender effects were found for jump height ($p < 0.001$, $p < 0.001$). Post-hoc testing showed that the jump height during jumping for maximum height condition was greater than that during increased knee flexion landing [0.02, 0.05 m] and soft landing [0.03, 0.04 m] conditions for both males and females. Males

had greater jump height than females during all three stop-jump conditions [0.11, 0.21 m].

No interaction effect of condition and gender was present for stance time ($p = 0.782$). A condition effect was present for stance time ($p < 0.001$). Post-hoc testing showed that stance time during increased knee flexion landing condition was longer than that during jumping for maximum height [100.8, 159.4 ms] and soft landing [29.7, 93.5 ms] condition for both males and females. Stance time during soft landing condition was longer than that during jumping for maximum height condition for both males and females [46.0, 91.0 ms]. No gender effect was present for stance time ($p = 0.244$, [-14.9, 56.7 ms]).

Table 4.26. Means (standard deviations) and P-Values for ANOVAs for performance variables during jumping for maximum height, jump with increased knee flexion landing, and soft landing conditions

	Jump Max Height		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
Ankle Work (J/BW/BH)	0.06 (0.01)	0.06 (0.02)	0.06 (0.01)	0.06 (0.02)	0.06 (0.01)	0.06 (0.02)	0.453	0.352	0.455
Knee Work (J/BW/BH)	0.12 (0.03)	0.11 (0.02)	0.16 (0.04)	0.15 (0.05)	0.13 (0.02)	0.12 (0.03)	0.922	<0.001	0.212
Hip Work (J/BW/BH)	0.12 (0.04)	0.08 (0.04)	0.17 (0.07)	0.14 (0.04)	0.14 (0.06)	0.10 (0.04)	0.978	<0.001	0.017
Total Work (J/BW/BH)	0.30 (0.06)	0.25 (0.05)	0.39 (0.09)	0.34 (0.06)	0.33 (0.07)	0.29 (0.05)	0.921	<0.001	0.018
Approach speed (m/s)	2.42 (0.36)	2.26 (0.24)	2.28 (0.42)	2.24 (0.23)	2.13 (0.36)	2.05 (0.24)	0.339	<0.001	0.327
Jump Height (m)	0.57 (0.10)	0.39 (0.05)	0.52 (0.09)	0.37 (0.04)	0.52 (0.10)	0.37 (0.05)	0.057	<0.001	<0.001
Stance Time (ms)	331.76 (54.20)	304.86 (68.42)	453.24 (72.78)	443.58 (79.50)	399.91 (78.34)	373.75 (72.50)	0.782	<0.001	0.244

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No interaction effect of condition and gender was present for peak posterior GRF ($p = 0.620$, Table 4.27, Figure B.37, Figure B.38). A condition effect was present for peak posterior GRF ($p = 0.007$). Post-hoc testing showed that peak posterior GRF during soft landing condition was less than that during jumping for maximum height condition for both males and females [-0.05, 0.16 BW]. No gender effect was present for peak posterior GRF ($p = 0.816$, [-0.13, 0.16]).

No interaction or main effect was present for timing of peak posterior GRF ($p > 0.050$).

No interaction effect of condition and gender was present for vertical GRF at peak posterior GRF ($p = 0.721$, Figure B.39, Figure B.40). A condition effects was present for vertical GRF at peak posterior GRF ($p < 0.001$). Post-hoc testing showed that vertical GRF at peak posterior GRF during jumping for maximum height was greater than that during soft landing [0.24, 0.54 BW] and increased knee flexion landing [0.03, 0.38 BW] conditions for both males and females. Vertical GRF at peak posterior GRF during increased knee flexion landing was greater than that during soft landing for both males and females [0.07, 0.30 BW]. No gender effect was present for vertical GRF at peak posterior GRF ($p = 0.388$, [-0.17, 0.42 BW]).

No interaction effect of condition and gender was present for initial knee flexion angle ($p = 0.815$, Figure B.41, Figure B.42). Condition and gender effects were found for initial knee flexion angle ($p < 0.001$, $p < 0.001$). Post-hoc testing showed that the initial knee flexion angle during increased knee flexion landing condition was greater than that during jumping for maximum height [7.4, 10.8 deg]

and soft landing [7.7, 11.9 deg] conditions for both males and females. Males had greater initial knee flexion angle than females during all three stop-jump conditions [4.7, 13.3 deg].

No interaction effect of condition and gender was present for maximum knee flexion angle ($p = 0.973$). Condition and gender effects were found for maximum knee flexion angle ($p < 0.001$, $p = 0.023$). Post-hoc testing showed that the maximum knee flexion angle during increased knee flexion landing was greater than that during jumping for maximum height [17.7, 27.5 deg] and soft landing [8.8, 19.1 deg] conditions for both males and females. Maximum knee flexion angle during soft landing was greater than that during jumping for maximum height for both males and females [5.3, 11.9 deg]. Males had greater maximum knee flexion angle than females during all three stop-jump conditions [1.1, 14.5 deg].

No interaction effect of condition and gender was present for range of motion of knee flexion ($p = 0.982$). A condition effect was present for range of motion of knee flexion ($p < 0.001$). Post-hoc testing showed that the range of motion of knee flexion during jumping for maximum height condition was less than that during increased knee flexion landing [-18.6, -8.3 deg] and soft landing [-12.7, -5.9 deg] conditions. No gender effect was present for range of motion of knee flexion ($p = 0.698$, [-7.3, 4.9 deg]).

An interaction effect of condition and gender was present for initial knee flexion velocity ($p = 0.023$, Figure B.43, Figure B.44). The initial knee flexion velocity during soft landing condition was greater than that during jumping for maximum

height condition for females. Females had greater initial knee flexion velocity than males during soft landing condition.

No interaction effect of condition and gender was present for knee flexion velocity at peak posterior GRF ($p = 0.746$). No condition effect was present for knee flexion velocity at peak posterior GRF ($p = 0.060$). A gender effect was present for knee flexion velocity at peak posterior GRF ($p = 0.014$). Females had greater knee flexion velocity at peak posterior GRF than males during all three stop-jump conditions [14.2, 116.4 deg/s].

No interaction effect of condition and gender was present for knee flexion angle at peak posterior GRF ($p = 0.547$). Condition and gender effects were found for knee flexion angle at peak posterior GRF ($p < 0.001$, $p = 0.001$). Post-hoc testing showed that the knee flexion angle at peak posterior GRF during increased knee flexion landing condition was greater than that during jumping for maximum height [6.7, 11.1 deg] and soft landing [6.5 10.5 deg] conditions for both males and females. Males had greater knee flexion angle at peak posterior GRF than females during all three stop-jump conditions [3.1, 11.2 deg].

No interaction or main effect was present for knee internal rotation angle or knee varus angle at peak posterior GRF ($p > 0.050$, Figure B.45, Figure B.46, Figure B.47, Figure B.48).

No interaction effect of condition and gender was present for knee flexion moment at peak posterior GRF ($p = 0.172$, Figure B.49, Figure B.50). A condition effect was present for knee flexion moment at peak posterior GRF ($p = 0.023$). Subjects demonstrated knee extension moment at peak posterior GRF. Post-hoc

testing showed that the knee extension moment at peak posterior GRF during soft landing was less than that during jumping for maximum height [-0.02, 0.00 BW*BH] and increased knee flexion landing [-0.02, 0.00 BW*BH] conditions for both males and females. No gender effect was present for knee flexion moment at peak posterior GRF (p = 0.751, [-0.02, 0.02 BW*BH]).

No interaction effect of condition and gender was present for knee internal rotation moment at peak posterior GRF (p = 0.600, Figure B.51, Figure B.52). A condition effect was present for knee internal rotation moment at peak posterior GRF (p = 0.030). Subjects demonstrated external rotation moments at peak posterior GRF. Post-hoc testing showed that the knee external rotation moment during increased knee flexion landing condition was less than that during jumping for maximum height [-0.02, 0.00 BW*BH] and soft landing [-0.01 0.00 BW*BH] conditions for both males and females. No gender effect was present for knee internal rotation moment at peak posterior GRF (p = 0.896, [-0.02, 0.02 BW*BH]).

No interaction effect of condition and gender was present for knee varus moment at peak posterior GRF (p = 0.804, Figure B.53, Figure B.54). A condition effect was present for knee varus moment at peak posterior GRF (p = 0.014). Subjects demonstrated valgus moments at peak posterior GRF during jumping for maximum height condition. Post-hoc testing showed that the knee valgus moment at peak posterior GRF during jumping for maximum height condition was greater than that during increased knee flexion landing [0.00, 0.02 BW*BH] and soft landing [0.00, 0.01 BW*BH] conditions for both males and females. No gender effect was present for knee varus moment at peak posterior GRF (p = 0.952, [-0.02, 0.02 BW*BH]).

Table 4.27. Means (standard deviations) and P-Values for ANOVAs for ACL loading factor variables during jumping for maximum height, jump with increased knee flexion landing, and soft landing conditions

	Jump Max Height		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
PPGRF (BW)	-0.64 (0.26)	-0.65 (0.26)	-0.60 (0.26)	-0.59 (0.20)	-0.52 (0.22)	-0.57 (0.21)	0.620	0.007	0.816
Time_PPGRF (ms)	30.60 (8.51)	31.27 (10.11)	29.94 (7.38)	31.64 (9.85)	31.53 (6.19)	30.59 (8.36)	0.470	0.969	0.851
VGRF_PPGRF (BW)	1.63 (0.63)	1.52 (0.46)	1.46 (0.61)	1.28 (0.40)	1.23 (0.47)	1.14 (0.36)	0.721	<0.001	0.388
Ini_KF (Deg)	29.39 (6.72)	20.36 (6.05)	38.75 (7.30)	29.18 (7.30)	28.37 (8.55)	19.95 (5.81)	0.815	<0.001	<0.001
Max_KF (Deg)	77.78 (11.28)	69.76 (9.44)	100.36 (15.74)	92.30 (14.76)	86.08 (13.21)	78.68 (9.29)	0.973	<0.001	0.023
ROM_KF (Deg)	48.40 (8.45)	49.40 (10.41)	61.61 (14.45)	63.12 (15.45)	57.71 (12.16)	58.72 (10.82)	0.982	<0.001	0.698
Ini_KFV (Deg/s)	173.92 (104.75)	257.50 (87.65)	180.81 (133.63)	195.53 (124.97)	195.35 (132.51)	310.91 (85.42)	0.023	0.003	0.032
KFV_PPGRF (Deg)	427.02 (79.37)	504.07 (80.26)	434.91 (96.44)	487.68 (114.77)	463.41 (112.18)	529.45 (66.93)	0.746	0.060	0.014
KF_PPGRF (Deg)	38.51 (7.26)	31.70 (6.74)	48.16 (7.33)	39.86 (6.65)	38.61 (7.29)	32.37 (5.81)	0.547	<0.001	0.001
KIR_PPGRF (Deg)	-1.11 (5.01)	1.19 (3.52)	-1.31 (5.67)	1.78 (3.67)	-1.34 (4.38)	1.18 (4.69)	0.519	0.645	0.079
KVA_PPGRF (Deg)	-1.00 (5.20)	0.12 (7.07)	-2.34 (4.43)	-0.35 (6.33)	-1.47 (4.58)	-0.07 (6.44)	0.547	0.086	0.426
KFM_PPGRF (BW*BH)	-0.06 (0.04)	-0.07 (0.03)	-0.07 (0.05)	-0.06 (0.02)	-0.06 (0.03)	-0.05 (0.02)	0.172	0.023	0.751
KIRM_PPGRF (BW*BH)	-0.01 (0.03)	-0.01 (0.04)	0.00 (0.03)	0.00 (0.02)	-0.01 (0.02)	-0.01 (0.02)	0.600	0.030	0.896
KVAM_PPGRF (BW*BH)	-0.01 (0.04)	-0.01 (0.02)	0.00 (0.04)	0.00 (0.02)	0.00 (0.03)	0.00 (0.02)	0.804	0.014	0.952

Note: PPGRF: Peak posterior GRF; Time_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion: KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF.

Peak ACL Force

No interaction effect of condition and gender was present for peak ACL force ($p = 0.172$, Table 4.28). A condition effect was present for peak ACL force ($p < 0.001$). Post-hoc testing showed that peak ACL force during jumping for maximum height condition was greater than that during increased knee flexion landing [0.12, 0.35 BW] and soft landing [0.06, 0.28 BW] conditions for both males and females. No gender effect was present for peak ACL force ($p = 0.315$, [-0.35, 0.12 BW]).

No interaction or main effect was present for timing of peak ACL force ($p > 0.050$).

An interaction effect of condition and gender was present for peak ACL force caused by anterior shear force ($p = 0.028$). Post-hoc testing showed that peak ACL force caused by anterior shear force during jumping for maximum height condition was greater than that during increased knee flexion landing condition for females. Females had greater peak ACL force caused by anterior shear force than males during jumping for maximum height conditions.

No interaction or main effect was present for peak ACL force caused by internal - external rotation moments ($p > 0.050$).

No interaction effect of condition and gender was present for peak ACL force caused by valgus - varus moment ($p = 0.522$). No condition effect was present for peak ACL force caused by valgus - varus moment ($p = 0.266$). A gender effect was present for peak ACL force caused by valgus - varus moment ($p = 0.011$). Males had greater peak ACL force caused by valgus - varus moment than females during all three stop-jump conditions [0.04, 0.28 BW].

Table 4.28. Means (standard deviations) and P-Values for ANOVAs for ACL force variables during jumping for maximum height, jump with increased knee flexion landing, and soft landing conditions

	Jump Max Height		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
ACL_Peak (BW)	0.76 (0.46)	0.99 (0.49)	0.62 (0.30)	0.65 (0.33)	0.66 (0.40)	0.75 (0.29)	0.172	<0.001	0.315
Timing_ACL (ms)	58.14 (40.00)	43.87 (21.95)	63.94 (49.91)	46.01 (31.79)	60.95 (31.80)	38.43 (16.71)	0.770	0.631	0.054
ACL_AS (BW)	0.31 (0.35)	0.77 (0.47)	0.20 (0.26)	0.40 (0.27)	0.32 (0.37)	0.57 (0.27)	0.028	<0.001	0.004
ACL_IR (BW)	0.07 (0.29)	0.01 (0.13)	0.03 (0.12)	0.04 (0.09)	0.03 (0.18)	-0.01 (0.07)	0.300	0.516	0.506
ACL_VV (BW)	0.38 (0.23)	0.21 (0.19)	0.40 (0.18)	0.21 (0.23)	0.31 (0.23)	0.19 (0.18)	0.522	0.266	0.011

Note: ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external rotation moment; ACL_VV: Peak ACL force caused by valgus – varus moment; Timing_ACL: Timing of peak ACL force.

4.7.2. Side-cutting

Performance Measures

No interaction effect of condition and gender was present for ankle work ($p = 0.400$, Table 4.29). A condition effect was present for ankle work ($p = 0.003$). Post-hoc testing showed that ankle work during soft landing condition was greater than that during cutting with maximum speed [0.00, 0.01 J/BW*BH] and increased knee flexion landing [0.00, 0.01 J/BW*BH] conditions for both males and females. No gender effect was present for ankle work ($p = 0.626$, [-0.02, 0.01 J/BW*BH]).

No interaction effect of condition and gender was present for knee work ($p = 0.110$). A condition effect was present for knee work ($p < 0.001$). Post-hoc testing showed that knee work during increased knee flexion landing condition was greater than that during cutting with maximum speed [0.02, 0.04 J/BW*BH] and soft landing [0.01, 0.02 J/BW*BH] conditions for both males and females. Knee work during soft landing conditions was greater than that during cutting with maximum speed conditions for both males and females [0.01, 0.02 J/BW*BH]. No gender effect was present for knee work ($p = 0.775$, [-0.02, 0.01 J/BW*BH]).

No interaction effect of condition and gender was present for hip work ($p = 0.429$). A condition effect was present for hip work ($p < 0.001$). Post-hoc testing showed that hip work during increased knee flexion landing condition was greater than that during cutting with maximum speed [0.05, 0.09 J/BW*BH] and soft landing [0.04, 0.08 J/BW*BH] conditions for both males and females. Hip work during soft landing conditions was greater than that during cutting with maximum speed

conditions for both males and females [0.00, 0.03 J/BW*BH]. No gender effect was present for hip work ($p = 0.430$, [-0.02, 0.04 J/BW*BH]).

No interaction effect of condition and gender was present for total work ($p = 0.895$). A condition effect was present for total work ($p < 0.001$). Post-hoc testing showed that total work during increased knee flexion landing condition was greater than that during cutting with maximum speed [0.08, 0.12 J/BW*BH] and soft landing [0.05, 0.09 J/BW*BH] conditions for both males and females. Total work during soft landing conditions was greater than that during cutting with maximum speed conditions for both males and females [0.02, 0.05 J/BW*BH]. No gender effect was present for total work ($p = 0.748$, [-0.03, 0.04 J/BW*BH]).

No interaction effect of condition and gender was present for approach speed ($p = 0.304$). Condition and gender effects were found for approach speed ($p < 0.001$, $p< 0.001$). Post-hoc testing showed that approach speed during cutting with maximum speed condition was faster than that during increased knee flexion landing [0.36, 0.57 m/s] and soft landing [0.54, 0.72 m/s] conditions for both males and females. Post-hoc testing showed that approach speed during increased knee flexion landing condition was faster than that during soft landing conditions for both males and females [0.06, 0.27 m/s]. Males had faster approach speed than females during all three side-cutting conditions [0.19, 0.61 m/s].

No interaction effect of condition and gender was present for take-off speed ($p = 0.110$). Condition and gender effects were found for take-off speed ($p < 0.001$, $p = 0.003$). Post-hoc testing showed that take-off speed during cutting with maximum speed condition was faster than that during increased knee flexion landing [0.40,

0.60 m/s] and soft landing [0.47, 0.68 m/s] conditions for both males and females. Males had faster take-off speed than females during all three side-cutting conditions [0.13, 0.61 m/s].

No interaction effect of condition and gender was present for stance time ($p = 0.776$). A condition effect was present for stance time ($p < 0.001$). Post-hoc testing showed that stance time during increased knee flexion landing was longer than that during cutting with maximum speed [118.0, 166.9 ms] and soft landing [31.3, 75.9 ms] conditions for both males and females. Stance time during soft landing condition was longer than that during cutting with maximum speed condition for both males and females [64.0, 113.7 ms]. No gender effect was present for stance time ($p = 0.347$, [-65.0, 23.5 ms]).

Table 4.29. Means (standard deviations) and P-Values for ANOVAs for performance variables during cutting with maximum speed, increased knee flexion landing, and soft landing conditions

	Cut Max Speed		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
Ankle work (J/BW/BH)	0.06 (0.02)	0.06 (0.02)	0.06 (0.01)	0.06 (0.02)	0.06 (0.02)	0.07 (0.02)	0.400	0.003	0.626
Knee work (J/BW/BH)	0.06 (0.02)	0.07 (0.02)	0.09 (0.03)	0.09 (0.02)	0.07 (0.02)	0.08 (0.02)	0.110	<0.001	0.775
Hip work (J/BW/BH)	0.09 (0.04)	0.07 (0.02)	0.15 (0.07)	0.15 (0.06)	0.10 (0.05)	0.08 (0.04)	0.429	<0.001	0.430
Total work (J/BW/BH)	0.20 (0.05)	0.19 (0.02)	0.30 (0.09)	0.30 (0.06)	0.24 (0.07)	0.23 (0.05)	0.895	<0.001	0.748
Approach speed (m/s)	4.01 (0.33)	3.68 (0.31)	3.58 (0.41)	3.18 (0.42)	3.45 (0.30)	2.98 (0.34)	0.304	<0.001	<0.001
Take-off speed (m/s)	4.36 (0.43)	4.01 (0.26)	3.82 (0.46)	3.55 (0.41)	3.85 (0.38)	3.37 (0.39)	0.110	<0.001	0.003
Stance time (ms)	290.58 (54.29)	306.26 (33.22)	432.82 (91.01)	449.00 (87.07)	372.10 (70.26)	402.53 (103.57)	0.776	<0.001	0.347

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No interaction effect of condition and gender was present for peak posterior GRF ($p = 0.511$, Table 4.30, Figure B.55, Figure B.56). A condition effect was present for peak posterior GRF ($p < 0.001$). Post-hoc testing showed that the peak posterior GRF during the soft landing condition was less than that during the cutting with maximum speed [-0.05, 0.24 BW] and increased knee flexion landing [0.13, 0.30 BW] conditions for both males and females. No gender effect was present for peak posterior GRF ($p = 0.587$, [-0.19, 0.11 BW]).

No interaction effect of condition and gender was present for timing of peak posterior GRF ($p = 0.151$). A condition effect was present for timing of peak posterior GRF ($p < 0.001$). Post-hoc testing showed that the timing of peak posterior GRF during the cutting with maximum speed condition was earlier than that during cutting with soft landing [-8.9, -1.1 ms] and increased knee flexion landing [-10.5, -3.0 ms] conditions for both males and females. No gender effect was present for timing of peak posterior GRF ($p = 0.480$, [-8.0, 3.8 ms]).

No interaction effect of condition and gender was present for vertical GRF at peak posterior GRF ($p = 0.560$, Figure B.57, Figure B.58). A condition effect was present for vertical GRF at peak posterior GRF ($p < 0.001$). Post-hoc testing showed that the vertical GRF at peak posterior GRF during the soft landing condition was less than that during the cutting with maximum speed [-0.81, -0.31 BW] and increased knee flexion landing [-0.62, -0.22 BW] conditions for both males and females. No gender effect was present for vertical GRF at peak posterior GRF ($p = 0.868$, [-0.19, 0.11 BW]).

No interaction effect of condition and gender was present for initial knee flexion angle ($p = 0.527$, Figure B.59, Figure B.60). Condition and gender effects were found for initial knee flexion angle ($p < 0.001$, $p = 0.027$). Post-hoc testing showed that the initial knee flexion during cutting with increased initial knee flexion condition was greater than that during cutting with maximum speed [2.4, 6.5 deg] and soft landing [6.8, 11.8 deg] conditions for both males and females. The initial knee flexion during cutting with maximum speed condition was greater than that during soft landing condition for both males and females [2.9 6.7 deg]. Males had greater initial knee flexion than females during all three side-cutting conditions [0.6, 8.4 deg].

No interaction effect of condition and gender was present for maximum knee flexion angle ($p = 0.183$). A condition effect was present for maximum knee flexion ($p < 0.001$). Post-hoc testing showed that the maximum knee flexion during increased knee flexion landing was greater than that during cutting with maximum speed [15.3, 20.0 deg] and soft landing [9.0, 13.4 deg] conditions for both males and females. The maximum knee flexion during soft landing condition was greater than that during cutting with maximum speed condition for both males and females. No gender effect was present for maximum knee flexion angle ($p = 0.590$, [-3.3, 5.7 deg]).

No interaction effect of condition and gender was present for range of motion of knee flexion ($p = 0.726$). A condition effect was present for range of motion of knee flexion ($p < 0.001$). Post-hoc testing showed that range of motion of knee flexion during cutting with maximum speed was less than that during increased knee flexion landing and soft landing conditions for both males and females [10.3, 16.0

deg]. No gender effect was present for range of motion of knee flexion (p = 0.273, [-9.3, 2.7 deg]).

No interaction effect of condition and gender was present for initial knee flexion velocity (p = 0.739, Figure B.61, Figure B.62). A condition effect was present for initial knee flexion velocity (p < 0.001). Initial knee flexion velocity during soft landing condition was greater than that during cutting with maximum speed [116.2, 180.4 deg/s] and increased knee flexion landing [80.6, 168.6 deg/s] conditions for both males and females. No gender effect was present for initial knee flexion velocity (p = 0.084, [-138.9, 9.2 deg/s]).

No interaction effect of condition and gender was present for knee flexion velocity at peak posterior GRF (p = 0.132). A condition effect was present for knee flexion velocity at peak posterior GRF (p = 0.002). The knee flexion velocity at peak posterior GRF during cutting with maximum speed condition was less than that during increased knee flexion landing [-104.3, -5.9 deg/s] and soft landing [-116.1, -34.5 deg/s] conditions for both males and females. No gender effect was present for knee flexion velocity at peak posterior GRF (p = 0.181, [-115.1, 22.6 deg/s]).

No interaction effect of condition and gender was present for knee flexion at peak posterior GRF (p = 0.639). A condition effect was present for knee flexion at peak posterior GRF (p < 0.001). Post-hoc testing showed that knee flexion at peak posterior GRF during increased knee flexion landing condition was greater than that during cutting with maximum speed [4.7, 8.2 deg] and cutting with soft landing [5.2, 9.5 deg] conditions for both males and females. No gender effect was present for knee flexion at peak posterior GRF (p = 0.080, [-0.4, 6.9 deg]).

No interaction or main effect was present for knee internal rotation angle or varus angle at peak posterior GRF ($p > 0.050$, Figure B.63, Figure B.64, Figure B.65, Figure B.66).

No interaction effect of condition and gender was present for knee flexion moment at peak posterior GRF ($p = 0.787$, Figure B.67, Figure B.68). Condition and gender effects were found for knee flexion moment at peak posterior GRF ($p = 0.040$, $p = 0.042$). Subjects demonstrated knee extension moment at peak posterior GRF. Post-hoc testing showed that knee extension moment at peak posterior GRF during increased knee flexion landing condition was greater than that during soft landing condition for both males and females [0.01, 0.05 BW*BH]. Males had greater knee extension moment than females during all three side-cutting tasks [0.00, 0.05 BW*BH].

No interaction effect of condition and gender was present for knee internal rotation moment at peak posterior GRF ($p = 0.128$, Figure B.69, Figure B.70). A condition effect was present for knee internal rotation moment at peak posterior GRF ($p = 0.012$). Post-hoc testing showed that knee internal rotation moment at peak posterior GRF during soft landing was less than that during cutting with maximum speed [-0.01, -0.05] and increased knee flexion landing [-0.01, -0.04] conditions for both males and females. No gender effect was present for knee internal rotation moment at peak posterior GRF ($p = 0.921$, [-0.03, 0.03]).

No interaction or main effect was present for knee varus moment at peak posterior GRF ($p > 0.050$, Figure B.71, Figure B.72).

Table 4.30. Means (standard deviations) and P-Values for ANOVAs for ACL loading factor variables during cutting with maximum speed, increased knee flexion landing, and soft landing conditions

	Cut Max Speed		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
PPGRF (BW)	-0.73 (0.30)	-0.67 (0.28)	-0.76 (0.37)	-0.78 (0.20)	-0.60 (0.25)	-0.51 (0.20)	0.511	<0.001	0.587
Time_PPGRF (ms)	25.23 (11.52)	28.10 (8.09)	30.81 (10.15)	36.06 (11.79)	32.63 (12.79)	30.71 (9.85)	0.151	0.001	0.480
VGRF_PPGRF (BW)	1.93 (0.81)	1.95 (0.88)	1.77 (0.72)	1.83 (0.49)	1.46 (0.66)	1.29 (0.40)	0.560	<0.001	0.868
Ini_KF (Deg)	25.70 (8.26)	22.30 (4.34)	31.38 (7.83)	25.58 (5.30)	21.35 (8.17)	17.03 (6.54)	0.527	<0.001	0.027
Max_KF (Deg)	49.12 (7.66)	50.25 (3.62)	68.86 (9.27)	65.77 (6.75)	56.96 (10.94)	55.32 (6.05)	0.183	<0.001	0.590
ROM_KF (Deg)	23.44 (10.12)	27.95 (6.42)	37.49 (11.75)	40.19 (8.38)	35.62 (12.09)	38.29 (9.92)	0.726	<0.001	0.273
Ini_KFV (Deg/s)	-56.68 (139.16)	-8.49 (111.50)	-45.24 (160.42)	27.47 (99.62)	78.88 (135.82)	152.54 (97.89)	0.739	<0.001	0.084
KFV_PPGRF (Deg)	251.51 (180.21)	311.91 (94.89)	296.88 (154.41)	396.73 (85.30)	357.79 (104.58)	356.25 (102.99)	0.132	0.002	0.181
KF_PPGRF (Deg)	29.27 (6.09)	26.77 (3.43)	35.97 (6.28)	32.96 (6.14)	29.21 (8.70)	25.04 (5.41)	0.639	<0.001	0.080
KIR_PPGRF (Deg)	-2.91 (3.81)	-1.17 (2.10)	-2.70 (4.26)	-0.54 (2.46)	-3.00 (3.87)	-1.24 (1.74)	0.563	0.109	0.075
KVA_PPGRF (Deg)	-2.68 (3.93)	-2.32 (5.82)	-2.75 (4.30)	-1.92 (6.09)	-3.46 (3.57)	-2.13 (5.89)	0.372	0.392	0.609
KFM_PPGRF (BW*BH)	-0.07 (0.06)	-0.05 (0.04)	-0.10 (0.08)	-0.07 (0.04)	-0.07 (0.05)	-0.04 (0.03)	0.787	0.040	0.042
KIRM_PPGRF (BW*BH)	0.06 (0.07)	0.08 (0.07)	0.07 (0.06)	0.05 (0.04)	0.04 (0.04)	0.04 (0.03)	0.128	0.012	0.921
KVAM_PPGRF (BW*BH)	0.02 (0.04)	0.00 (0.04)	0.01 (0.03)	-0.01 (0.03)	0.01 (0.03)	0.00 (0.03)	0.494	0.557	0.101

Note: PPGRF: Peak posterior GRF; TIME_PPGRF: Timing of PPGRF; VGRF_PPGRF: Vertical GRF at PPGRF; Ini_KF: Initial knee flexion angle; Max_KF: Maximum knee flexion angle; ROM_KF: Range of motion of knee flexion: KF_PPGRF: Knee flexion angle at PPGRF; KIR_PPGRF: Knee internal rotation angle at PPGRF; KVA_PPGRF: Knee varus angle at PPGRF; KFM_PPGRF: Knee flexion moment at PPGRF; KIR_PPGRF: Knee internal rotation moment at PPGRF; KVA_ PPGRF: Knee varus moment at PPGRF.

Peak ACL Force

No interaction effect of condition and gender was present for peak ACL force ($p = 0.629$, Table 4.31). A condition effect was present for peak ACL force ($p = 0.002$). Post-hoc testing showed that peak ACL force during cutting with maximum speed condition was greater than that during increased knee flexion landing [0.12, 0.57 BW] and soft landing [0.12, 0.60 BW] conditions for both males and females. No gender effect was present for peak ACL force ($p = 0.950$, [-0.26, 0.28 BW]).

No interaction effect of condition and gender was present for timing of peak ACL force ($p = 0.276$). A condition effect was present for timing of peak ACL force ($p = 0.008$). Post-hoc testing showed that the timing of peak ACL force during cutting with maximum speed condition was earlier than that during increased knee flexion landing [-37.2, -5.4 ms] and soft landing [-26.1, -9.5 ms] conditions for both males and females. No gender effect was present for timing of peak ACL force ($p = 0.380$, [-22.1, 8.6 ms]).

No interaction effect of condition and gender was present for peak ACL force caused by anterior shear force ($p = 0.702$). A condition effect was present for peak ACL force caused by anterior shear force ($p = 0.046$). Post-hoc testing showed that peak ACL force caused by anterior shear force during increased knee flexion landing condition was less than that during cutting with maximum speed [-0.46, -0.01 BW] and soft landing [-0.33, -0.04 BW] conditions for both males and females. No gender effect was present for peak ACL force caused by anterior shear force ($p = 0.421$, [-0.37, 0.16 BW]).

No interaction for main effect was present for peak ACL force caused by internal - external rotation moment ($p > 0.050$).

No interaction effect of condition and gender was present for peak ACL force caused by varus - valgus moment ($p = 0.598$). A condition effect was present for peak ACL force caused by varus - valgus moment ($p = 0.008$). Post-hoc testing showed that peak ACL force caused by varus - valgus moment during soft landing condition was less than that during cutting with maximum speed [-0.47, -0.10 BW] and increased knee flexion landing [-0.21, -0.02 BW] conditions for both males and females. No gender effect was present for peak ACL force caused by varus - valgus moment ($p = 0.258$, [-0.11, 0.41 BW]).

Table 4.31. Means (standard deviations) and P-Values for ANOVAs for ACL force variables during cutting with maximum speed, increased knee flexion landing, and soft landing conditions

	Cut Max Speed		Increased Flexion		Soft Landing		P-Value		
Variables	Male	Female	Male	Female	Male	Female	Interaction	Condition	Gender
ACL_Peak (BW)	1.31 (0.92)	1.19 (0.50)	0.88 (0.31)	0.93 (0.46)	0.87 (0.45)	0.91 (0.32)	0.629	0.002	0.950
Timing_ACL (ms)	33.25 (17.20)	44.18 (19.12)	62.56 (48.72)	57.48 (39.64)	49.37 (28.39)	63.66 (25.81)	0.276	0.008	0.380
ACL_AS (BW)	0.64 (0.63)	0.76 (0.63)	0.45 (0.36)	0.48 (0.34)	0.57 (0.56)	0.73 (0.43)	0.702	0.046	0.421
ACL_IR (BW)	-0.02 (0.41)	-0.06 (0.32)	-0.02 (0.40)	0.07 (0.15)	-0.08 (0.34)	-0.04 (0.19)	0.427	0.218	0.710
ACL_VV (BW)	0.69 (0.77)	0.49 (0.51)	0.45 (0.37)	0.38 (0.43)	0.39 (0.37)	0.22 (0.17)	0.598	0.008	0.258

Note: ACL_Peak: Peak ACL force; ACL_AS: Peak ACL force caused by anterior shear force; ACL_IR: Peak ACL force caused by internal - external rotation moment; ACL_VV: Peak ACL force caused by valgus - varus moment; Timing_ACL: Timing of Peak ACL force.

4.8. Family-Wise Type I Error Rate

For hypothesis 1, three significant pair-wise comparisons were present (Table 4.32). The family-wise Type I error rate for hypothesis 1 was 0.0012 (Equation 3.62).

For hypothesis 2, nineteen significant pair-wise comparisons were present (Table 4.32). The family-wise Type I error rate for hypothesis 2 was 0.0124.

The family-wise Type I error rate for both hypothesis 1 and 2 was 0.0136.

Table 4.32. Type I Error Rates for Significant Pair-Wise Comparisons for Hypothesis 1

Significant Pair-Wise Comparisons for Hypothesis 1	Type I Error Rates
Jumping fast had greater peak ACL force than jumping for 60% of maximum height	6.0E-06
Jumping fast had greater peak ACL force than jumping for maximum height	4.2E-07
Cutting with maximum speed had greater peak ACL force than cutting with 60% of maximum speed	1.2E-03

Table 4.32. Type I Error Rates for Significant Pair-Wise Comparisons for Hypothesis 2

Significant Pair-Wise Comparisons for Hypothesis 2	Type I Error Rate
Jumping with increased knee flexion landing had less peak ACL force than jumping for maximum jump height	2.2E-04
Jumping with increased knee flexion landing had greater total work than jumping for maximum jump height	2.4E-09
Jumping with increased knee flexion landing had less jump height than jumping for maximum jump height	7.2E-07
Jumping with increased knee flexion landing had longer stance time than jumping for maximum jump height	1.6E-10
Jumping with soft landing had less peak ACL force than jumping for maximum jump height	4.1E-3
Jumping with soft landing had greater total work than jumping for maximum jump height	1.3E-4
Jumping with soft landing had less approach speed than jumping for maximum jump height	2.4E-07
Jumping with soft landing had less jump height than jumping for maximum jump height	4.9E-10
Jumping with soft landing had longer stance time than jumping for maximum jump height	4.8E-07
Cutting with increased knee flexion landing had less peak ACL force than cutting with maximum speed	3.8E-03
Cutting with increased knee flexion landing had greater total work than cutting with maximum speed	4.2E-12
Cutting with increased knee flexion landing had less approach speed than cutting with maximum speed	1.2E-10
Cutting with increased knee flexion landing had less take-off speed than cutting with maximum speed	8.9E-12
Cutting with increased knee flexion landing had longer stance time than cutting with maximum speed	1.3E-13
Cutting with soft landing had less peak ACL force than cutting with maximum speed	4.2E-3
Cutting with soft landing had greater total work than cutting with maximum speed	4.6E-05
Cutting with soft landing had less approach speed than cutting with maximum speed	5.6E-16
Cutting with soft landing had less take-off speed than cutting with maximum speed	7.4E-13
Cutting with soft landing had longer stance phase than cutting with maximum speed	2.0E-08

4.9. Summary of Results

Subjects in this study performed all testing tasks with similar reliability. The data generally met the statistical assumptions for repeated measure ANOVA in terms of outliers, normality, and homoscedasticity.

The peak ACL force occurred within 60 ms after initial contact during all stop-jump conditions and cutting with maximum speed and cutting with soft landing conditions. The magnitudes peak ACL forces ranged from 0.64 to 1.34 body weights during different jumping and cutting conditions. Tibial anterior shear force was the major contribution to peak ACL force. Valgus - varus moments also significantly contributed to peak ACL force. Internal - external rotation moments had a small contribution to peak ACL force. Peak ACL force decreased during jumping and cutting with increased knee flexion conditions. 100% ankle or 100% hip co-contraction caused less than 10% changes in the magnitude and timing of peak ACL force.

For specific aim 1 in stop-jump, it was confirmed that jumping fast condition had the shortest stance time. Jumping for maximum height condition had the highest jump height. Jumping for 60% of maximum height had approximate 60% of the jump height during jumping for maximum height condition.

With regard to kinetic variables, jumping fast condition had the greatest posterior GRF, vertical GRF, and knee extension moment among three jumping conditions. Jumping for maximum height condition had greater posterior GRF and vertical GRF but similar knee extension moment and timing of peak posterior GRF

compared to jumping for 60% of maximum height condition. With regard to kinematic variables, jumping fast condition had the lowest maximum knee flexion angle, knee flexion range of motion, and knee flexion velocity during early landing phase among three jumping conditions. Jumping for 60% of maximum height condition had the lowest knee flexion angle during early landing phase among three jumping conditions. With regard to ACL force variables, jumping fast condition demonstrated the greatest peak ACL force among three jumping conditions. Jumping for maximum height and jumping for 60% of maximum height conditions had similar peak ACL force. In addition, males had greater knee flexion but less knee flexion velocity during early landing phase than females. Males and females had similar peak ACL force.

For specific aim 1 in side-cutting, it was confirmed that cutting with 60% of maximum speed condition had approximate 60% of approach and take-off speeds and longer stance time compared to cutting with maximum speed condition.

With regard to kinetic variables, cutting with maximum speed condition demonstrated greater posterior GRF, vertical GRF, knee extension moment, knee internal rotation moment, and knee varus moment compared to cutting with 60% of maximum speed condition. With regard to kinematic variables, cutting with maximum speed condition had greater knee flexion angle but less knee flexion velocity during early landing phase and less knee flexion range of motion compared to cutting with 60% of maximum speed condition. Cutting with maximum speed condition had greater peak ACL force than cutting with 60% of maximum speed condition. In addition, males and females had similar peak ACL force.

For specific aim 2 in stop-jump, it was confirmed that increased knee flexion landing had the greatest knee flexion angle throughout the entire landing phase among three jumping conditions. It was confirmed that soft landing had the lowest posterior GRF, vertical GRF, knee extension moments, and the greatest knee flexion velocity during early landing phase among three jumping conditions. Increased knee flexion landing and soft landing had decreased peak ACL force compared to jumping for maximum height condition. In addition, males had greater knee flexion but less knee flexion velocity during early landing phase than females. Males and females had similar peak ACL force.

With regard to changes in performance outcomes, increased knee flexion landing and soft landing increased knee work, hip work, total work, stance time and decreased jump height compared to jumping for maximum height condition. Soft landing also decrease approach speed compared to jumping for maximum height condition.

For specific aim 2 in side-cutting, it was confirmed that increased knee flexion landing had the greatest knee flexion angle throughout the entire landing phase among three cutting conditions. It was confirmed that soft landing had the lowest posterior GRF, vertical GRF, knee extension moments, and the greatest knee flexion velocity during early landing phase among three cutting conditions. Increased knee flexion landing and soft landing had decreased peak ACL force compared to cutting with maximum speed condition. In addition, males had greater knee flexion during early landing phase than females. Males and females had similar peak ACL force.

With regard to changes in performance outcomes, increased knee flexion landing and soft landing increased knee work, hip work, total work, stance time and decreased approach speed and take-off speed compared to cutting for maximum speed condition.

The family-wise Type I error rate for all significant pair-wise comparisons for hypothesis 1 and 2 was 0.0136.

CHAPTER V

DISCUSSION

5.1. Reliability

Subjects in this study performed all testing tasks with similar reliability. One of the potential limitations of this study was the reliability of the data. Testing tasks including jumping for 60% of maximum height, jumping with increased knee flexion, cutting with 60% of maximum speed, and cutting with increased knee flexion were novel to the subjects. Subjects, therefore, might not be able to consistently perform these novel tasks after 5 practice trials. The results of this study showed that most CMCs during the four novel tasks were similar to the CMCs during jumping for maximum height and cutting for maximum speed. These results indicated that subjects were able to perform each of those novel tasks with the similar reliability as they performed those tasks they frequently performed after five practice trials, and that the reliability of the performances of those novel tasks was not be a factor that affect the quality of the data in this study.

The reliability of the data obtained in this study was similar to those reported in the literature. In terms of the magnitudes of CMCs, excellent reproducibility have been observed in most sagittal plane variables, while moderate reproducibility have been observed in some non-sagittal plane variables. Milner et al. (Milner et al., 2011)

assessed the within-session reliability of knee biomechanics during a stop-jump task. The subjects in their study were instructed to jumping for maximum height. The CMCs reported in their study were 0.85 for vertical GRF, 0.96 for knee flexion-extension angle, 0.94 for knee flexion-extension moment, 0.80 for knee valgus-varus angle, and 0.67 for knee valgus-varus moment, respectively, which are similar to those observed in this study (Table 4.2). The CMCs for knee flexion-extension moment, knee valgus-varus angle, and knee valgus-varus moment in the current study were slightly less than those reported in the study by Milner et al. (2011).

The slightly greater CMCs in the study by Milner et al. (2011) were likely due to lower cut-off frequencies they used to filter their raw data. In the study by Milner et al. (2011), the sampling frequency for kinematic data was 240 Hz and the sampling frequency for GRF data were 1200 Hz. In the current study, we used 10 Hz as the cut-off frequency for motion data. 10 Hz was estimated optimum cut-off frequency based on a sampling rate of 120 Hz (Yu et al., 1999). Because the sampling frequency in the study by Milner et al. (2011) was greater than the sampling frequency in the current study, their estimated optimum cut-off frequency for kinematic data should be greater than 10 Hz (Yu et al., 1999). However, Milner et al. (2011) filtered their kinematic data with a cut-off frequency of 6 Hz. GRF data were highly accurate and not differentiated during calculations. Theoretically, GRF data did not need to be filtered. 200 Hz was used as the cut-off frequency for GRF data in the current study, because a high frequency noise was observed using Fourier analysis. Milner et al. (2011) used the same sampling frequency for GRF data as in the current study, but they filtered their GRF data with a cut-off frequency of 60 Hz.

Milner et al. (2011) mentioned that the cut-off frequencies were determined by the residual analysis of the data. Yu et al. (1999) demonstrated that the residual analysis developed by Winter et al. (1974) significantly under estimated the cut-off frequency and resulted in over smoothed data. It is likely that Milner et al. (2011) over smoothed both kinematic and GRF data, which resulted in a seemingly better reliability at the cost of the validity. The cut-off frequencies and CMCs in the current study should be more realistic than those reported by Milner et al. (2011), and consequently the reliability of the kinematic and kinetic data were more realistic than those reported by Milner et al. (2011). The CMCs of the biomechanical data of the side-cutting tasks were generally greater than the CMCs during the stop-jump task in the current study. No reliability of these data was found in the literature.

5.2. Data Screening

The data generally met the statistical assumptions for repeated measure ANOVA. A few outliers were observed. Including the outliers had minimal effect on the statistical outcomes. Therefore, the outliers were included in the statistical analysis. A small portion of the variables violated the assumption of normality. Repeated measure ANOVA was robust against moderate violation of normality (Collier et al., 1967; Howell, 2009). Therefore, the violations of normality should have minimal effects on the statistical outcomes. The violation of between - subject homoscedasticity might raise concern about the gender effects in repeated measure ANOVA. However, only a small portion of the variables violated this assumption and

ANOVA was robust against moderate violation of between - subject homoscedasticity (Box, 1954; Howell, 2009). In addition, if the assumption of within-subject homoscedasticity was violated, the Greenhouse - Geisser correction was applied to adjust the p values (Greenhouse and Geisser, 1959). In summary, repeated measure ANOVA was considered an appropriate statistical method to analyze the data in the current study.

5.3. Face Validity of ACL Loading Model

An ACL loading model was developed to estimate ACL force from lower extremity kinematics and kinetics in the current study. The face validity of the model was evaluated through four comparisons between the current study and the literature: (1) timing of peak ACL force, (2) magnitude of peak ACL force, (3) composition of ACL peak force, (4) relationship between peak ACL force and knee flexion angle.

5.3.1. Timing of Peak ACL Force

The face validity of the ACL loading model was supported by the absolute timing of peak ACL force. Krosshaug et al. (2007) estimated that the timing of ACL injuries in basketball was 17 to 50 ms after initial contact. Koga et al. (2010) estimated that the timing of ACL injuries in team handball was approximately 40 ms after initial contact. In the current study, the estimated timing of peak ACL forces

were before 40ms after initial contact for jumping fast and cutting with maximum speed conditions, and between 40 ms and 50 ms after initial contact for jumping for 100% maximum height and jumping for 60% of maximum height conditions. The estimated timing of peak ACL forces was after 50ms after initial contact for the other conditions. The delayed timing of peak ACL forces during soft landing and increased knee flexion landing conditions were likely due to change in landing techniques. The delayed timing of peak ACL forces during cutting with 60%of maximum speed was likely due to a small external loading. Overall, the estimated absolute timing of peak ACL force in the current study was consistent with the timing of ACL injuries reported in the literature.

The face validity of the ACL loading model was also supported by the timing of peak ACL force relative to peak impact GRF. Cerulli et al. (2003) used an implanted strain gauge device to measure *in vivo* ACL strain during a single leg hop landing task. The investigators found that the peak ACL strain and peak impact GRF occurred approximately at the same time. Taylor et al. (2011) integrated marker-based motion analysis with fluoroscopic and magnetic resonance imaging technique to measure in vivo ACL strain during a drop vertical jump task. The investigators found that the overall peak ACL strain occurred approximately 55 ms before the foot contact the ground, when the knee flexion angle was the lowest. The ACL strain started to decrease when the knee started to flex. An increase in ACL strain was observed after the foot contacted the ground and reached a local maximum at the peak impact GRF. The studies by Cerulli et al. (2003) and Taylor et al. (2011) suggested that peak ACL strain after landing was likely to occur at the timing of peak

impact GRF. In the current study, the timing of peak ACL force occurred slightly later than the timing of peak posterior GRF. However, the differences between the timing of peak ACL force and peak posterior GRF were only less than 13 ms for cutting with maximum speed, jumping fast, and jumping for 60% of maximum height. The differences between the timing of peak ACL force and peak posterior GRF were less than 20 ms for jumping for maximum height condition. The differences between the timing of peak ACL force and peak posterior GRF were more than 20 ms for the other conditions. The large differences during soft landing and increased knee flexion landing conditions were likely due to changes in landing techniques.

The delayed timing of peak ACL force compared to timing of peak posterior GRF was likely due to the inclusion of valgus / varus moments loading. In the current model, the ACL force caused by valgus/varus moments was modeled as a function of knee flexion angle and external knee valgus/varus moments with or without the presence of anterior shear force. The contribution of valgus/varus loading to ACL loading was largely independent of anterior shear force loading mechanism. Because the maximum valgus/varus moments usually occurred during middle stance phase, the valgus / varus moments loading delayed timing of peak ACL force to a certain degree. The current model might overestimate the valgus/varus moments loading by assuming valgus/varus loading and anterior shear force loading being additive to each other. Overall, the timing of peak ACL force relative to peak impact GRF were slightly delayed in the current study.

There were several components that contributed to the early timing of peak ACL force. First of all, the knee flexion angle was small during the early phase of the

landing. As demonstrated by Taylor et al. (2011), knee flexion angle was the most important component in determining ACL loading. Secondly, a great peak impact vertical GRF could load the ACL through a tibiofemoral contact force with a posterior tibial plateau slope (Meyer and Haut, 2005; Meyer and Haut, 2008). A great vertical impact GRF could also generate great external valgus and varus moments which could load the ACL (Berns et al., 1992; Markolf et al., 1995). Thirdly, peak impact vertical GRF and peak impact posterior GRF usually occurred at the same time. To maintain the dynamic equilibrium, a great impact posterior GRF was associated with a great knee extension moment. Quadriceps was the major muscle that generated knee extension moment and tibial anterior shear force. A great peak posterior GRF could load the ACL through a great tibial anterior shear force. In the current model, all of these loading components have been included and contributed to the prediction of the timing of peak ACL force.

5.3.2. Magnitude of Peak ACL Force

The face validity of the ACL loading model was supported by the magnitude of peak ACL force. Cerulli et al. (2003) measured the *in vivo* ACL strain for one male subject in the landing of a single leg hop task, which is similar to the landing of the stop-jump tasks in the current study. The in vivo measured peak ACL strain for this male subject was 5.47%. Taylor et al. (2011) measured in vivo ACL strain for 8 male subjects during a drop vertical jump task, and reported a peak strain of 12% before initial foot contact with the ground and a peak strain of 7% after initial foot contact

with the ground. Chandrashekar et al. (2006) reported that the average length of male ACL was 30mm and the average stiffness of male ACL was 306 N/mm. Combined the results of these studies together, the estimated peak ACL force was 505 N in the study by Cerulli et al. (2003) and 1109 N before foot contact with the ground and 647 N after foot contact in the study by Taylor et al. (2011), respectively. In the current study, the averaged estimated peak ACL force was 607± 326 N during jumping for maximum height condition, which was similar to the *in vivo* measured ACL forces in similar tasks reported in the previous studies, and thus supported the face validity of the ACL loading model.

5.3.3. The Composition of Peak ACL Force

The face validity of the ACL loading model was partially supported by the composition of peak ACL force. The results of this study showed that tibial anterior shear force applied was the major loading mechanism of peak ACL force. The current model was based on the data reported by Markolf et al. (1995). Beside the study by Markolf et al. (1995), other previous studies have also demonstrated that anterior shear force was the major loading mechanism of ACL (Berns et al., 1992; Durselen et al., 1995; Fleming et al., 2001; DeMorat et al., 2004). Durselen et al. (1995) found that an application of a 140 N quadriceps force significantly increased the ACL strain from 20 to 60 degrees of knee flexion. DeMorat et al. (2004) demonstrated that a 4500 N quadriceps muscle force caused ACL injuries at 20 degrees of knee flexion in *vitro*. Berns et al. (1992) measured the in vitro ACL strain

caused by a 200 N tibial anterior shear force applied to the tibia. Their results demonstrated that the ACL strain was primarily caused by the anterior shear force, and that neither pure valgus / varus moments nor internal / external rotation moment had significant effects on the ACL strain. Fleming et al. (2001) tested the effects of tibial anterior shear force on ACL strain in *vivo*. A 130N anterior shear force was applied to the subjects' tibia when the knee was flexed at 20 degrees. For both weight bearing and non-weight bearing conditions, ACL strain increased as the anterior shear force increased. However, valgus / varus moments and external rotation moments had small effects on ACL strain. In the current study, anterior shear force contributed 45 - 81% to the peak ACL force during all jumping and cutting conditions. Anterior shear force contributed to more than 55% of the peak ACL force during jumping fast and cutting with maximum speed conditions during which ACL injuries were more likely to occur. However, while Internal / external rotation moments had very small contribution to the ACL force, valgus / varus moments (valgus moments for most jumping and cutting conditions) contributed 23 - 49% of the peak ACL force.

The effects of valgus / varus moments and angle on ACL loading mechanism, ACL injury risk, and ACL injury mechanism are still not completely understood. Previous investigators have demonstrated that isolated valgus moment had small effect on ACL loading. Markolf et al. (1995) showed that a pure 10 Nm valgus moments only loaded the ACL to approximate 50 N at 0 - 90º knee flexion angles in *vitro*. Berns et al. (1992) observed that a 20 Nm valgus moment could not significantly load the ACL at 30º of knee flexion in *vitro*. Fleming et al. (2001) showed

that 0-20 Nm valgus moments did not significantly increase ACL strain under both weight bearing and no weight bearing conditions in *vivo*. Previous researchers have also shown that medial collateral ligament is the major structure to resist valgus moments. Mazzocca et al. (2002) demonstrated that ACL strain increased substantially only after MCL ruptured under valgus loading in *vitro.* Matsumoto et al. (2001) showed that when the medial collateral ligament was severed, a significant increased in valgus rotation angle and a large medial knee joint space were observed in *vitro*. When the ACL was severed, a significant increase in internal rotation but a small medial knee joint space was observed. Lujan et al. (2007) showed that ACL transaction had little effect on MCL strain during 10 Nm valgus loading in *vitro*. Bendjaallah et al. (1997) employed a finite element model to analyze the effects of varus – valgus moments on knee ligament forces. The investigators showed that a 10 Nm valgus moment only loaded the ACL to less than 50 N with an intact MCL. However, a 10 Nm valgus moment loaded the ACL up to 300 N without a MCL. Seering et al. (1980) evaluated the contributions of different ligaments and tissues to valgus loading in two cadaveric knees. When the valgus angle was 8 degrees, the MCL contributed to 55% and 71% of the resistance to valgus loading. However, the ACL only contributed to 0% and 3% of the resistance to valgus loading. Shin et al. (2009) employed a knee model to study the effect of isolated valgus moments on ACL strain during a simulated landing task. The researchers found that valgus moment increased ACL loading at a low valgus loading level. ACL strain became insensitive to valgus moment when the valgus moment increased to more than 50 Nm. The findings suggest that valgus moment may not be sufficient to cause

an ACL tear without completely tear the MCL. However, only 6% patients who had ACL injuries completely ruptured their medial collateral ligaments (Fayad et al., 2003). This evidence suggests that the anterior shear force is the major loading mechanism of ACL, while valgus moments should be considered a secondary loading mechanism.

On the other hand, although isolated valgus loading is not likely to cause an ACL injury, previous investigators found that valgus loading could increase ACL loading in combination with other loading. In the study by Markolf et al. (1995), an addition of 10 Nm valgus moment to a 100 N anterior shear force increased the ACL force compared to a pure 100 N anterior shear force loading when the knee flexion angle was more than 5 degrees. In the study by Bern et al. (1992), an addition of 20 Nm valgus moment to a 100 N anterior shear force increased the ACL force compared to a pure 100 N anterior shear force loading at 30 degrees of knee flexion angle. Shin et al. (2011) applied dynamic loading during a single leg landing to a knee model to predict ACL strain. The researchers showed that the peak ACL strain increased when valgus moments or internal rotation moments increased. In addition, combined knee valgus and internal rotation moments generated greater ACL strain than either alone. Withrow et al. (2006) used a simulation apparatus to study the effect of valgus alignment on ACL loading during a simulated landing task. The researchers demonstrated that combined impulsive force, muscle forces, and valgus moment increased ACL strain more than the same loading without the valgus moment. In addition, valgus moment has been identified as a prospective risk factor for ACL. Hewett et al. (2005) conducted a study in attempt to determine the risk

factors of ACL injury for female young athletes. Compared to uninjured athletes, the injured athletes demonstrated greater initial and maximum knee valgus angles, smaller maximum knee flexion angles, greater peak external hip flexion moments, and greater peak ground reaction force during landing phase as well as greater peak external knee valgus moments and less stance time during the stance phase. Regression analysis showed that the peak knee valgus moment predicted ACL injury with 73% specificity and 78% sensitivity.

The underlining mechanism of increased ACL loading during combined valgus moment and other loading condition is not clear. However, a few postulations might be proposed. The interplay of valgus moment and anterior shear force might be similar to interplay of knee flexion angle and anterior shear force. Simple knee flexion angle could not significantly load the ACL, but knee flexion angle can change the loading effect of anterior shear force on ACL force through modifying ACL elevation angle (Li et al., 2005). Similarly, a valgus moment can cause a knee valgus angle which might change the loading structure of ACL and exaggerate the loading effect of anterior shear force on ACL force. However, it is unknown whether ACL elevation angle changes according to knee valgus angle. This postulation needs further investigations. The combined loading of valgus moment and compressive might be associated with a posterior tibial plateau slope. A compressive load applied in the setting of a posterior tibial slope can increase ACL loading via relative anterior tibial translation (Meyer and Haut, 2005; Meyer and Haut, 2008). Recent studies suggest that the slope in the lateral compartment may be of greater relative importance (Hashemi et al., 2010; Simon et al., 2010). Knee valgus during landing

would accentuate the effects of the lateral compartment, particularly under circumstances of high compressive forces and low knee flexion angles (Chaudhari and Andriacchi, 2006; Boden et al., 2009). Therefore, a valgus angle might increase the anterior shear force generated by compressive force and increase ACL force.

In the current study, the valgus moment effect on ACL force was modeled using the data reported by Markolf et al. (1995). The effect of valgus moment on ACL force was modeled differently with or without the presence of an anterior shear force. With an anterior shear force, the valgus moment effect on ACL force was increased compared to without an anterior shear force to represent combined loading effect on ACL force. In addition, it was assumed that the loading effect of valgus moment on ACL loading was linear without upper limit. The current model might overestimate the valgus effect from a few perspectives.

First of all, the combined loading effect of anterior shear force and valgus moment on ACL force can be modeled in two different ways. The current study assumed that valgus moment had a larger effect on ACL force with the presence of an anterior shear force. However, it could also be assumed that anterior shear force had a larger effect on ACL force with the presence of a valgus angle which resulted from a valgus moment. Previous investigators have shown that pure valgus moment has small effect on ACL loading (Berns et al., 1992; Markolf et al., 1995). Therefore, it might be more reasonable to assume that valgus angle exaggerate the anterior shear force effect instead of anterior shear force exaggerate the valgus moment effect. However, in the study by Markolf et al. (1995), the valgus angle was not reported. The only way to model the combined loading effect of anterior shear force

and valgus moment in the current study was to isolate the valgus moment and assumed valgus moment was linear to ACL force. Therefore, the valgus moment effect in the current study could be largely overestimated because this valgus moment effect could actually be an anterior shear force effect. The valgus moment contribution to peak ACL force in the current study should not be interpreted as an isolated valgus moment effect on ACL force but the combined effect of anterior shear force and valgus moment on ACL force.

Secondly, the assumed linear relationship between ACL force and valgus moment and the unlimited upper boundary of valgus moment effect on ACL force could overestimate ACL force. Shin et al. (2009) showed that the relationship between ACL loading and valgus moment was actually nonlinear. Because valgus moment caused tibia external rotation which could unload the ACL and MCL was the major ligament to valgus moment, ACL strain became relatively insensitive to valgus moment when the valgus moment increased to more than 50 Nm. Matsumoto et al. (2001) showed that medial knee joint space largely increased only when the medial collateral ligament was severed. Mazzocca et al. (2003) demonstrated that ACL strain increased substantially only after MCL ruptured which would cause large medial knee joint space. The findings suggested that the valgus loading effect on ACL force should be limited to a low level before knee joint space increases. The valgus loading should be mostly resisted by the MCL. Without considering the nonlinear relationship between valgus moment and ACL loading, the current study could overestimate the valgus moment effects.

Thirdly, the magnitude of anterior shear force and valgus moment applied in previous studies should raise attention. In the study by Markolf et al. (1995) and many other studies, the applied anterior shears force was usually about 100 N and the applied valgus moment was usually about 10 – 20 Nm. The applied anterior shear force in previous studies was much smaller than the actual anterior shear force during jumping and cutting tasks, while the applied valgus moment was in previous studies was close to the actual valgus moment during jumping and cutting tasks. The data reported by Markolf et al. (1995) might overestimate valgus moment effect on ACL loading. If the anterior shear force in the study by Markolf et al. (1995) was greater, an addition of valgus moment to anterior shear force might not significantly increase ACL loading. The underrepresented anterior shear force might overestimate the valgus loading effect on ACL force.

Although the current model might overestimate the valgus moment effect on ACL loading, the overestimation should have relatively small effect on the interpretations on the results. If we ignore the valgus / varus moment effect on ACL loading, considering the internal / external rotation moment effect was small, the peak ACL force would be approximate the peak ACL force caused by anterior shear force component in the current analysis. The statistical test outcomes for peak ACL force and peak ACL force caused by anterior shear force in the current analysis were similar except for the comparison between cutting with maximum speed condition and cutting with 60% of maximum speed condition. Cutting with maximum speed condition had greater external loading but increased knee flexion during early landing phase compared to cutting with 60% of maximum speed condition. The

greater peak ACL force during cutting with maximum speed condition compared to cutting with 60% of maximum speed condition was mainly due to greater peak ACL force caused by valgus / varus moment. In addition, the comparisons between males and females would be different because females had greater peak ACL force caused by anterior shear force during many jumping and cutting conditions. Ignoring valgus / varus moment would present a greater peak ACL force in females compared to males. However, the conclusion of females being at greater level of ACL loading as a percentage of maximum ACL loading would be same.

In summary, the current model showed that anterior shear force had a major contribution to peak ACL force during most jumping and cutting conditions. However, the valgus / varus moment effect on ACL force could be overestimated. It should be noticed that the valgus / varus moment effect in the current study should not be interpreted as an isolated valgus / varus loading effect but a combined effect of anterior shear force and valgus / varus loading. The overestimation of valgus - varus moment effect might have influence on the interpretation of cutting speed effect on ACL loading. However, the interpretation of changes in ACL loading between other jumping and cutting conditions should be similar if we exclude valgus / varus loading mechanism or simply assess ACL loading based on kinematic and kinetic data.

5.3.4. The Relationship between Peak ACL Force and Knee Flexion Angle

The results of this study showed that the peak ACL force was sensitive to knee flexion angle during landing. Taylor et al. (2011) showed that ACL force was

sensitive to knee flexion angle during pre-landing. A recent study done by the same group showed that landing with increased initial knee flexion angle decreased peak ACL length during the pre-landing and landing phases of a drop vertical jump task (Brown et al., 2012). In the study by Brown et al. (2012), subjects received verbal instruction to increase the initial knee flexion. Subjects increased the initial knee flexion by 4 degrees during the increased knee flexion landing compared to regular landing. Similar to the findings by Taylor et al. (2012), Brown et al. (2012) found that maximum ACL length occurred during pre-landing. A four degree increase in initial knee flexion decreased the maximum ACL strain from 12% to 8% during pre-landing. Brown et al. (2012) did not report descriptive data for changes in maximum ACL strain during the landing phase. From the graphs reported by Brown et al. (2012), it was observed that the maximum ACL length during the landing phase of the increased knee flexion landing condition was slightly less than that during the regular landing condition. In the current study, subjects received verbal instruction to increase the initial knee flexion during a stop-jump task. Subjects increased initial knee flexion by 10 degrees during the increased knee flexion landing condition compared to jumping for maximum height condition. The peak ACL force during the increased knee flexion condition was 26% less than that during the jumping for maximum height condition. The percentage of decrease in peak ACL force during landing in the current study appeared to be greater than the percentage of decrease in peak ACL strain during landing in the study by Brown et al. (2012). The greater percentage of decrease in the current study could be due to a greater increase in initial knee flexion angle (10 degrees vs. 4 degrees) and inherent limitations of the

model. The decreased peak ACL during increased knee flexion landing supported the face validity of the biomechanical model for the sensitivity of peak ACL force to knee flexion angle in the current study.

The decrease in ACL peak force caused by increased knee flexion was because of changes in patella tendon - tibia shaft angle, hamstring tendon - tibia shaft angle, and ACL elevation angle which were all considered in the current model. A less knee flexion was associated with a greater patella tendon-tibial shaft angle (Nunley et al., 2003), a less hamstring tendon-tibia shaft angle (Nunley et al., 2003; Lin et al., 2009), and a greater ACL elevation angle which all contributed to a greater ACL loading (Li et al., 2005). The decrease in ACL force during increased knee flexion condition was mainly due to the decrease in ACL force caused by anterior shear force in the current study.

5.4. Sensitivity Analysis

The major assumption of the model was that there was no muscle co-contraction at the ankle and hip joints. A sensitivity analysis was conducted to assess the effect of different percentages of co-contraction at ankle and hip on the estimate of peak ACL force.

5.4.1. Ankle Co-contraction

Co-contraction at the ankle joint slightly increased the magnitude of peak ACL force and decreased the timing of peak ACL force. Co-contraction at the ankle joint increased the gastrocnemius muscle force. Because gastrocnemius muscle pulled the femur in the posterior direction and generated knee flexion moment, an increase of gastrocnemius muscle force increased the tibial anterior shear force and ACL force. However, the percentage of changes in peak ACL force caused by ankle co-contraction was small. The major ankle plantarflexors were the gastrocnemius and soleus muscles. The major ankle dorsiflexors were the tibialis anterior, extensor hallucis longus, and extensor digitorum longus muscles. The sum of peak forces of major plantarflexors (5500 N) was much greater than the sum of peak force of dorsiflexors (1184 N) (Arnold et al., 2010). The moment arms of dorsiflexors were similar to the moment arms of plantarflexors (McCullough et al., 2011). In addition, the muscle activation level of tibialis anterior was similar to the muscle activation levels of gastrocnemius and soleus muscles during jump landing task (Iida et al., 2011). Therefore, the moment generated by ankle dorsiflexors should be small compared to ankle plantarflexors, so the con-contraction level should be low during jump landing tasks. In addition, because the gastrocnemius-tibia shaft angle was only 3 degrees, the effects of gastrocnemius muscle force on tibial anterior shear force and ACL force were small. Therefore, co-contraction at the ankle joint had small effects on magnitude and timing of peak ACL force.

5.4.2. Hip Co-contraction

Co-contraction at the hip joint slightly decreased the magnitude of peak ACL force and had small effects on timing of peak ACL force. Hip co-contraction increased the hamstring muscle force. The hamstring muscle was pulling the tibia toward the posterior direction. A pure increase in hamstring muscle force would decrease the anterior shear force and ACL force. However, hamstring muscle force also generated knee flexion moments. An increased in hamstring force required an increase in patellar tendon force to generate knee extension moment to maintain the same knee joint resultant moment. A part of the patellar tendon force would apply an anterior shear force on the tibia. In addition, the increases in hamstring force and quadriceps force increased the tibiofemoral compression force. Because of a posterior tibial plateau slope, the increase in tibiofemoral compression force also generated an anterior shear force. Therefore, the eventual effects of increase in hamstring force on ACL force were determined from the sum of the posterior shear force caused by increased hamstring force, the anterior shear force caused by increased patellar tendon force, the anterior shear force caused by increased tibiofemoral compression force, and knee flexion angle.

Previous investigators have simulated hamstring co-contraction effects on ACL force. In the study by O'Connor et al. (1993), the quadriceps force was changing according to changes in hamstring force to generate an extension moment to counterbalance the flexion moment of the hamstring. The co-contraction of quadriceps and hamstring actually increased the ACL force when the knee flexion

angle was less than 22 degrees. Yu and Garrett (2005) simulated the effects of hamstring contraction on ACL force at the timing of peak posterior GRF during a stop-jump task. Hamstring forces were set at different levels and at different knee flexion angles. Quadriceps forces were adjusted to satisfy the knee joint resultant moments. ACL forces were estimated from the quadriceps force, hamstring forces, joint resultant force, and ACL elevation angle. The hamstring co-contraction did not decrease ACL force until the knee flexion angle was greater than 15 degree for males and 20 degrees for females. Previous studies suggested that the effects of hamstring co-contraction on ACL force were depended on knee flexion angle and gender.

In the current study, the simulation results showed that the hamstring co-contraction did not decrease ACL force until the knee flexion angle was greater than 25 degrees for males and 26 degrees for females. The reason that the cut-off knee flexion angles in the current study were greater than previous studies (O'Connor, 1993; Yu and Garrett, 2005) was because of the inclusion of tibiofemoral compressive force and posterior tibial plateau slope. If the compressive force loading mechanism was ignored in the current study, the cut-off knee flexion angles would be 14 degree for males and 16 degrees for females. The current study confirmed that effects of hamstring co-contraction on ACL force were depended on knee flexion angle and gender.

The simulation results explained the small effects of hip co-contraction on peak ACL force during jumping and cutting tasks. In the current study, the peak ACL force typically occurred at 30-40 degrees of knee flexion. At 30-40 degrees of knee

flexion, the posterior shear force generated by the hamstring force caused by co-contraction was only slightly greater than the anterior shear force generated by the quadriceps and the compressive force. Therefore, co-contraction at the hip joint had small effects on magnitude and timing of peak ACL force.

A combination of co-contractions at the ankle and hip joints had small effects on the magnitude and timing of peak ACL force. Because the co-contraction at the ankle increased the peak ACL force and the co-contraction at the hip decreased the peak ACL force, the combination of ankle and hip co-contraction counterbalanced each other to a certain degree and resulted in small changes in peak ACL force.

5.5. Comparisons of ACL Loading Models

Previous investigators have used musculoskeletal modeling to evaluate ACL force during dynamic movements (Pflum et al., 2004; Kernozek and Ragan, 2008; Laughlin et al., 2011). Kernozek et al. (2008) used an EMG driven model to estimate ACL force during a double-leg drop landing task. The estimated peak ACL force was only 94 N. The estimated timing of peak ACL force occurred between the initial contact and maximum vertical GRF. Laughlin et al. (2011) used optimization model to predict peak ACL force during soft and stiff single leg landing tasks. The estimate peak ACL force was 440 N during soft landing and 497 N during stiff landing. The estimated timing of peak ACL force occurred 7 -10 ms after initial contact. Pflum et al. (2004) used an EMG driven model to estimate peak ACL force during a double-

leg drop landing task. The estimated peak ACL force was 253 N. The timing of peak ACL force occurred at 40 ms after the initial contact.

The estimated peak ACL forces in previous modeling studies were generally less than the peak forces estimated from the studies by Cerulli et al. (2003) and Taylor et al. (2011). The discrepancies among studies could be due to different testing protocols, subject characteristics, and modeling methods. In the study by Kernozek et al. (2008), the hamstring muscle force and gastrocnemius muscle force were estimated from normalized EMG. The calculations of anterior shear force and ACL force were similar to the current study. Because it was a simple drop landing task, the posterior GRF which was an important ACL loading factor (Yu et al., 2006) could be small. In addition, only the sagittal plane loading mechanism was modeled. Therefore, it was reasonable to observe that the estimated peak ACL force was small compared to the other studies.

Laughlin et al. (2011) used OpenSim to calculated muscle forces. The authors employed similar methods in the study by Kernozek et al. (2008) to calculate peak ACL force. To resolve muscle redundancy problem, OpenSim used an optimization method to minimize a cost function to calculate muscle forces. The current study assumed no muscle contraction at the ankle an hip joints which was similar to a minimal cost function. Therefore, the results in the current study should be compatible to Laughlin et al.'s study. Because Laughlin et al. (2011) only modeled the ACL sagittal plane loading mechanism, it was reasonable to observe a less peak ACL force in their study compared to the current study.

Pflum et al. (2004) used a forward dynamic model to estimate muscle forces. The model included 54 musculotendinous units. EMG data were recorded for 7 muscles or obtained from previous literature. The input muscle excitations for the muscles were manually adjusted to match the measured kinematic and kinetic data. The relative position between tibia and patellar and femur and joint torques were calculated from GRF, muscle forces, and lower extremity motions. The bone poisons and joint torques were used as inputs to a knee model to calculate ACL force. This was an elaborate model with thorough consideration of muscle forces and joint ligaments. However, in order to match the simulated results with measured data, the muscle excitations were subjectively adjusted. The knee model to calculated ACL force was based on a mechanical force-strain model. Compared to the mechanical model, the current study calculated the ACL force based on an *in vitro* simulation study (Markolf et al., 1995). The model in the current study might give a better representation of the physiological structure, while the model in the study by Pflum et al. (2004) might had a better consideration of mechanical structure. The relatively low peak ACL force observed in the study by Pflum et al. (2004) might be because of a less challenging landing task and a pure mechanical model to calculate ACL force.

In the current study, muscle forces were estimated from a torque driven model. Forces acting on the knee joint were calculated from muscle forces and knee geometry. The effects of anterior shear force and non-sagittal plane moments on ACL force were based on an in vitro simulation study. The estimated peak ACL forces in the current study were close to the estimated ACL forces in previous *in vivo*

studies. The current model demonstrated good validity in timing of peak ACL force, compositions of peak ACL force, and sensitivity of peak ACL force to knee flexion angle. However, many assumptions had been made and the content validity of the model was not evaluated.

The current model assumed that there was no muscle co-contraction at the ankle and hip joints. In addition, it was assumed that force distribution between gastrocnemius and soleus and the force distribution between hamstring and gluteus maximus was depended on their peak muscle forces and moment arms without considering muscle activation levels. Although sensitivity analysis and previous literature suggested that the assumptions might not cause a fatal flaw to the model, a lack of physiological input of muscle activation level was still the major limitation of the model. Muscle activation levels could be different in different jumping and cutting conditions can cause different ACL loading. Compared to EMG driven models, the current model simplified the calculations but lost the validity of physiological input of muscle activation level.

The current model also assumed that muscle moment arms obtained from the literature were proportional to subjects' body height or segment length. The lines of action of muscle force were assumed to be the same across different subjects. The effect of muscle activation on the length of muscle moment arm was negligible. The moments generated by passive tissues including ligaments and joint capsule were negligible. The effects of friction force between femur and tibia was negligible. The ACL loadings caused by anterior shear force, internal rotation moment, and varus / valgus moment were additive. As discussed previously, because the three loading

mechanisms were modeled largely independent to each other, the effects of certain component on ACL loading might be overestimated. In addition, the ACL force estimated from the model reached a peak value during the early landing phase and then drop to 0 during most time of later landing phase. During later landing phase, the greater knee flexion angle caused a posterior tibial shear force and therefore the ACL force caused by anterior shear force was constrained to be 0. Previous in vivo studies found that the ACL strain was changing continuously during the landing phase. Therefore, the 0 ACL force during most time of later landing phase might not be realistic. All these assumptions of the model limited its application and generalization to the real world, especially when the content validity of the model was not evaluated.

In summary, previous modeling studies usually involved electromyography signal process and complicated optimization. The current model simplified the calculations and estimated ACL force directly from the time series of lower extremity kinematics and kinetics data. The model demonstrated several good face validities. However, because of a lack of physiological input of muscle activation and other assumptions, the exact ACL force estimated from the model should be interpreted with caution. The current model might be used a tool to assess the resultant effects of kinematic and kinetic variables on ACL loading and evaluate the general trend of changes in ACL loading during different conditions. The estimated ACL force can be used along with kinematic and kinetic variables that can affect ACL loading to give a better understanding of changes in ACL loading. Because of the simplification, the current model may have significant advantages in application.

5.6. Specific Aim 1: Effects of Performance Demands on ACL Loading

Specific Aim 1 was to determine the effects of changes in performance demands on ACL loading in recreational athletes while performing stop-jump and side-cutting tasks. It was hypothesized that ACL loading would increase when the athletes jumped with a higher jump height and a shorter stance time during a stop-jump task. It was also hypothesized that ACL loading would increase when the athletes cut with a faster speed and a shorter stance time during a side-cutting task.

5.6.1. Condition effects

<u>Stop-Jump</u>

The results of this study support the hypothesis that ACL loading would increase when athletes jumped with a shorter stance time during a stop-jump task. However, the results did not support the hypothesis that ACL loading would increase when athletes jumped for a higher jump height during a stop-jump task.

Jumping with a fast speed rather than jumping for a maximum height is very common in sports. For example, a volleyball player needs to jump fast to spike a ball when the ball is set relatively low for spiking. A basketball player needs to jump fast for a rebound when competing with an opponent. Previous video analysis studies have demonstrated that ACL injuries usually occurred during quick deceleration athletic tasks (Boden et al., 2000; Krosshaug et al., 2007; Koga et al., 2010). Walsh et al. (Walsh et al., 2004) investigated the effects of drop heights and contact time

on GRF, joint moments, joint power, and joint work during a drop jump task. The purpose of their study was to provide information for improving jumping performance. The subjects dropped from 4 drop heights with 5 different contact times and jumped for a maximum height. The maximum vertical GRF increased as the contact time decreased. However, no other kinetic or kinematic variables that were related to ACL loading were reported.

The effects of jumping speed on ACL loading during a stop-jump task were investigated in the current study. Stop-jump tasks were commonly used to assess ACL injury risks in previous studies (Chappell et al., 2002; Chappell et al., 2005; Yu et al., 2006; Chappell et al., 2007). Different from drop landing and drop vertical jump tasks (Devita and Skelly, 1992; Hewett et al., 2005), stop-jump tasks started with an approach run and involved sudden deceleration in the anterior-posterior direction during the landing. Because the initial approach velocity was forward and downward, a posterior braking GRF and a vertical landing GRF were needed to quickly reduce the approach momentum during the landing. The posterior GRF was considered an important ACL loading factor (Yu et al., 2006). Consistent with the study by Walsh et al. (2004), the current study found a greater impact vertical GRF when subjects landed and jumped with a shorter stance time. In addition, greater impact posterior GRF and knee extension moments were also observed. To achieve the goal to jump fast, subjects landed with a rigid pattern as indicated by small initial knee flexion velocity, maximum knee flexion, and range of motion. The landing pattern was different from jumping for maximum height condition during which subjects had a relatively large knee range of motion and dissipated the landing force

over a longer period of time. The rigid landing pattern ensured that the subject could absorb the approach momentum in a short time and reduce the total stance time. At the mean time, the impact force increased and the timing of impact force decreased as compensations for decrease in landing time.

The landing pattern was a reflection of how subjects prepared and conducted their movements from a motor control aspect. The decreased knee flexion velocity at initial contact suggested that subjects had already programmed the rigid landing pattern before the landing. The earlier timing of peak impact force suggested that the subjects not only deceased the total stance time but also shortened the relative time for critical events. After landing, subjects generated great joint extension moments to push against the ground. Great GRFs were generated to stop the body moving forward and downward. It was observed that subjects need minimal practice to successfully and consistently perform the jumping fast condition during data collection, although the official trials were collected after 5 practice trials. It was speculated that subjects had commonly performed fast jumps during real practice and competition. A motor program for fast jump has already formulated before the testing.

The increased ACL loading factors and peak ACL forces in jumping fast condition suggests an increased risk for ACL injuries. Jumping fast condition had 3 degree less knee flexion at peak posterior GRF compared to jumping for maximum height condition. Jumping fast condition also had 26% more peak posterior GRF, 22% more vertical GRF, and 28% more knee extension moments at peak posterior GRF than jumping for maximum height condition. The estimated peak ACL force during

jumping fast condition was 34% more than the peak ACL force during jumping for maximum height conditions. The increased vertical GRF was associated with increased tibiofemoral contact force. The increased posterior GRF was associated with increased knee extension moment. The decreased knee flexion, increased tibiofemoral contact force, and increased knee extension moments increased the tibial anterior shear force and ACL force through sagittal plane loading mechanism. The increased vertical GRF which was associated with increased valgus / varus moments increased ACL loading through non- sagittal plane loading mechanism.

The results of this study provide significant information for understanding non-contact ACL injury. As mentioned previously, Taylor et al. (2011) observed a maximum ACL strain during pre-landing of a drop vertical jump task. The authors suggested that a hypothetical injury scenario could be disruptions of the timing of landing events. One example could be the athletes being perturbed during the midair and the predetermined neuromuscular programming of landing being changed. In the current study, although the subjects landed with a rigid pattern during jumping fast condition, the initial knee velocity was still in the flexion direction and the knee flexion angle at peak posterior GRF was 8 degrees greater than the initial knee flexion angle. Because of this flexion pattern, the peak ACL force during jumping fast condition was well below the maximum ACL loading capacity (Chandrashekar et al., 2006). However, because the peak ACL force during jumping fast condition was closer to the maximum ACL loading capacity compared to other conditions, relatively small perturbations to the landing patterns might result in ACL injuries. As an addition to theory proposed by Taylor et al. (2011), the current study suggest that an

injury event was likely to occur when an athlete plan to landing in a short stance time but the motor programming is perturbed before the landing or during early landing.

Contradictory to our hypothesis, no difference was observed estimated peak ACL force between jumping for maximum height and jumping for 60% of maximum height conditions. Most previous investigators tested athletes during jump landing task with maximum jump height as the performance demand (Chappell et al., 2002; Hewett et al., 2005; Yu et al., 2006; Lin et al., 2008). It was assumed that jumping for a maximum jump height represented an injury scenario for ACL injuries. The current study evaluated the effects of jump height following landing on ACL loading during a stop jump task. However, the results suggested that ACL loading did not change when the athletes jumped with a higher jump height during a stop-jump task.

Subjects changed their landing patterns according to different jump heights. How high a subject would jump was largely depended on the concentric work done by the lower extremities during the take-off phase. To achieve a given jump height, subjects needed to generate certain mechanical work from the end of landing to take-off. It was speculated that after years of practice and competition, subjects had formulated an optimal control strategy to jumping for maximum height. This control strategy required subjects to reach certain body posture at the end of landing to generate the maximum work during the take-off phase. Because the jumping for 60% of maximum height required less mechanical work done during the take-off, subject adjusted their body into a more upright posture and reduced the range of motion from the end of landing to take-off. Subjects reduced the range of motion to generate concentric work and therefore decreased the jump height. As the results showed,

less knee flexion angles during the early and middle phase were observed during the jumping for 60% of maximum height condition compared to jumping for maximum height condition. Because of the less knee flexion angles, the jumping for 60% of maximum height condition had similar peak ACL force caused by anterior shear force compare to the jumping for maximum height condition although the jumping for maximum height condition had higher peak posterior GRF and vertical GRF.

The results of this study provide significant information for understanding the biomechanics of landing. Previously investigators who evaluated the effects of performance demands on lower extremity biomechanics during jump landing tasks focused on the drop height and drop distance effects (McNitt-Gray, 1993; Zhang et al., 2000). McNitt-Gray (1993) evaluated lower extremity kinetics during landing from three drop heights (0.32, 0.72, 1.28m). The peak vertical GRF increased as the drop heights increased. Subjects landed with less initial knee, and hip flexions, but increased flexion velocities and range of motions as the drop heights increased. Zhang et al. (Zhang et al., 2000) studied lower extremity biomechanics during landing from three drop heights (0.3, 0.6, 1m). The vertical GRF associated with toe-touch and heel contact increased as the drop heights increased. The joint range of motions increased as the drop height increased. In the studies by McNitt-Gray et al. (1993) and Zhang et al. (2000), subjects conducted simple landing without a consecutive jump. The increased range of motion during higher drop height conditions could be a strategy to increase the landing time and compensate with the increased landing force. Studying drop height and drop distance effects was

important for us to understanding basic landing biomechanics. However, considering landing from a drop height or distance was not commonly performed in sports, the generalization of the findings to the real world was limited. The current study evaluated stop-jump tasks which were commonly performed during sports such as basketball, volleyball, and soccer. The performance demands including jump height and jump speed were important performance demands during real competitions.

Side-cutting

The results of this study support the hypothesis that ACL loading would increase when athletes cut with a faster speed and a shorter stance time during a side-cutting task. Side-cut task is commonly performed in sports such as basketball and soccer. Athletes usually need to conduct a side-cutting task with a fast speed to achieve a great performance. During a side-cutting task, subjects approached to the cutting step with an initial forward and downward velocity. A posterior braking GRF and vertical landing GRF were needed to reduce a part of the anterior approach momentum and all the downward approach momentum. After the cutting step, subjects continued to run toward the forward and lateral direction. A part of initial forward momentum was transferred into take-off forward momentum. A medial GRF was generated during the take-off to change the running direction into 45 degree towards the lateral side.

Previous studies generally tested subjects with cutting speed as a controlled variable (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Pollard et al., 2006; Landry et al., 2007). One thing should be noticed was the method used to

quantify the approach speed. In the study by Pollard et al. (2006), the approach speed was the average speed 3 - 5 meters before cutting step. The cutting speed controlled between 5.5 and 7.5 m/s. In the study by Landry et al. (2007), the approach speed was the instantaneous speed before the cutting step. The cutting speed was controlled around 3.5 m/s. In the study by Malinzak et al. (2001), the approach and take-off speed was the instantaneous speed before and after the cutting step. The approach speed was approximate 5m/s and the take-off speed was approximate 4.5 m/s. In the current study, the maximum approach speed were 3.7 - 4m/s which were close to the approach speeds in the studies by Landry et al. (2007) and Malinzak et al. (2001)., but less than the speed in the study by Pollard et al. (2006). It was possible that subjects started with a fast approach speed but slowed down before the cutting step, so the instantaneous speed before the cutting step was less than the average speed before the cutting step. The purpose of the current study was to quantify the approach speed on cutting mechanics, so it was more reliable to use cutting speed right before the cutting step to exclude the confounding effects caused by average speed over a long distance.

Comparisons between the data reported in the current study and previous studies were limited. Previous investigators who studied cutting biomechanics generally reported peak kinematic and kinetic variables during the entire stance phase instead of variables at critical loading events (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Pollard et al., 2006; Landry et al., 2007). Benjaminse et al. (2011) reviewed 7 studies that evaluated gender effects on lower extremity biomechanics during cutting tasks. The mean initial knee flexion among

different studies was approximately 17 degrees. The mean maximum knee flexion was approximately 60 degrees. The initial knee flexion angle in previous studies was greater than the initial knee flexion angle during the cutting with 60% of maximum speed condition but less than the initial knee flexion angle during cutting with maximum speed condition in the current study. The maximum knee flexion angle in previous studies was greater than the maximum knee flexion angles during both cutting conditions. The difference among studies could be caused by different task demands such as cutting speed, cutting angle, and cutting anticipation.

Subjects landed with a reduced range of motion to reduce the stance time during cutting with maximum speed condition. Interestingly, subjects demonstrated a knee extension velocity at the initial contact during the with maximum speed condition. Subjects started with more initial knee flexion but reduced the initial flexion velocity to achieve a reduced range of motion during cutting with maximum speed condition. The extension velocity at initial contact suggested a pre-programmed movement pattern before the landing to achieve the goal to cut fast. On the other hand, subjects landed with less initial knee flexion and went through a greater range of knee flexion motion during cutting with 60% of maximum speed condition. From a pure knee flexion aspect, the cutting with 60% of maximum speed condition actually had similar ACL loading caused by anterior shear force because of the decreased knee flexion angle during the early phase compared to cutting with maximum speed condition.

The rigid landing pattern caused great impact GRF during cutting with maximum speed condition. The peak posterior GRF, vertical GRF, and knee

extension moment at peak posterior GRF during cutting with maximum speed condition were more than 2 times of those during cutting with 60% of maximum speed condition. The great GRF were caused by a rigid body structure as well as great joint moments. The great extension moment suggested that subjects actively push against the ground to generate great GRF. The increased posterior GRF and vertical GRF allowed the subjects to complete landing in a short time and reduce the total stance time. The earlier timing of impact force suggested an overall shift of critical events to the earlier phase. From a sagittal plane loading aspect, cutting with maximum speed condition had greater kinetic ACL loading factors which were associated with greater anterior shear force and compressive force.

Significant differences were observed in non-sagittal plane motions and moments between two cutting speed conditions. Different from the stop-jump task, the side-cutting task involved many rotational and medial-lateral movements. Because the tibia alignment of the cutting leg was usually not in the sagittal plane, greater impact vertical GRF caused great external valgus-varus moments. The rotation between the cutting foot and ground generated great external internal-external rotation moments. The greater non-sagittal plane movement could be a result of the overall rigid landing pattern. Because the non-sagittal plane loading was mainly absorbed by passive tissue at the knee joint, the greater non-sagittal plane keen motion was a reflection of the greater non-sagittal plane loading. Consistent with the increased knee external rotation angle, increase knee internal rotation moment was observed in cutting with maximum speed condition. Passive tissue generated internal knee internal rotation moment to resist the knee external rotation

motion. Similarly, passive tissue generated internal knee varus moment to resist the knee valgus motion during cutting with maximum speed condition. A part of the resisting moments generated by passive tissue came from the ACL (Berns et al., 1992; Markolf et al., 1995). Increased non-sagittal plane moments could increase the ALC loading.

The estimated peak ACL force during cutting with maximum speed condition was greater than the peak ACL force during cutting with 60% of maximum speed condition. The greater peak ACL force was due to an increase in peak ACL force caused by valgus-varus moments. Interestingly, the peak ACL forces caused by anterior shear force were similar between two cutting speed conditions. The cutting with 60% of maximum speed condition had less knee flexion angle while cutting with maximum speed condition had greater sagittal plane loading. The loading structure of knee flexion angle and the magnitude of sagittal plane loading counterbalanced each other and resulted in similar peak ACL force caused by anterior shear force between two cutting speed conditions. The results indicated the importance of considering both loading structure and magnitude of loading when evaluating ACL loading. Because of the great impact GRF which resulted in greater valgus-varus moments, the cutting with maximum speed condition had greater peak ACL force caused by valgus-varus moments.

The results of this study provided significant information for understanding ACL injury mechanism. ACL injuries were more likely to occur when subject cut with a fast speed. Subjects landed with a reduced range of motion pattern and had great external loading during cutting with maximum speed condition. Subjected landed

with more initial knee flexion at initial contact which significantly decreased the sagittal plane ACL loading. However, the knee extension velocity at initial contact and large external loading make subjects susceptible to ACL injuries under perturbations. If an athlete's motor program is perturbed before the landing which resulted in a straight knee in combination with initial knee extension velocity and great external loading, ACL could be at great risk for injury.

Previous investigators who studied performance demands on ACL loading during cutting tasks focused on reaction and fatigue effects (McLean et al., 2004; Tsai et al., 2009). McLean et al. (2004) studied the effects of defensive opponent on lower extremity biomechanics during a side-cutting task. The authors found that subjects had increased the medial GRF and the hip and knee flexion and abduction angle when cutting with a defensive player. The authors suggested that the presence of a defensive player could increase the knee loading and might bring the movement closer to ACL injuries. Chappell et al. (2005) studied the effects of fatigue on jump landing mechanics. The investigators found that the subjects landed with decreased sagittal plane motion and increased non-sagittal plane motion after fatigue. Tsai et al. (2009) found that fatigue increased peak internal knee adduction moments and peak knee abduction angle during a side-cutting task. Borotikar et al. (2008) studied lower extremity biomechanics during a single leg landing task. The authors showed that ACL injury risks were the greatest during unanticipated condition after fatigue. As an addition to previous studies, the current study suggested that fast speed should also be considered a hazardous factor for ACL injuries.

Summary

ACL loading increased when the movement speed increased during stop-jump and side-cutting tasks. The increased ACL loading was caused by a more rigid landing pattern which resulted in shorter stance time but increased peak external loading. Jump height actually did not have significant effect on ACL loading. Subject had similar peak external loading but decreased knee flexion when the jump height decreased. The results of this study provided important information in understanding ACL injury mechanism, screening ACL injury risks, preventing ACL injuries.

Movement speed was a sensitive performance demand to ACL loading. ACL injuries were likely to occur when subject planned to jump or cut with a short stance time, but the motor programming was perturbed during pre-landing or early landing. The perturbed programming could result in a decreased knee flexion in combination with decreased knee extension velocity and great external loading which could cause great ACL loading. Previous investigators generally tested athletes with maximum jump height as the performance demand. However, jump height was not a sensitive performance demand to ACL loading. Testing athletes with a non-sensitive performance demand may result in misleading testing results. Future descriptive studies as well as injury screening and injury intervention studies should consider testing subjects with jumping or cutting fast as the performance demand to have a better representation of ACL injury scenario. Technique training program should focus on modifying athletes' techniques during fast movement tasks. Jumping or cutting with a slow speed might decrease the risk to suffer ACL injuries. However,

jumping or cutting with a fast speed might not be avoidable for many players during real competitions.

5.6.2. Gender Effects

Stop-Jump

Males and females had similar responses to changes in jumping conditions. Different knee sagittal plane motions were observed between males and females. Females had less knee flexion at initial contact and peak posterior GRF and less maximum knee flexion angle than males during all three jumping conditions. Females had greater knee flexion velocity during the early phase compared to males during all three jumping conditions.

Previous studies have repeatedly found that females had less knee flexion angle during early phase of jump landing tasks compared males (Decker et al., 2003; Yu et al., 2005; Yu et al., 2006; Chappell et al., 2007). Decker et al. (2003) found that female recreational athletes landed with less knee flexion at initial contact compared to males during a drop landing task. Yu et al. (2005) demonstrated that female adolescent soccer players had decreased knee flexion angles at initial contact during a stop-jump task compared to males. Yu et al. (2006) showed that female recreational athletes had less knee flexion angles at initial contact and maximum knee flexion during a stop-jump task compared to males. Chappell et al. (2007) found that female recreational athletes had decreased knee flexions before the landing of a stop-jump task compared to males. Different from previous studies

which only looked at stop-jump tasks for a maximum jump height, the current study also evaluated jumping fast and jumping for 60% of maximum height conditions. Interestingly, no matter which condition it was, females always had less knee flexion angle during early and middle phase of the landing than males. It was unclear why females landed with less knee flexion compared to males. Possible explanations may lay in the gender differences in strength, neuromuscular control pattern, and anatomical characteristics. However, previous studies have shown that land patterns were not likely predicted by muscular strength and anthropometric factors (Bennett et al., 2008; Beutler et al., 2009). Future studies were needed to understand why females landed with less knee flexion angels compared to males.

Although females landed with less knee flexion, females had greater knee flexion velocity during the early phase compared to males. Yu et al. (2006) did not found significant difference in knee flexion velocity at initial contact between male and female recreational athletes during a stop-jump task. At the meantime, Yu et al. found that hip and knee flexion velocity at initial contact was negatively correlated with peak posterior GRF and peak vertical GRF. The authors suggested that active hip and knee motion could affect the impact force and ACL loading. In the current study, the increased flexion velocity resulted in 1-3 more degrees of knee range of motion from initial contact to peak posterior GRF in females compared to males. However, males still had 5-9 more degrees of knee flexion angle at peak posterior GRF than females. In the current study, the female subjects had an average of 12.8 years of sports experience. It was possible that the female subjects had adapted a pattern to compensate the decreased knee flexion with an increased knee flexion

velocity in order to decrease the impact GRF as suggested by Yu et al. (2006). Because of the increased knee flexion velocity in females, no significant difference was observed in impact GRF between males and females.

Although no significant difference was found in peak ACL force between males and females, females were still at greater risk of ACL injury compared to males. Chandrashekar et al. (2006) found that the maximum loading capacities of ACL were approximately 2.4 body weights for males and 1.8 body weights for females. In the current study, if the peak ACL forces were normalized to maximum loading capacities, females would have greater relative ACL forces compared to males. Females were considered more vulnerable to movement perturbations because their ACL forces were closer to the maximum loading capacity. The results were consistent with previous literature which demonstrated a greater injury rate per exposure in females compared to males (Agel et al., 2005; Hootman et al., 2007). In addition, males and females tent to load the ACL differently. Females had greater peak ACL force caused by anterior shear force while males had greater peak ACL force caused by valgus-varus moments.

Side-Cutting

Males and females had similar response to changes in cutting conditions in most ACL loading variables. Males tent to have a greater initial knee flexion angle than females during cutting with maximum speed condition while females tent to have a greater initial knee flexion angle than males during cutting with 60% of maximum speed condition.

Males increased the maximum knee flexion during cutting with maximum speed condition compared to cutting with 60% of maximum speed condition. Females had similar maximum knee flexion for two cutting speed conditions. Males increased the initial knee flexion more than females did when the cutting speed increased from 60% to 100% of maximum speed. Males utilized a deeper knee flexion motion to absorb the landing force and generate take-off force when the cutting speed increased. The deeper knee flexion was protective to ACL because of previously mentioned ACL loading mechanism. Males were more effective in utilizing different movement patterns according to changes in cutting speed compared to females.

Previous investigators (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Sigward and Powers, 2006b) usually controlled the cutting speed to a certain range when studying gender effects. The cutting speed was controlled as an absolute value instead of a percentage of maximum cutting speed. However, males usually had a fast cutting speed than females. The same absolute cutting speed might represent different percentages of maximum speed for each individual. Comparing males and females with the same absolute speed might have limited application to the real word.

Investigators of previous studies have studied gender effects on lower extremity kinematic and kinetic during side-cutting tasks (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Sigward and Powers, 2006b). Malinzak et al. (2001) compared the lower extremity biomechanics and muscle activities between males and females during a side-cutting task. The authors found that

females had less knee flexion angle and more knee valgus angle during the stance phase of a side-cutting task compared to males. McLean et al. (2004) found that females had decreased maximum knee flexions and maximum knee internal rotations but increased maximum knee valgus during a site-cutting task with defensive players when compared to males. Mclean et al. (2005) demonstrated that female basketball players had greater knee valgus and less knee flexions during early and middle phase of side-step task when compared to males. Sigward and Powers (2006b) found that female collegiate soccer players had greater external knee adduction moments and small external knee flexion moments during the early phase of a side-cutting task compared to males. However, no difference was observed in knee kinematics. Benjaminse et al. (2011) reviewed 7 studies that assessed gender effects on lower extremity biomechanics during cutting task. The investigators summarized that inconsistent results were found in gender effects on lower extremity biomechanics. Females tend to have less peak knee flexion, more peak valgus angle, and more peak external valgus moments compared to males. However, the effect sizes were small in many studies and the clinical relevance was questionable.

Similar to stop-jump, although no significant difference was observed in peak ACL force between males and females during side-cutting, females were still at greater risk of ACL injury because females' ACL has less ultimate strength (Chandrashekar et al., 2006).

Summary

Knee sagittal plane motion at early landing phase was the most prominent difference between males and females, which provide significant information for understanding injury mechanism and developing injury screening and prevention program. Males had greater knee flexion angle but less knee flexion velocity during early landing phase of stop-jump tasks compared to females. Males had a greater increase in initial knee flexion when the cutting speed increased from 60% to 100% compared to females. Females had peak ACL forces closer to the ultimate strength compared to males.

Females had greater ACL injury rates per sports exposure than males in most sporting events (Agel et al., 2005; Hootman et al., 2007). Considering the gender differences in knee sagittal plane motion and the importance of knee sagittal plane motion in loading ACL, knee sagittal plane motion should be considered a key factor in understanding gender disparity in ACL injury rates as well as an important intervention target in ACL injury prevention. Females are considered more vulnerable to movement perturbations because females' ACL has less ultimate strength. Males and females load the ACL differently and might have different injury mechanisms. Females need to achieve better movement patterns than males in order to reach the same injury risk level as males. Females need to have more knee flexion to protect the ACL because of the greater patellar tendon - tibia shaft angle (Nunley et al., 2003), greater ACL elevation angle (Li et al., 2005), and less ACL ultimate strength (Chandrashekar et al., 2006). Injury prevention training might need to focus more on reduce sagittal plane ACL loading for females.

5.7. Specific Aim 2: Movement Pattern Effects on Performance

Specific Aim 2 was to determine the effects of changes in movement patterns that should decrease ACL loading on the performance outcomes. It was hypothesized that soft landing and landing with increased knee flexion at initial contact would decrease ACL loading, but also decrease jump height and increase stance time and mechanical work during a stop-jump task. It was also hypothesized that soft landing and landing with increased knee flexion at initial contact would decrease ACL loading, but also decrease cutting speed and increase stance time and mechanical work during a side-cutting task. For Specific Aim 2, the main purpose was to compare performance outcomes among different jumping and cutting conditions. The secondary purpose was to compare ACL loading between males and females.

5.7.1. Condition Effects

<u>Stop-Jump</u>

The results of this study supported the hypothesis that soft landing and landing with increased knee flexion at initial contact would decrease ACL loading for both males and females. The decrease in ACL loading during soft landing condition was mainly due to decrease in magnitudes of internal and external loading. The decrease in ACL loading during increased knee flexion landing condition was mainly due to change in loading structure.

The results of this study also supported the hypothesis that soft landing and landing with increased knee flexion at initial contact would decrease jump height and increase stance time and mechanical work compared to regular landing during a stop-jump task. Soft landing also decreased the approach speed compared to jumping for maximum height condition.

The soft landing condition had the lowest peak posterior GRF, vertical GRF at peak posterior GRF, knee extension moments at peak posterior GRF and the greatest initial knee flexion velocity, and knee flexion velocity at peak posterior. The estimated peak ACL force was less in the soft landing condition compared to jumping for maximum height condition. Subjects were simply instructed to land softly and jump as great as possible without receiving any specific movement pattern instruction. Interestingly, subjects were able to employ a self selected movement pattern to decrease the impact GRF. The greatest initial knee flexion velocity among three jumping conditions suggested that subjects pre-programmed the active knee flexion motion before the landing. The forward and downward momentum of the body was slowly absorbed with a greater range of motion during the soft landing. The increase in range of motion and landing time dissipated the landing force and significant decrease the peak impact GRF and knee extension moments at the impact GRF. It was observed that subjects needed minimal practice to successfully and consistently perform the soft landing condition. It was speculated that subjects had already formulated soft landing motor program before the testing.

Many previous investigators have studied instructions of soft landing on biomechanical injury risk factors during jump landing tasks (Cronin et al., 2008;

Onate et al., 2001). Cronin et al. (2008) et al. found that technique instructions included landing on the forefoot, knee over toes, flexing knees before the landing, and deep knee flexion decreased peak vertical GRF during the landing after a volleyball spike. Onate et al. (2001) found that landing with forefoot, normal varus / valgus, and deep flexion decreased peak vertical GRF during the landing after a maximum vertical jump. McNair et al. (2000) showed that landing techniques with landing on forefoot and increasing knee flexion before landing decreased peak vertical GRF during a drop landing task. Prapavessis et al. (1999; 2003) found that instructions of landing with toe and deep knee flexion decreased peak GRF during a drop landing task in children and high school students. Cowling et al. (2003) showed that increasing knee flexions decreased peak vertical and posterior GRF during a single leg landing task. Podraza and White (2010) found that increased initial knee flexions were associated with decreased the peak vertical and posterior GRF during a single leg landing. Previous studies showed that explicit instructions of soft landing techniques such as landing with forefoot and deep knee flexion could decrease impact GRF. In the current study, the results suggested that athletes with a relatively long sports experience were able to land softly without specific instructions. Subject's movement patterns during soft landing were similar to some of the instructions given by previous investigators such as active knee flexion and deep knee flexion. It was necessary to give athletes explicit instructions to make them land softly. However, the question became why athletes still landed relatively hard during jumping for maximum height condition even though they knew how to land softly. The later discussion of performance outcomes addressed this question.

Landing with increased knee flexion at initial contact also decreased ACL loading. The reason to use the instruction of increased initial knee flexion was because of the importance of knee flexion in determining ACL loading (Taylor et al., 2011). In addition, previous *in vivo* study showed that increased initial knee flexion decreased peak ACL strain (Brown et al., 2012). In the current study, the increased knee flexion condition had the greatest initial knee flexion angle, knee flexion angle at peak posterior GRF, and maximum knee flexion angle among three jumping conditions. Increased knee flexion landing condition had the lowest peak ACL force among three jumping conditions.

Although subjects were only instructed to increase their initial knee flexion, increased knee flexion angles throughout the entire landing phase were observed. From a mechanical aspect, because subjects landed with increased initial knee flexion, the moment arm of body center of mass relative to the knee joint increased at the initial contact. Because of the increased moment arm, the knee joint needed to absorb a greater angular momentum. At the mean time, subjects slowly generated joint moments to reduce the angular momentum. Therefore, the knee joint went through a greater range of motion to absorb the angular momentum. From another point of view, because of the more flexed body posture, the subjects were less likely to use passive tissue to absorb the initial momentum. Joints needed to generate more active work to absorb the initial momentum. As discussed previously, ACL loading was largely depended on knee flexion angle. It was not surprised to observe the lowest peak ACL force during the increased knee flexion landing condition among three jumping conditions. During data collection, subjects were instructed to

land with increased knee flexion at initial contact. A specific knee flexion angle subjects had to achieve was not required. The purpose was to have subjects increase the initial knee flexion with their own patterns to keep the integrity of the movement. Subject usually took 3-5 practice trials to achieve a consistent movement pattern. Although subjects were able to achieve consistent movement pattern, it was speculated that subjects did not commonly perform similar task during real practice or competition. A motor program for landing with increased initial knee flexion was not likely formulated before the testing.

Landing with increased knee flexion have been commonly included in technique instructions to decrease impact GRF as discussed previously (Prapavessis and McNair, 1999; Onate et al., 2001; Prapavessis et al., 2003; Cronin et al., 2008). Technique training including increased knee flexion landing has also been included in many long term ACL injury prevention programs. Myer et al. (2005) found that a 6 week neuromuscular training program improved lower extremity biomechanics in female athletes. During the plyometric and dynamic movement training, the athletes were instructed to land softly and athletically with deep knee flexion. After the training, athletes increased their knee range of motion during a drop vertical jump task. Myklebust et al. (2003) studied the effects of a neuromuscular training program on injury rate in female handball players over three seasons. The intervention included floor, balance mat, and wobble board exercises. The athletes were instructed to land with increase hip and knee flexion during the landing exercise. A significant reduction in non-contact ACL injury in the second season was reported in comparison to the control season. Mandelbaum et al. (2005)

studied the effects of a warm-up program on prevention of ACL injury for female soccer players in two seasons. In the program, there was an emphasis on soft landing with deep hip and knee flexion. An incidence rate of 0.05 (injuries / 1000 exposures) was found in the intervention group compared to 0.47 in the control group in the first year. An incidence rate of 0.13 exposures was found in the intervention group compared to 0.51 in the control group in the second year. The results of this study supported the theory that increasing knee flexion could decrease ACL loading and should be considered during technique training program. On the other hand, knowing the changes in performance outcomes caused by deep knee flexion landing could give us a comprehensive understanding of increased knee flexion training.

The increased joint work during soft landing and increase knee flexion conditions were associated with increased joint range of motion. The increased joint range of motion decreased the peak impact GRF and peak joint moments, but prolonged the landing time. Because joint work is the time integration of joint power, the increased landing time overcame the decreased in average joint power and resulted in increase in joint work. From a motor control aspect, the subject used more muscle work to absorb the impact over a longer period of time and therefore increased the total work. The results indicated that subjects mainly increased the knee and hip work to achieve the goal to land softly. The change in ankle joint work was not significant. We did not instruct the subjects to change their foot striking pattern during the soft landing. Subjects utilized similar ankle movement pattern during different landing conditions, so no change was observed in ankle work.

The increased joint work during soft landing and increased knee flexion conditions was consistent with previous studies. Zhang et al. (2000) studied the effects of landing techniques (soft, normal, and stiff) on lower extremity biomechanics when landing from different drop heights. The soft landing decreased peak GRF and increased knee and hip joint range of motions. The soft landings also increased the eccentric work at knee and hip. Devita and Skelly (1992) studied the effects of landing stiffness on lower extremity kinetics. The soft landing had a maximum of 117 degree of knee flexion and stiff landing had a maximum of 77 knee flexion. The soft landing decreased the peak vertical GRF. However, the soft landing increased the knee and hip work as well as the total lower extremity work during the impact phase.

The increased joint work might be considered a decrease in performance in some situations. Subjects increased 30% more total work in the increase knee flexion condition and increased 10% more total work in the soft landing condition compared to jumping for maximum height condition. Because of the associations between energy expenditure and mechanical work (McCaulley et al., 2007), subjects spent more energy to achieve the same jump height if they land softly or land with increased knee flexion. During real competitions, subjects usually need to compete at a great intensity for a period of time. If athletes can not have enough time to recover, a great energy cost during each movement means that athletes need to reduce the total number of movements or playing time. In addition, subjects might reach fatigue earlier if the energy cost was greater during each movement. However, increased joint work should not always be interpreted as a decrease in performance.

In sports such as skinning and basketball, athletes usually need to lower their center of mass to maintain balance and react quickly to external changes during ready and defensive positions. Compared to a standing posture, a squatting posture might increase the mechanical work but also increase the performance of athletes during competitions.

Previous investigators have found that fatigue could increase in the biomechanical injury risk factors during jump landing tasks (Chappell et al., 2005; Borotikar et al., 2008). Chappell et al. (2005) studied the effects of fatigue on knee kinetics and kinematics during forward, vertical, and backward stop-jump tasks. Subjects increased the tibial anterior joint resultant force and valgus moments and decreased knee flexion angle during all stop-jump tasks when fatigued. Borotikar et al. (2008) studied the effects of fatigue on lower extremity kinematics during anticipated and unanticipated single leg landing tasks. Fatigue increased initial hip extension and internal rotation and peak knee abduction and internal rotation. These two studies suggested that individuals were likely to change their movement into a more rigid pattern after fatigued. This rigid movement pattern might be a strategy to decrease joint work and reduce energy cost, but the ACL loading was likely to be greater in rigid landing. Therefore, if subjects use the soft landing and increased knee flexion landing during the earlier phase of competition to decrease ACL loading, subjects might reach fatigue earlier and actually changed their movement pattern into a more risky pattern after fatigued during the later phase of competition.

Decrease approach speed during soft landing should be considered a decrease in performance in many real competitions. Subjects were instructed to

approach as fast as possible during all stop-jump tasks. The decreased approach speed during soft landing condition might an apart of the strategy subjects used to land softly. A decrease approach speed meant less approach momentum subjected need to absorb during the landing phase. Therefore, a less posterior GRF impulse was need to absorb the forward approach momentum. As discussed previously, subjects utilized active flexion motion following the landing to land softly. Additionally, subjects also approached slower to achieve the goal to land softly. This decrease in approach speed might not be desirable in real competition. For example, a soccer player might need to approach to a certain position as fast as possible to head a soccer ball before an opponent reaches the location. A slow approach speed might largely decrease the chance that the soccer player can head the ball.

Increased stance time during soft landing and increased knee flexion landing should be considered a decrease in performance in many real competitions. The soft landing and increase knee flexion landing conditions increased the stance time by more than 40% and 20% respectively compared to jumping for maximum height condition. The increased stance was desirable in reducing impact GRF because the impact force was dissipated over a longer period of time. However, the increased stance time might not be desirable in terms of performance if the jump speed is important during real competition. For example, a basketball player might lose a rebound if he/she misses the timing of the rebound even he /she can jump high.

A few previous studies reported changes in stance time when evaluating jump landing techniques (Mizner et al., 2008; Walsh et al., 2007). Devita and Skelly et al. (1992) found that the time to second peak vertical GRF was longer in soft landing

compared to stiff landing. Mizner et al. (2008) found that soft landing decreased peak vertical GRF, peak external knee abduction moments, and increased peak knee flexion angles during a vertical drop jump task. Soft landing also increased landing time. Walsh et al. (2007) found that the instructions of soft landing had no effects on peak vertical GRF and contact time during a drop vertical jump task for males. Soft landing decreased the peak vertical GRF and increased the stance time for females. However, only 6 subjects were included in each group in the study of Walsh et al. (2007). An increase in sample size might demonstrate significant differences for stance time in males as well. Myers and Hawkins (2010) investigated the effects of alterations to techniques on ACL loading and performance during a stop-jump task. The verbal instructions included increasing the amplitude of the jump prior to landing, increasing the amount of knee flexions at landing, and striking the ground with the toes first. The changes in technique increased knee flexion angles and decreased anterior tibial shear force. However, contradictory to the current study, the subjects maintained their approach speeds and contact time after the modifications in techniques. One drawback of Myers and Hawkins's study was that the order of two jump landing styles was not randomized. The changes in landing biomechanics and performance outcomes could be due to learning effects.

Soft landing and increased knee flexion landing also decreased jump height compared to jumping for maximum height condition. However, the decrease in jump height was relatively small compared to the increases in stance time and total mechanical work. The mean jump height during increased knee flexion landing condition was 92% of the jump height during the jumping for maximum height

condition. The mean jump height during soft landing condition was 93% of the jump height during the jumping for maximum height. In the study of Walsh et al. (2007), the authors did not observe change in flight time in soft landing compared to stiff landing. In the study of Myers and Hawkins (2010) , the jump height of subject actually increased. In the study of Dowling et al. (2012), subjects conducted 3 baseline drop vertical jump testing with maximum jump height. The subjects then received technique training with feedback to increase knee and hip flexions during 15-20 drop vertical jumps. The training effects were evaluated using 3 more drop vertical jumps. The authors found that subjects were able to maintain the maximum jump height during the training and training evaluation testing compared to baseline testing. The current study showed that the acute effect of landing softly or landing with increase knee flexion had significant but relatively small effects on maximum jump height. The jump height should be largely depended on the elastic energy stored during the eccentric phase and concentric work done by during the take-off phase (Anderson and Pandy, 1993; Bobbert et al., 1996). During the jumping for maximum height condition, it was speculated subjects have formatted an optimal pattern to combine the stored elastic energy and the concentric muscle contraction to reach maximum jump height. Considering subjects had greater maximum knee flexion angle during the soft landing and increased knee flexion landing conditions, the knee joint actually went through a greater range of motion from the end of landing to take-off. Subjects should be able to generate a similar amount of concentric work. However, the changes in movement pattern might cause a decrease in elastic energy storage and utilization and thus decrease in jump height.

Side-Cutting

The results of this study supported the hypothesis that soft landing and landing with increased knee flexion at initial contact would decrease ACL loading, but also decrease cutting speed and increase stance time and mechanical work during a side-cutting task for both males and females.

The decreased in ACL loading during the soft landing condition was mainly due to decreases in magnitudes of external and internal loading. The soft landing condition during the side-cutting had the lowest peak posterior GRF, vertical GRF at peak posterior GRF, knee internal rotation moment at peak posterior GRF and less the initial knee flexion velocity, and knee flexion velocity at peak posterior GRF among the three cutting conditions. The peak ACL force was also less during soft landing condition compared to cutting with maximum speed condition. Subjects conducted the soft landing with their own pattern without explicit instruction. Subjects decreased the initial knee flexion but increased the knee flexion velocity during the early phase of landing to slowly absorb the impact GRF over a long period of time. The active knee flexion motion, large knee range of motion, and long landing time significantly decrease the peak impact GRF. Subjects were likely to pre-program this soft landing pattern before the landing. Similar to stop-jump tasks, subjects needed minimal practice to successfully and consistently perform the soft landing condition in side-cutting.

The decreased in ACL loading during the increased initial knee flexion condition was mainly because of increased knee flexion during the landing. The increased knee flexion condition had the greatest initial knee flexion angle, knee

flexion angle at peak posterior GRF, and maximum knee flexion angle. The increased knee flexion condition had similar peak posterior GRF and vertical GRF at peak posterior GRF compared to cutting with maximum speed condition. However, because of the greater knee flexion during early landing, the increase knee flexion condition had less peak ACL force compare to cutting with maximum speed condition. The less peak ACL force was mainly due to decreased peak ACL force caused by anterior shear force.

Greater knee flexion angles were observed throughout the landing phase, although subjects were only instructed to increase their initial knee flexion angle at landing. As discussed previously in stop-jump task, because the subjects landed with increased initial knee flexion, the moment arm of body center of mass relative to the knee joint increased. The knee joint needed to go through a greater range of motion to counteract the increased angular momentum. However, different from stop-jump task, the increase in range of motion during side-cutting did not result in decreased posterior GRF and vertical GRF. Different from stop-jump during which the initial forward impulse of the body needed to complete absorbed during the stance phase, a greater percent of forward impulse can be retained during the side-cutting because of continuous forward movement after the cutting. A greater knee flexion angle during the landing phase might result in unnecessarily absorption of the forward impulse. It was observed that subjects usually demonstrated a braking motion which was associated with a great posterior GRF during the landing of increased knee flexion condition. Then subjects pushed against the ground to generate additional forward and lateral force to cut. In addition, subjects usually

lowered the body more during the increase knee flexion conditions. A lower body position was associated greater downward velocity of the body and could increase the vertical GRF. Therefore, the increased braking motion and lower body position could be the cause of similar impact GRF between cutting with maximum speed and increased knee flexion conditions.

A few previous studies have evaluated the effects of technique modifications on lower extremity biomechanics during cutting tasks (Dempsey et al., 2007; Dempsey et al., 2009; Cortes et al., 2012). Dempsey et al. (2007) studies the effects of side cutting techniques on lower extremity biomechanics. The different techniques included torso lean, knee flexion, and foot placement. The cutting speed was controlled among different cutting conditions. The authors found that placing the cutting foot wide from pelvis and leaning/rotating trunk toward the opposite side of cutting direction increased the knee valgus and internal rotation moments compared to regular cutting. The flexed knee condition had greater initial knee flexion than the regular cutting. The authors suggested that wide foot placements, torso leaning / rotating in the opposite cutting direction may place an athlete at greater risk of ACL injury. Based on the results, Dempsey et al. (2009) conducted a 6-week technique training study. Subjects were trained to bring the cutting foot closer to the midline of body and ensure the cutting foot was not turned in or out, and maintain an upright torso during the side-cutting. Subjects significantly decreased peak valgus moments during both planned and unplanned side-cutting after training. No differences were observed in initial and maximum knee flexion angles. Cortes et al. (2012) compared the lower extremity biomechanics between forefoot and rearfoot landing techniques

during a side-cutting task. Rearfoot landing decreased knee flexion, knee internal adduction moments, and increase knee valgus and hip flexion angel at initial contact. The rearfoot landing also increased peak knee and hip flexion angle. Previous studies suggested that trunk position, foot placements, and foot landing patterns could affect ACL loading during cutting tasks. In the current study, the subjects were instructed to land with increased initial knee flexion, because of the important of knee flexion angle in determining ACL loading (Taylor et al., 2011). The other techniques were not included in order to keep a focus on the most important technique. Subjects were able to cut softly without any explicit instruction. Soft landing and increase knee flexion landing could reduce peak ACL loading during side-cutting tasks.

Soft landing and increased knee flexion landing conditions increase total mechanical work. Increased knee flexion landing increased the total work by 55% while the soft landing increased the total work by 17% compared to cutting with maximum speed condition. The increased total work during soft landing and increased knee flexion landing conditions was mainly due to increases in knee and hip work. The increased work was consistent with previous jump landing studies (Devita and Skelly, 1992; Zhang et al., 2000). As discussed previously, the increased total work in soft landing and increased knee flexion conditions were mainly due to increased joint range of motion and prolonged the landing time. In addition, the total work during increased knee flexion landing conditions was much greater than the soft landing condition. As mentioned previously, the increased knee flexion landing imposed unnecessary braking motion. This braking motion meant

that the subjects produced more eccentric work to absorb the initial approach momentum during landing. Subjects also produced more concentric work to generate take-off momentum during take-off. On the other hand, a greater percent of approach momentum was transferred into the take-off momentum during soft landing condition. Therefore, a greater increase in mechanical work was observed in increased knee flexion condition compared to soft landing condition.

The increased total work during soft landing and increased knee flexion landing should be considered a decrease in performance during many real competitions. Athletes such as soccer players need to conduct many cutting tasks during a game. If the total energy storage is relatively constant, an increase in energy cost for each movement means that the player needs to reduce the total number of movements or reducing the playing time. Athletes might also reach fatigue earlier if the energy cost is great and there is no enough time to recover. As discussed previously, fatigue could cause more rigid landing pattern (Chappell et al., 2005; Borotikar et al., 2008) and might increase the risk of ACL injuries.

Decreased approach and take-off speeds in soft landing and increase knee flexion conditions should be considered a decrease in performance. The approach speed was the lowest during soft landing condition. As discussed previously, the decrease in approach speed decreased the initial forward momentum and could be a part of the strategy to land softly. The increased knee flexion condition usually involved a braking motion during the landing. Subjects might felt this braking motion and slowed down during the approach to decrease the amount of braking impulse. Although subjects were only instructed to land softly during the cutting, subjects had

a soft pattern during the whole stance phase. Subjects did not push against the ground as hard as cutting with maximum speed condition to generate a fast speed. In addition, a greater part of initial approach speed was lost during the landing because of a longer stance time. Therefore, the take-off speed was less during the soft landing condition. Similar to the soft landing condition, the less take-off speed during the increased knee flexion condition was likely due to a large lose of initial approach speed during the landing. On the other hand, during the cutting with maximum speed condition, subjects maintained a fast approach speed and mainly used the cutting step to change the running direction. Subjects started to extend the knee even before the landing. This active and strong extending motion largely decreased the landing time and enabled the subjects to maintain a fast speed. Previous studies usually studied gender effects and intervention effects when the speed was controlled to a certain range (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005). However, during real competitions, subjects usually need to conduct a cutting motion as fast as possible to maximize the performance. Evaluating the intervention effects on body biomechanics and cutting speed when the subjects were intruded to cut as fast as possible might be more applicable to the real world.

Increased stance time in soft landing and increase knee flexion conditions should be considered a decrease in performance. Soft landing and increased knee flexion landing increased the stance time by about 30% and 50% respectively compared to cutting with maximum speed condition. Previous investigators who studied gender or intervention effects on cutting biomechanics did not report

differences in stance time (Malinzak et al., 2001; McLean et al., 2004; Dempsey et al., 2007). In the current study, the increased stance was desirable in reducing ACL loading. However, the increased stance time meant the subjects took a longer time to complete the cutting step. During real competitions, the cutting speed should be considered a combination of approach speed, stance time, and take-off speed which give an overall estimate of how fast the athletes complete the cutting task. Soft landing and increased knee flexion landing decreased cutting speed in all three aspects.

Summary

Soft landing and landing with increased knee flexion decreased ACL loading during stop-jump and side-cutting tasks. However, soft landing and landing with increased knee flexion also decreased jump height, movement speed, and increased total mechanical work, which meant decreased performance. The results had important implications for ACL injury prevention.

Experienced athletes could achieve soft landing pattern without any explicit landing instructions. The reason why they did not conducted soft landing as their preferred landing pattern was probably because of the decrease in performance. Different from a lab setting, performance is very important during real competitions for athletes to achieve their sports goals. The results suggested that simply instructing athletes to land soft or land with increased knee flexion might have limited application in the real competitions. When the performance is the priority, soft landing and increased knee flexion landing may be sacrificed. The results of this

study revealed the limitations of certain training methods that only focused on the biomechanical risk factors of ACL injuries without fully considering changes in performance

The results of this study did not mean that soft landing or increased knee flexion landing were not beneficial or should not be included in injury prevention program. After a period of training and physical adaption, athletes may develop new landing patterns that would decrease ACL loading without compromising performance. For example, if subjects' muscle strength and power generation were enhanced at deep knee flexion position after training, subjects might land with more knee flexion but still being able to maintain the same stance time as compared to pre-training. In addition, if the subjects' endurance is enhanced after training, a greater cost of energy for each movement will not necessarily cause an early fatigue. However, all of those changes are more likely to occur after long term training instead of immediate technique training. Techniques training that decrease ACL loading without compromising performance should be explored in future studies.

5.7.2. Gender Effects

<u>Stop-Jump</u>

Males and females had similar responses to landing instructions. However, different knee sagittal plane motions were observed between males and females.

Females had less knee flexion during early landing phase than males during all three jumping tasks. Previous studies have repeatedly found that females had

less knee flexion angle during early phase of jump landing tasks compared males (Malinzak et al., 2001; Yu et al., 2005; Yu et al., 2006; Chappell et al., 2007). Previous investigators who assessed landing technique effects on landing biomechanics usually only included females or males (Devita and Skelly, 1992; Zhang et al., 2000; Mizner et al., 2008; Myers and Hawkins, 2010). The potential gender bias to landing instructions received less attention. Walsh et al. (2007) studied the effects of instructions of soft landing on peak GRF, contact time, and flight time during a drop vertical jump task in basketball players. The instructions of soft landing had not effects on peak GRF, contact time and flight time for males. However, instructions of soft landing decreased the peak GRF and increased the stance time but had no effects on flight time for females. The authors suggested that females responded differently to instructions compared to males. The results of Walsh et al. (2007) tend to be inconsistent with the results in the current study. In the current study, no condition by gender interaction effect was observed among the three jumping conditions. Males and females had similar responses to landing instructions. As mentioned previous, the study by Walsh et al. (2007) study was a between-subject design study. The number of subjects in the technique instruction group was only 6. The statically power of the study by Walsh et al. (2007) was low and an increase in sample size might demonstrate more significant differences. The discrepancies between studies could also be caused by different subject characteristic and testing tasks.

Females also had greater knee flexion velocity during the early phase compared to males during all three jumping conditions. As discussed previously,

female might have adapted a pattern to compensate the decreased knee flexion with an increased knee flexion velocity in order to decrease the impact GRF. No significant difference was observed in peak ACL force between males and females. However, females had greater relative ACL force compared to males because of weaker ACL. Because of the less knee flexion angles, females had increased peak ACL force caused by anterior shear force compared to males.

Side-Cutting

Males and females had similar responses to landing instructions during side-cutting. However, different knee sagittal plane motions at initial contact were observed between males and females.

Females had less knee flexion at initial contact during all three cutting tasks. Previous studies have shown that females had less knee flexion angle during the early phase of side-cutting task (Malinzak et al., 2001; McLean et al., 2005). The current study found that the gender effects of initial knee flexion still existed when the subjects were instructed to land softly and land with increased initial knee flexion angle. No gender bias to landing instructions was observed. Because of the greater initial knee flexion velocity in females, the knee flexion angles at peak posterior GRF became similar between males and females during these three cutting conditions. No significant difference was observed in peak ACL force between males and females.

Previous studies have also shown that females had more peak valgus angle and more peak external valgus moments during side-cutting compared to males

(Benjaminse et al., 2011). In the current study, no gender effects were found in valgus moments and valgus angle at peak posterior GRF. The discrepancies between studies could be caused by the timing of varus-valgus moment as well as the control of the cutting tasks.

Summary

Males and females had similar responses to landing instructions during stop-jump and side-cutting tasks. Gender differences in sagittal plane motion were observed during the landing conditions with and without technique instructions. Males had greater knee flexion during early landing phase compared to females.

The results of this study have implications for developing ACL injury prevention programs. Males and females were both capable to changing their movement patterns and they were likely to have similar response to the same technique training. However, this did not mean that training programs should not be gender specific. Females had less knee flexions which contributed to a greater ACL loading caused by anterior shear force. In addition, females had peak ACL forces closer to the ultimate strength compare to males. Therefore, gender differences with an emphasis in sagittal plane motion should still be considered an important factor in developing ACL injury prevention programs.

5.8. Relationships between performance and ACL loading

A strong tradeoff relationship was found between jump speed and ACL loading during stop-jump. Results for Specific Aim 1 suggested that ACL loading was largely affected by changes in jump speed during stop-jump. The jumping fast condition had the greatest ACL loading. On the other hand, the results for Specific Aim 2 suggested that jump speed was strongly affected by changes in ACL loading during stop-jump. The increased knee flexion landing had the lest ACL loading but the longest stance time. The combined results for Specific Aims 1 and 2 suggested a strong tradeoff relationship between jump speed and ACL loading.

A weak relationship was found between jump height and ACL loading during stop-jump. Results for Specific Aim 1 suggested that the ACL loading was not affected by changes in jump height in stop-jump. The peak ACL forces were similar between jumping for maximum height and jumping for 60% of maximum height. On the other hand, the results for Specific Aim 2 suggested that jump height was slightly affected by changes in ACL loading in stop-jump. The combined results for Specific Aims 1 and 2 suggested a weak relationship between jump height and ACL loading.

A strong tradeoff relationship was found between cutting speed and ACL loading during side-cutting. Results for Specific Aim 1 suggested that the ACL loading was largely affected by changes in cutting speed during side-cutting. The cutting with maximum speed condition had greater ACL loading than cutting with 60% maximum speed condition. On the other hand, the results for Specific Aim 2 suggested that cutting speed was strongly affected by changes in ACL loading

during side-cutting. The soft landing and increased knee flexion landing had decreased ACL loading but also decreased cutting speed. The combined results for Specific Aims 1 and 2 suggested a strong tradeoff relationship between cutting speed and ACL loading.

The results of this study suggested a strong tradeoff relationship between movement speed and ACL loading during stop-jump and side-cutting tasks. Increasing movement speed is likely to increase ACL loading. Decreased ACL loading could be purely cause by decrease in movement speed. In addition, decreasing ACL loading was likely to increase mechanical work. Movement speed, jump height, and mechanical work should all be considered important performance during real competition. A good ACL injury prevention program should reduce ACL injury risk factors without compromising performance. However, a lack of considering both performance and ACL loading were observed in many previous studies.

Previous studies have investigated the effects of technique instructions on impact GRF during simple drop landing (Prapavessis and McNair, 1999; McNair et al., 2000; Onate et al., 2001; Cowling et al., 2003; Prapavessis et al., 2003; Cronin et al., 2008). Technique instructions such as forefoot landing, knee over toe, normal valgus/varus, and deep knee flexion were effective in reducing impact GRF. Based on the results, the investigators suggested that injury risks might be reduce by modifying landing techniques. However, no performance variable such as landing time or mechanical work was reported in most studies. If the modified techniques resulted in increase in lading time and mechanical work, athletes were not likely to

perform the modified techniques during real competition if performance was the priority.

Limited performance variables were reported or controlled in some previous technique instruction studies (Onate et al., 2005; Mizner et al., 2008; Myers and Hawkins, 2010). However, the report of changes in performance was usually incomplete. Onate et al. (2005) found self or combination video feedback increased knee range of motions and decrease peak vertical GRF. The jump heights were used as a covariance during the analysis. However, the authors did not report changes in contact and mechanical work. Mizner et al. (2008) found that soft landing with landing on toe and deep knee flexion decreased peak vertical GRF, peak knee abduction angles, peak external knee abduction moments, and increased peak knee flexion angles. The authors reported no changes in jump heights and increased landing time. However, the changes in mechanical work were not reported. Myers and Hawkins (2010) investigated the effects of alterations to techniques on ACL loading and performance during a stop-jump task. The changes in technique increased knee flexion angles and decreased anterior tibial shear force. Subjects increased their jump heights and maintained their approach speeds and contact time after the modifications in techniques. The authors concluded that the changes in techniques decreased ACL loading and increased performance. However, the changes in mechanical work were not reported.

A lack of report of changes in performance was also observed in long-term intervention studies. Myer et al. (2005) showed that a 6-week neuromuscular training program increased knee range of motions and decreased varus and valgus

moments during a drop vertical jump tasks in female athletes. Increases in jump heights were observed during a maximum vertical jump task. However, the changes in stance time, mechanical work, and jump height during the drop vertical jump between pre-training and post-training were not reported. The incomplete report of changes in performance raised concern about the true training effects. It was possible that the training modified subjects' movement pattern into low injury risk without compromising performance. It was also possible that subjects simply landed slowly during the post-training testing to reduce injury risk. Myer et al. (2006a; 2006b) also compared the effects of plyometric training with balance training on athletes' lower extremity biomechanics and performance. Both training methods decreased biomechanical ACL injury risk factors during drop vertical jump task. Increases in jump heights were observed during a maximum vertical jump. It should be noticed that the lower extremity biomechanics and performance were tested during different tasks. The increase in jump height during maximum vertical jump did not necessarily mean that there was an increase in height during the drop vertical jump. In addition, the changes in stance time, mechanical work, and jump height during the drop vertical jump between pre-training and post-training were not reported. Chappell and Limpisvasti (2008) found that a neuromuscular training program decreased dynamic knee valgus during a stop-jump task. The training also increased the initial knee flexions and maximum knee flexions during a drop jump tasks. The authors reported that the stance time was not significantly different during two testing tasks between pre-testing and post-testing. However, the changes in mechanical work and jump height during two testing tasks were not reported.

In summary, it was important to evaluate performance and ACL loading for a given athletic task, because of the tradeoff relationships between performance and ACL loading. Evaluating performance and ACL loading individually might favor one aspect without considering the other during each test. We need to consider performance and ACL loading as a single unit during movement evaluation and injury prevention. In addition, a comprehensive report of changes in performance is needed to truly reflect the training effects. Otherwise, it will be unknown whether it is the training effect or simply the tradeoff relationships that affect ACL loading. An ideal ACL injury prevention program should decrease ACL injury risk factors without compromising performance.

5.9. Implications to Injury Risk Screening and Injury Prevention

The purpose of injury screening is to identify high risk population and potentially develop population specific injury prevention program. Screening individuals under the circumstances when injuries are likely to occur might give a better representation of injury risks as compared to circumstances when injuries are not likely to occur. The current study showed that ACL injuries were likely to occur when individuals jumping and cutting with a fast speed. Future descriptive studies as well as injury prevention studies might consider evaluated ACL injury risks during movements with a fast speed. Significant intervention effects during low demanding tasks might not necessarily transfer to high demanding tasks. Previous investigators generally tested athletes with maximum jump height as the performance demand.

However, jump height is not a sensitive factor in determining ACL loading. It might be necessary to screen and test individuals at both fundamental and high performance level, because abnormities at fundament level might be exaggerated at the high performance level. In addition, conducting motion analysis might not be feasible at all settings. Screening individuals during simple and low cost fundamental tasks that have a good correlation of high level fast movement tasks should still be encouraged. Achieving a good movement pattern during fast movement tasks should be the goal at final stage in preventing ACL injuries. Because decrease in ACL loading might be simply caused by decrease in movement speed, movement assessments need to fully consider the changes in performance. A success technique training program should be able to modify athletes' techniques during fast movement tasks without compromise performance.

The current study showed that soft landing and increased knee flexion landing both decreased ACL loading. However, soft landing and increased knee flexion landing also decreased movement speed and increased mechanical work. It was concluded that simply instructing individuals to land softly or land with increased knee flexion might have limited generalization to the real world. However, the findings also provide some implications in long-term training. If long-term training can induce positive adaptation, individuals might be able to land softly and land with increased knee flexion without compromising performance.

The current study demonstrated that soft landing has similar ACL loading but faster movement speed and less mechanical work compared to increase knee flexion landing. Soft landing tent to be more effective in preserving performance

compared to increased knee flexion landing. One difference between soft landing and increased knee flexion landing from a motor control aspect is the point of focus. Soft landing is considered external focus instruction because subjects can utilize their own movement pattern to decrease the external impact force. Increased knee flexion angle landing is considered internal focus because subjects have to change their movements specifically according to instruction. Previous investigators showed that external focus training were more effective in acquisition of motor skills compared to internal focus (Wulf, 2001; Benjaminse and Otten, 2011). Internal focus might interfere with individuals' own movement pattern and cause a breakdown in the natural movement pattern (Benjaminse and Otten, 2011). As discussed before, subjects could conduct the soft landing condition with minimal practice which suggests relatively small change in natural movement pattern. However, the increased knee flexion angle landing usually need practice trials and had the longest stance times which suggest a breakdown in the natural movement pattern. Therefore, soft landing instruction might be more beneficial than increased knee flexion landing instruction from a short-term training point. Future studies that use external focus but are able to increase knee flexion angle during early landing phase are needed.

To achieve the goal to decrease ACL loading using soft landing and increased knee flexion landing during real competitions, maintaining or improving performance is important. In the current study, the decreased ACL loading during soft landing condition was caused by decreased external loading. The decreased external loading was achieved by increased joint range of motion during landing

which also caused the increase in stance time and mechanical work. One way to maintain stance time while landing softly is to increase individual's muscle power during the take-off phase. The increased muscle power can decrease the time of take-off phase and compensate with increased time during landing phase. In addition, if the individual's endurance is enhanced after training, a greater mechanical work for each movement will not necessarily cause an early fatigue.

On the other hand, the decreased ACL loading during increased knee flexion landing condition was caused changes in loading structure. The increase in stance time during increased knee flexion landing was likely due to increased angular momentum than needed to be absorbed by muscles and a weaker muscle power generation capacity at greater knee flexion posture. In addition, the prolonged landing time compromised the muscle stretch-shortening cycle (Komi and Gollhofer, 1997) and decreased the storage and utilization of elastic energy. Therefore, to maintain or improve performance while using increased knee flexion technique, it is important to increase muscle power generation capacity at great knee flexion posture. If the muscle power at great knee flexion posture increased, the knee can absorb the angular momentum for the knee joint in a short time. The knee can also push again the ground harder during take-off and decrease the time for take-off phase. In addition, it might also be important to change optimal muscle length into a longer length and change the muscle stretch-shortening cycle into a great knee flexion posture. In addition, an improved endurance is needed to compensate for the increased mechanical work.

The gender effect observe in the current study could also provide information in preventing ACL injuries. For recreational athletes, knee sagittal plane motion was the most prominent difference between males and females. However, it should be noticed that the gender effect might not be present for other population (Sigward and Powers, 2006b). The gender effect might also be caused by other factors such as level of competitions (Sigward and Powers, 2006a). However, for recreational athletes, males had greater knee flexion angle during early landing phase of stop-jump tasks compared to females. Females had peak ACL forces closer to the ultimate strength compared to males. Considering female recreational athletes had smaller knee flexion angle and females had greater patellar tendon - tibia shaft angle, greater ACL elevation angle, and less ACL ultimate strength, females need to achieve better movement patterns than males. Injury prevention training might need to focus more on reduce sagittal plane ACL loading for females.

5.10. Limitations

There were several major limitations of the current study:

1. Although the model has demonstrated good face validity with previous in vivo studies, the content validity of the model was still unknown. The model might overestimate the valgus - varus loading effect on ACL force. The model has included all the important ACL loading mechanisms. However, many assumptions have been made to estimate ACL force. All the assumptions

limited its application and generalization to the real world. The absolute ACL force estimated from the model needed to be interpreted with causation. However, the ACL loading was evaluated from a combination of ACL loading factors and the ACL force estimated from the model in the current study. Similar conclusion would be made if the valgus - varus loading mechanism was not included or only kinematic and kinetic variables were analyzed. The ACL force estimated from the model was used to confirm the resultant effect of kinematic and kinetic variables on ACL loading. Therefore, the validity of the current study should only be slightly affected if the ACL loading model was not fully valid.

2. The athletic tasks in the current study were limited to a single stop-jump or a single side-cutting task without repetitive motion. There are other athletic tasks such pivoting task or sports specific tasks that were not studied.

3. The performance demands in the current study were limited to jump height, jump speed, and cutting speed. There are other performance demands such as jumping and cutting directions that were not included but might affect ACL loading. Therefore, the performance demand in the current studies does not represent all the possible performance demands during real competitions.

4. The performance demands in the current study were limited to jump height, jump speed, and cutting speed. There are other performance demands such

as jumping and cutting directions that were not included but might affect ACL loading. Therefore, the performance demand in the current studies does not represent all the possible performance demands during real competitions.

5. The current study only studied soft landing and landing with increased initial knee flexion techniques. There are many other commonly used techniques such as forefoot landing might reduce ACL loading. The techniques included in the current studies were the major components but not all the components that were commonly used in technique training program.

6. Only adult athletes without major lower extremity injuries were included in the study. It was unknown the generalizations of the findings to adolescent and patient populations. The findings of the current study should be limited to adult and healthy population.

7. The effects of technique instruction on changes of performance were only immediate effect and should not be generalized to long term training effect. It was unknown the effects of long term soft landing or deep knee flexion landing training on both ACL loading and performance.

8. The tradeoff relationships between performance and ACL loading were based on that individuals did not change their physical capabilities. After long term training and physical adaptation, the tradeoff relationships might not hold.

5.11. Significance and Recommended Future Studies

A musculoskeletal model was developed to estimate ACL force from the time series data of lower extremity kinematics and kinetics. The model demonstrated good face validity in timing of peak ACL force, magnitude of peak ACL force, composition of peak ACL force, and sensitivity of peak ACL force to knee flexion angle. The major assumption of the model was no co-contraction at the ankle and hip joint. However, this assumption had small influence on the estimate of magnitude and timing of peak ACL force. Most previous studies evaluated ACL injury risks using lower extremity kinematics and kinetics. However, the changes in ACL loading become confounding when some variables suggested a greater ACL loading and some other variables suggested a less ACL loading. By applying the current model, investigators can evaluated the resultant effects of lower extremity kinematic and kinetic on ACL loading. Different form previous musculoskeletal models which usually involve electromyography signal process and complicated optimization, the current model only needs the time series data of lower extremity kinematics and kinetics. The simplicity of the model might widen its application for both research and applied settings. The model can be applied to other studies that investigate ACL injury risk in different tasks, different populations, different sports, and different interventions. In addition, the different contributions to ACL force can be assessed to provide insight into the ACL loading mechanism. The specific ACL loading pattern could provide important information in developing specific ACL injury prevention programs.

The Specific Aim 1 investigated the effects of performance demands on ACL loading during stop-jump and side-cutting tasks. ACL loading increased when jump speed and cutting speed increased. ACL loading was not affected by jump height. The results of this study provided important information in understanding ACL injury mechanism, screening ACL injury risks, preventing ACL injuries. ACL injuries were more likely to occur when the athletes move at a fast speed. However, in previous jump landing studies, investigators generally tested athletes with maximum jump height as the performance demand. Previous studies might actually tested subjects during a scenario that was not likely to cause ACL injuries. The current study suggested that future jump landing studies should consider testing subjects with jumping fast as the performance demands to have better representation of ACL injury scenario. Considering movement speed is the sensitive factors to ACL loading, technique training program should focus more on modifying athletes' techniques during fast movement tasks. For players whose priority was injury prevention but not performance, adapting a slow movement pattern might decrease their risk to suffer ACL injuries. For players whose priority is performance, fast movements might not be avoidable. A good ACL injury prevention training program should reduce their ACL loading without compromising their performance.

The Specific Aim 2 was to determine the effects of changes in the movement patterns that should decrease ACL loading on the performance outcomes. The results of this study provided important information in ACL injury prevention. Soft landing and increased knee flexion landing decrease ACL loading, but also caused slower movement speed, lower jump height, and great mechanical work. Simply

instructing athletes to land soft or land with increased knee flexion with an aim to decrease ACL force might be effective during lab setting but have limited application in the real world. For players whose priority was injury prevention, land softly and land with increased knee flexion might decrease the ACL injury risk. For players whose priority is to achieve great performance, they are not likely to utilized soft landing or increase knee flexion landing patterns because of the decrease in performance. The results of this study reveal the limitations of certain training methods that only focused on the biomechanical risk factors of ACL injury without fully considering changes in performance. A long term training program is likely needed to achieve the goal to modifying athletes' preferred movement pattern into lower risks as well as maintaining or improving performance.

The tradeoff relationship between performance and ACL loading during stop-jump and side-cutting tasks suggest the important of considering performance and ACL loading as a combined unite during movement evaluations and injury prevention training. The current study provides new information in developing evaluation tests and criteria for reporting intervention effects on movement patterns. When the intervention effects on ACL loading were evaluated, changes in ACL loading as well as changes in performance including movement speed, stance time, mechanical work, jump height, and other related performance factors need to be completely reported.

Gender effects were investigates as a secondary purpose in the current study. The major gender effects observed in the current study was sagittal plane motion. Males had greater knee flexion angle during the landing phase, while female

had greater knee flexion velocity during the early phase of landing. Sagittal plane should be considered an important factor in understanding gender disparity in ACL injury rates. The contributions of peak ACL force were different between males and females. Females had greater peak ACL force caused by anterior shear force, while males had greater peak ACL force caused by valgus/varus moments. We need to consider the gender effects when screening for great risk population and developing injury prevention program. Males and females had similar peak ACL force after normalized to body weight. However, because females have less ACL maximum loading capacity as a percentage of body weight, females are at greater risk for ACL injuries than males. Females need to achieve movement patterns that cause less ACL loading than males to reach the same risk level for ACL injuries as males. Sagittal plane motion should be the focus for preventing ACL injuries in females. Males and females are both capable to changing their movement patterns and they were likely to have similar response to the same instructions. The gender differences as well as the similar response to instruction should be considered during ACL injury prevention training program.

The current study provides some basic but important information that is needed to be conscribed in studying ACL injuries. Along with the results of this study, further studies are needed to provide a comprehensive understanding of ACL injuries and eventually decrease the overall rate of ACL injuries. Future studies included but not limited to:

1. Evaluating the content validity of the musculoskeletal model. The model cannot be considered as fully validated unless it has a good content validity. Future studies might use strain gauge or imaging techniques to measure in vivo ACL strain to validate the model.

2. Studying the effects of different combination of performance demands on ACL loading during athletic task. The current study only evaluated jump height and jump speed demands for stop-jump and cutting speed demand for side-cutting. Other task demands such as anticipated versus unanticipated task, jumping / cutting directions, and fatigue can be added to the current performance demands. Other tasks such as cross cutting, pivoting, or repetitive tasks can be studies to have a better understanding of ACL injury risks imposed by performance demands during different tasks.

3. Studying the effects of other technique modification on ACL loading as well as performance. The current study only studied soft landing and landing with increased initial knee flexion techniques. Further studies can investigate other commonly used techniques such as forefoot versus rearfoot landing, foot placement, active hip flexion, and trunk posture.

4. Studying the relationships between performance and ACL loading in other populations. Only adult athletes without major lower extremity injuries were included in the study. It is important to study adolescent athletes and ACL

injured athletes because they are at great risk for ACL injuries and ACL re-injuries.

5. Conducting video analysis studies to evaluate the effects of technique instructions on athlete's movement patterns in real world. The current study found that some technique instruction decreased performance. It was speculated that athletes are not likely to use these techniques during real competition when the performance is the priority. Video analysis studies during real competition are needed to test this postulation.

6. Comprehensively evaluating the training effects on both ACL loading and performance for current ACL injury prevention programs. A lack of report of changes in performance was a major drawback of previous intervention studies. Changes in performance should be considered being associated with the generalization of the program to the real world. Before a training program is used for large scale application, we need to have a complete understanding of the training effects on both ACL loading and performance.

7. Evaluating a deep knee flexion training program on ACL loading and performance. The finding of the current study suggested that deep knee flexion could largely decrease peak ACL loading. However, decreases were observed in jump speed and mechanical work. After long term training of deep knee flexion landing, subjects' strength and endurance could increase at

deep knee flexion position. The conditioning effects might compensate the deep knee flexion posture and make the subjects to be able to maintain the jump speed. The increased endurance might compensate with the increased mechanical work during each jump and avoid early fatigue.

8. Conducting a prospective study to identify ACL injury risk factors using jumping fast as the performance demands during jump landing task. A few prospective studies have identifies biomechanical risk factors for ACL injuries during jump landing tasks. However, all the performance demands were jump as high as possible. Considering ACL loading was greater during jumping fast condition, testing subjects under a situation that is more likely to have ACL injuries might give us better prospective prediction of ACL injuries.

5.12. Conclusions

The current study has the following conclusions:

1. ACL loading increased when the jump speed increased during a stop-jump task. ACL loading remain similar when the jump height increased during a stop-jump task. ACL injuries were more likely to occur when the athletes jump with a fast speed.

2. ACL loading increased when the cutting speed increased during a side-cutting task. ACL injuries were more likely to occur when the athletes cut with a fast speed.

3. For players whose priority was injury prevention but not performance, adapting a slow movement pattern could decrease ACL loading. For players whose priority is performance, fast movements might not be avoidable. Technique training might need to focus on modifying techniques during fast movements.

4. Soft landing and increased knee flexion landing decreased ACL loading during a stop-jump task. Soft landing and increased knee flexion landing decreased movement speed and jump height and increased mechanical work which indicated decreased performance during a stop-jump task.

5. Soft landing and increased knee flexion landing decreased ACL loading during a side-cutting task. Soft landing and increased knee flexion landing decreased movement speed and increased mechanical work which indicated decreased performance during a side-cutting task.

6. For players whose priority was injury prevention, land softly and land with increased knee flexion could decrease ACL loading. For players whose priority was performance, simply instructing them to land soft or land with

increased knee flexion might have limited application in the real world because of the decrease in performance.

7. There is a tradeoff relationship between movement speed and ACL loading during stop-jump and side-cutting tasks without long term training. We need to consider performance and ACL loading as a combined unite during movement evaluations and injury prevention training. We need to completely report the changes in performance in order to have a thorough understanding of training effects.

8. Long term training is likely needed to modify athletes' preferred movement pattern into lower injury risks without compromising performance.

9. The major gender difference during stop-jump and side-cutting was sagittal plane motion. Males had greater knee flexion angle during the landing phase. Female had greater knee flexion velocity during the early phase of landing.

10. Males and females loaded the ACL differently. ACL loading in females were closer to the maximum ACL loading capacity compared to males.

11. Males and females were both capable to changing their movement patterns and likely to have similar response to the same instructions.

12. The gender differences as well as the similar response to instruction should be considered during injury risk screening and injury prevention.

APPENDIX A

Results for Pilot Studies

Table A.1. A comparison of soft landing and landing with increased initial knee flexion during the stop-jump task in a pilot study (n=5)

Variables	Soft Landing	Increased Flexion
Initial Knee Flexion Angle (Degs)	27.4 (10.1)	36.7 (5.2)
Maximum Knee Flexion Angle (Degs)	93.1 (3.0)	120.6 (14.7)
Knee Flexion Range of Motion (Degs)	65.6 (12.7)	83.9 (10.3)
Peak Posterior GRF (N)	393.0 (106.5)	343.0 (77.7)
Knee Flexion at Peak Posterior GRF (Degs)	58.2 (5.1)	86.4 (26.4)
Vertical GRF at Peak Posterior GRF (N)	769.6 (176.8)	771.8 (197.0)
Knee Extension Moments at Posterior GRF (Nm)	123.8 (35.7)	156.4 (75.6)

Note: Values are means (standard deviations).

Table A.2. A comparison of soft landing and landing with increased initial knee flexion during the side-cutting task in a pilot study (n=5)

Variables	Soft Landing	Increased Flexion
Initial knee flexion (Degs)	27.2 (9.3)	31.2 (6.5)
Maximum knee flexion (Degs)	64.3 (2.9)	76.5 (6.5)
Knee flexion range of motion (Degs)	37.1 (8.6)	45.4 (8.2)
Maximum posterior GRF (N)	538.1 (172.5)	630.1 (297.1)
Knee flexion at maximum posterior GRF (Degs)	58.2 (5.1)	86.4 (26.4)
Vertical GRF at maximum posterior GRF (N)	1316.3 (270.9)	1411.9 (613.7)
Knee extension moments at posterior GRF (Nm)	120.4 (71.0)	146.2 (52.3)

Note: Values are means (standard deviations).

Table A.3. CMC for kinematic and kinetic variables during the stop-jump task in a pilot study (n=6)

Variables	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion
Posterior GRF	0.93 (0.02)	0.80 (0.17)	0.88 (0.12)	0.91 (0.04)
Vertical GRF	0.92 (0.03)	0.87 (0.07)	0.92 (0.05)	0.93 (0.03)
Ankle Dorsiflexion-Plantarflexion Angle	0.87 (0.15)	0.91 (0.03)	0.92 (0.04)	0.95 (0.03)
Knee Flexion-Extension Angle	0.94 (0.06)	0.94 (0.04)	0.97 (0.02)	0.98 (0.01)
Hip Flexion-Extension Angle	0.95 (0.06)	0.94 (0.04)	0.93 (0.04)	0.97 (0.02)
Ankle Dorsiflexion-Plantarflexion Moment	0.86 (0.16)	0.94 (0.03)	0.91 (0.04)	0.87 (0.12)
Knee Flexion-Extension Moment	0.93 (0.02)	0.91 (0.03)	0.95 (0.03)	0.95 (0.02)
Hip Flexion-Extension Moment	0.70 (0.12)	0.78 (0.09)	0.78 (0.14)	0.91 (0.05)

Note: Values are with means (standard deviations).

Table A.4. Averaged standard deviation for kinematic and kinetic variables during the stop-jump task in a pilot study (n=6)

Variables	Jump Fast	Jump Max Height	Jump 60% Max Height	Increased Flexion
Posterior GRF	52.5 (9.3)	47.0 (16.2)	41.2 (20.0)	30.1 (11.3)
Vertical GRF	105.1 (25.4)	80.9 (21.1)	77.5 (28.9)	48.6 (19.2)
Ankle Dorsiflexion-Plantarflexion Angle	3.4 (1.7)	4.3 (0.7)	3.1 (0.6)	3.2 (1.1)
Knee Flexion-Extension Angle	4.4 (2.4)	6.4 (2.5)	3.7 (1.5)	3.8 (1.4)
Hip Flexion-Extension Angle	4.2 (2.8)	6.2 (2.8)	5.0 (1.9)	5.4 (1.8)
Ankle Dorsiflexion-Plantarflexion Moment	16.1 (9.9)	9.9 (3.4)	11.9 (5.3)	9.6 (3.0)
Knee Flexion-Extension Moment	25.4 (11.1)	15.1 (6.4)	15.4 (7.9)	10.6 (3.6)
Hip Flexion-Extension Moment	39.8 (15.1)	24.5 (10.9)	24.2 (12.9)	15.3 (5.1)

Note: Values are means (standard deviations).

Table A.5. CMC for kinematic and kinetic variables during the side-cutting task in a pilot study (n=6)

Variables	Cut Fast	Cut Max Speed	Cut 60% Max Speed
Posterior GRF	0.91 (0.03)	0.93 (0.04)	0.95 (0.04)
Vertical GRF	0.94 (0.03)	0.95 (0.04)	0.93 (0.06)
Ankle Dorsiflexion-Plantarflexion Angle	0.85 (0.14)	0.91 (0.05)	0.90 (0.06)
Knee Flexion-Extension Angle	0.92 (0.03)	0.95 (0.03)	0.97 (0.03)
Hip Flexion-Extension Angle	0.95 (0.03)	0.95 (0.04)	0.94 (0.04)
Ankle Dorsiflexion-Plantarflexion Moment	0.78 (0.28)	0.93 (0.07)	0.86 (0.09)
Knee Flexion-Extension Moment	0.86 (0.06)	0.93 (0.05)	0.91 (0.06)
Hip Flexion-Extension Moment	0.78 (0.09)	0.78 (0.12)	0.78 (0.16)

Note: Values are means (standard deviations).

Table A.6. Averaged standard deviation for kinematic and kinetic variables during the side-cutting task in a pilot study (n=6)

Variables	Cut Fast	Cut Max Speed	Cut 60% Max Speed
Posterior GRF	51.3 (8.8)	31.5 (11.4)	38.8 (19.0)
Vertical GRF	97.4 (17.9)	81.7 (32.6)	67.5 (22.4)
Ankle Dorsiflexion-Plantarflexion Angle	4.7 (1.5)	3.9 (2.0)	5.4 (2.8)
Knee Flexion-Extension Angle	4.4 (1.2)	3.8 (2.6)	3.3 (0.9)
Hip Flexion-Extension Angle	5.1 (1.4)	4.4 (3.2)	6.6 (3.0)
Ankle Dorsiflexion-Plantarflexion Moment	17.9 (8.1)	11.0 (5.4)	13.9 (6.6)
Knee Flexion-Extension Moment	30.7 (11.9)	16.1 (7.5)	21.5 (8.3)
Hip Flexion-Extension Moment	42.8 (14.4)	26.8 (13.3)	32.6 (12.8)

Note: Values are means (standard deviations).

APPENDIX B

Time Series Plots of Kinematic and Kinetic Variables

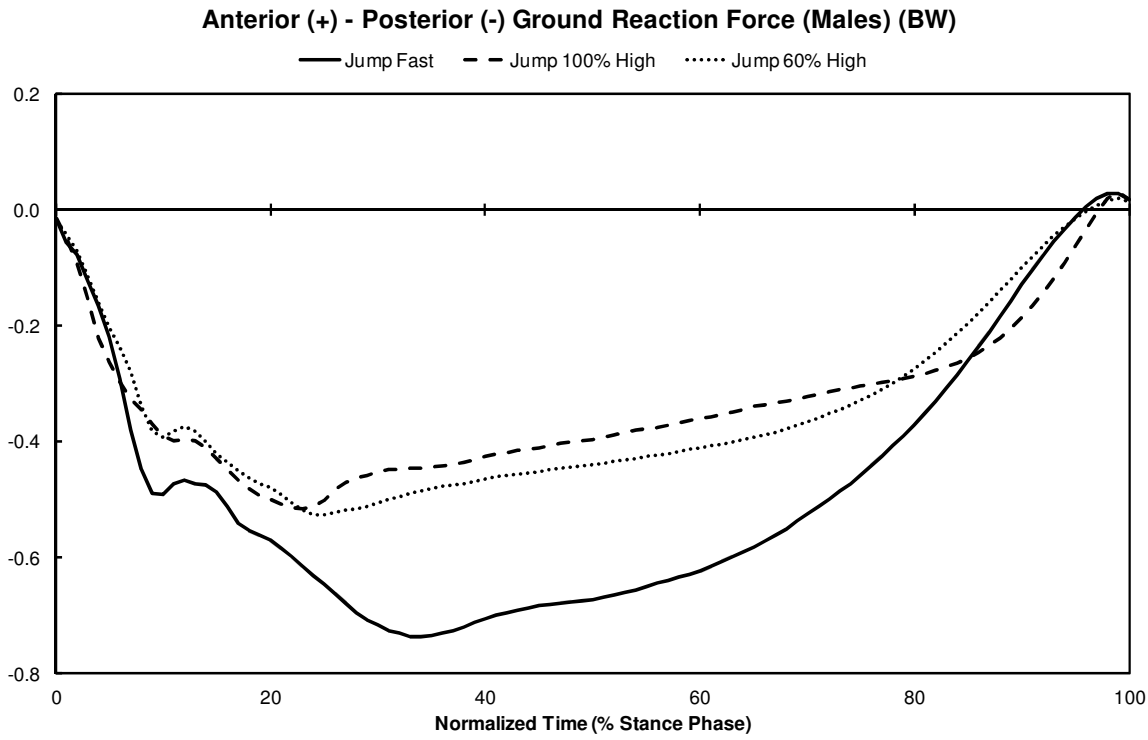

FIGURE B.1. Anterior (+) - posterior (-) ground reaction force during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

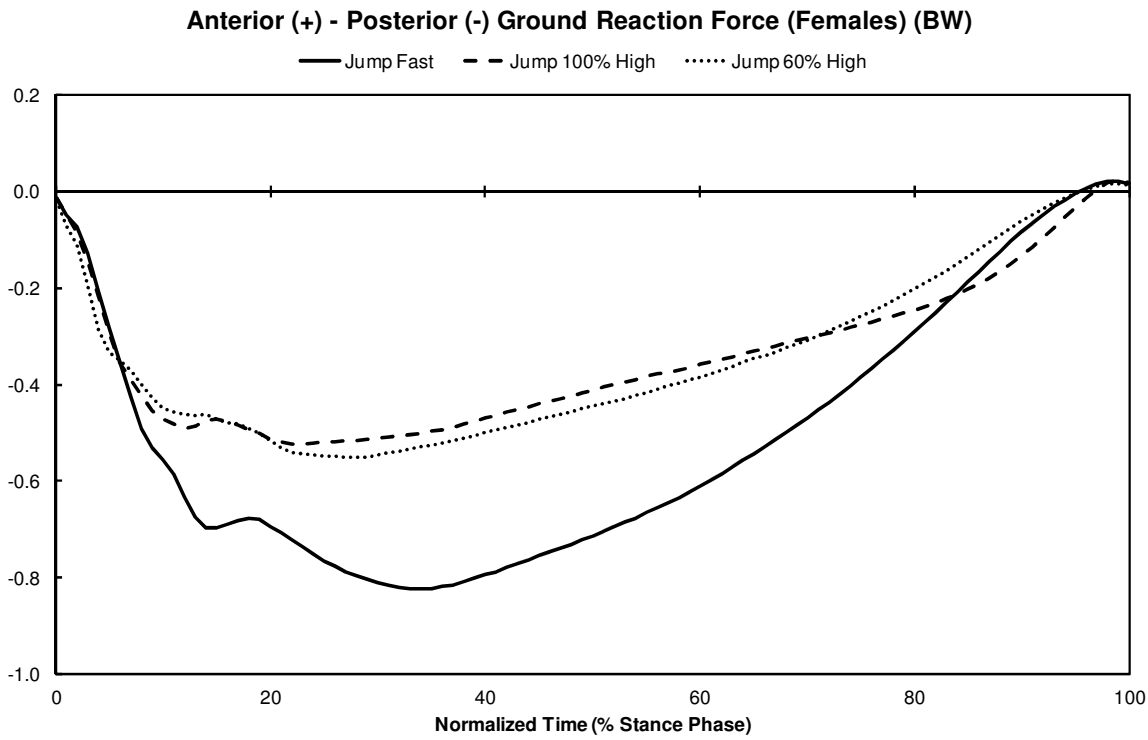

FIGURE B.2. Anterior (+) - posterior (-) ground reaction force during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

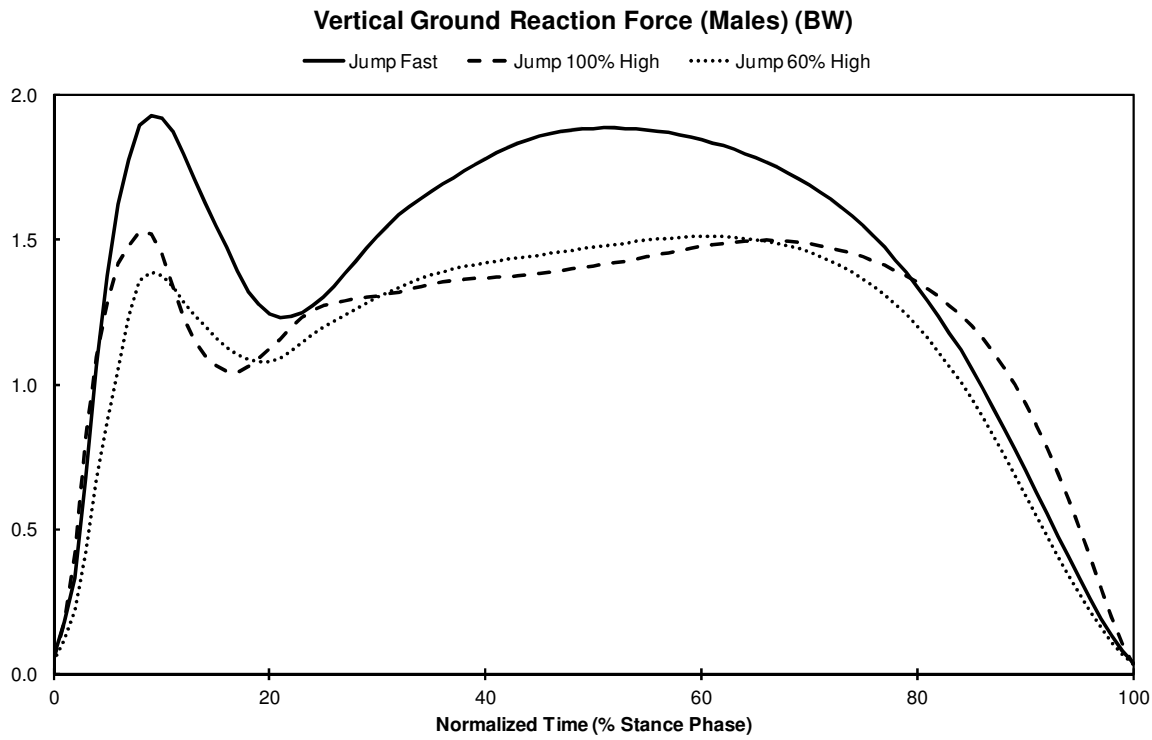

FIGURE B.3. Vertical ground reaction force during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

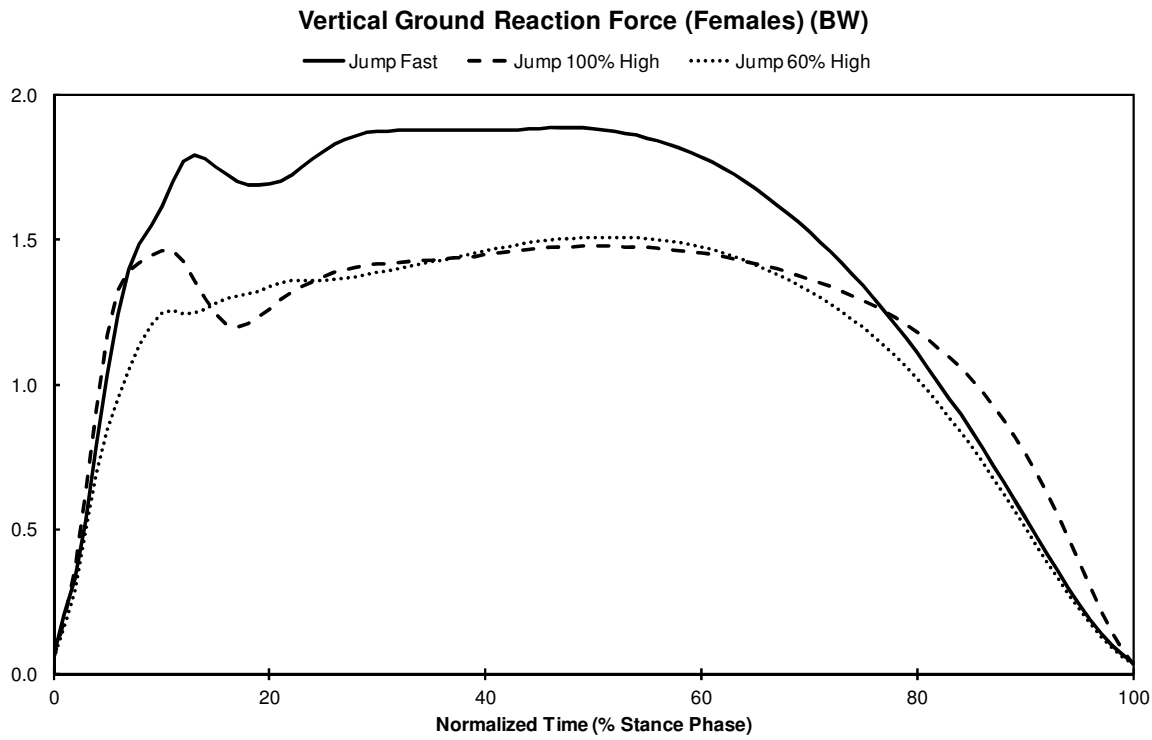

FIGURE B.4. Vertical ground reaction force during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

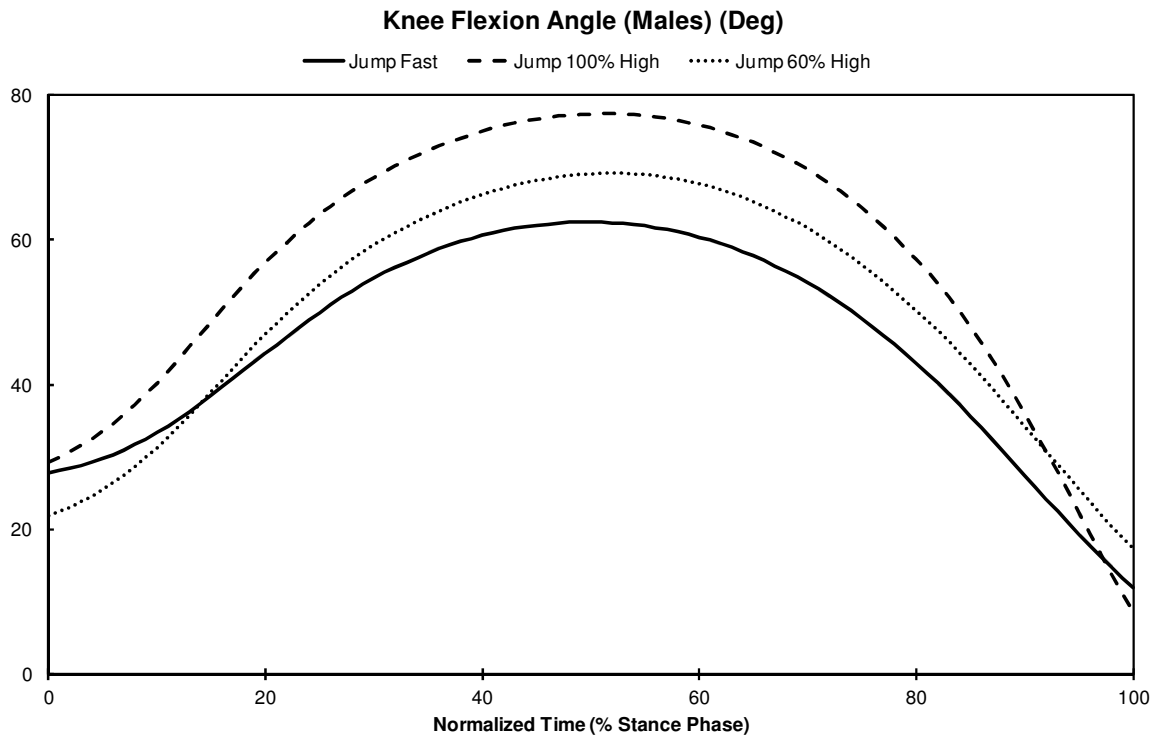

FIGURE B.5. Knee flexion angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

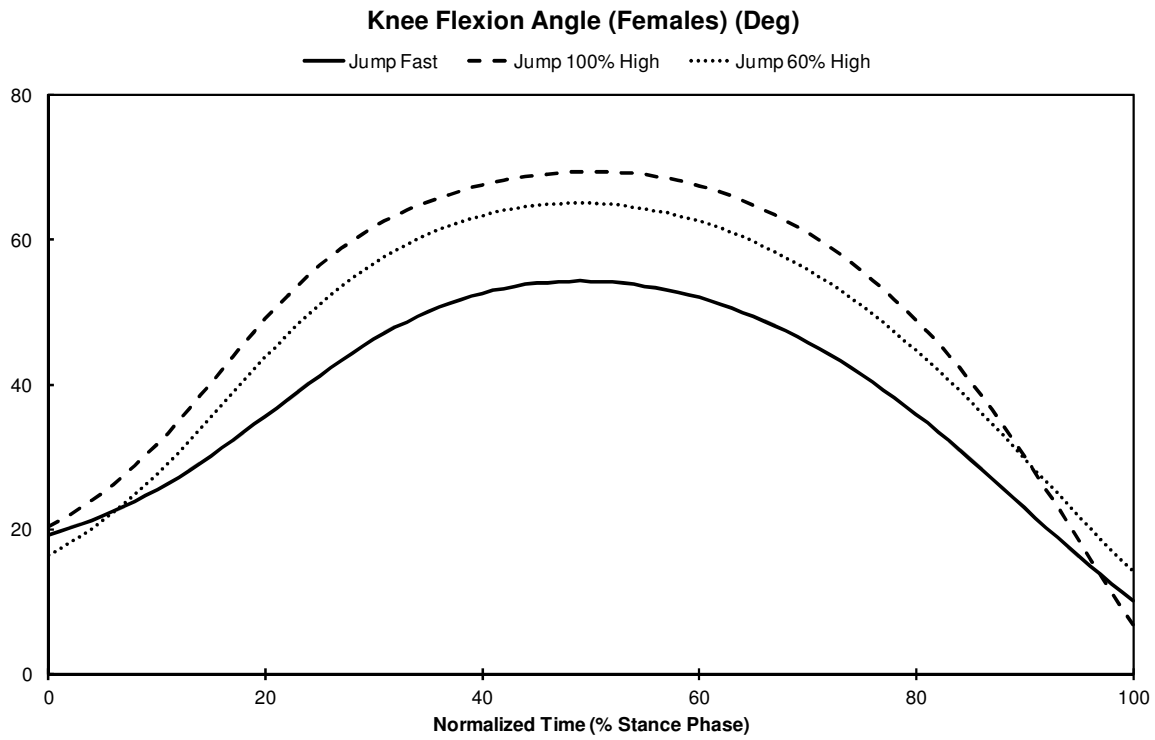

FIGURE B.6. Knee flexion angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

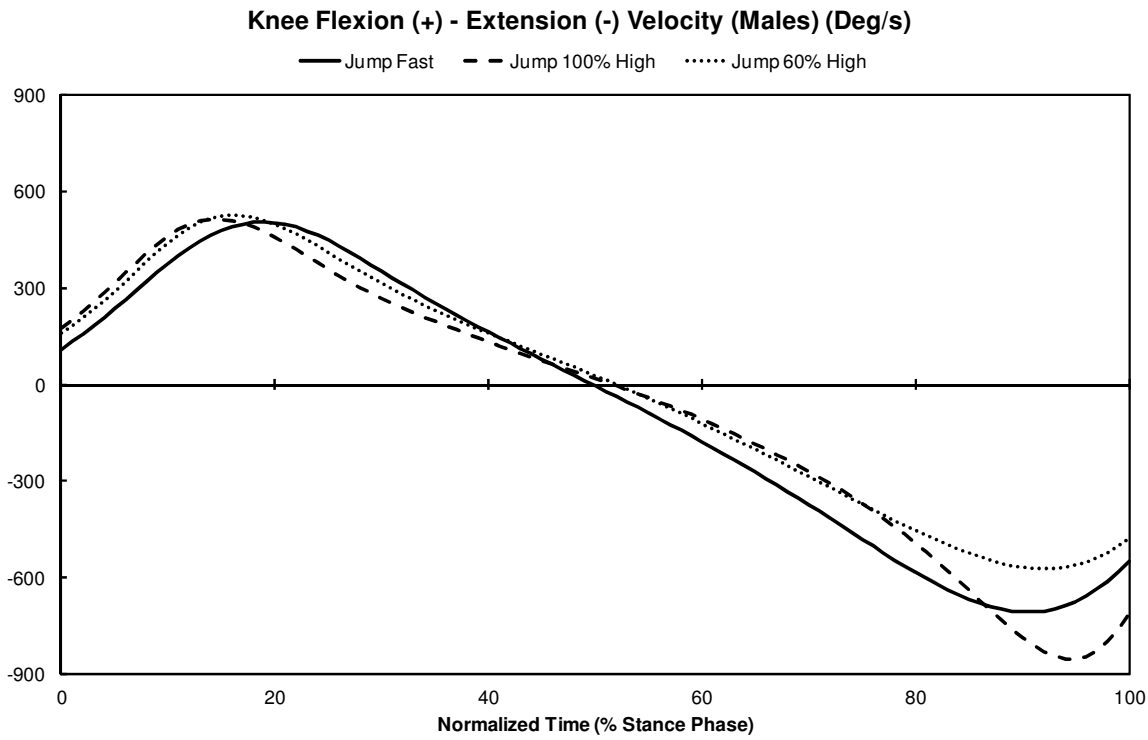

FIGURE B.7. Knee flexion (+) - extension (-) velocity during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

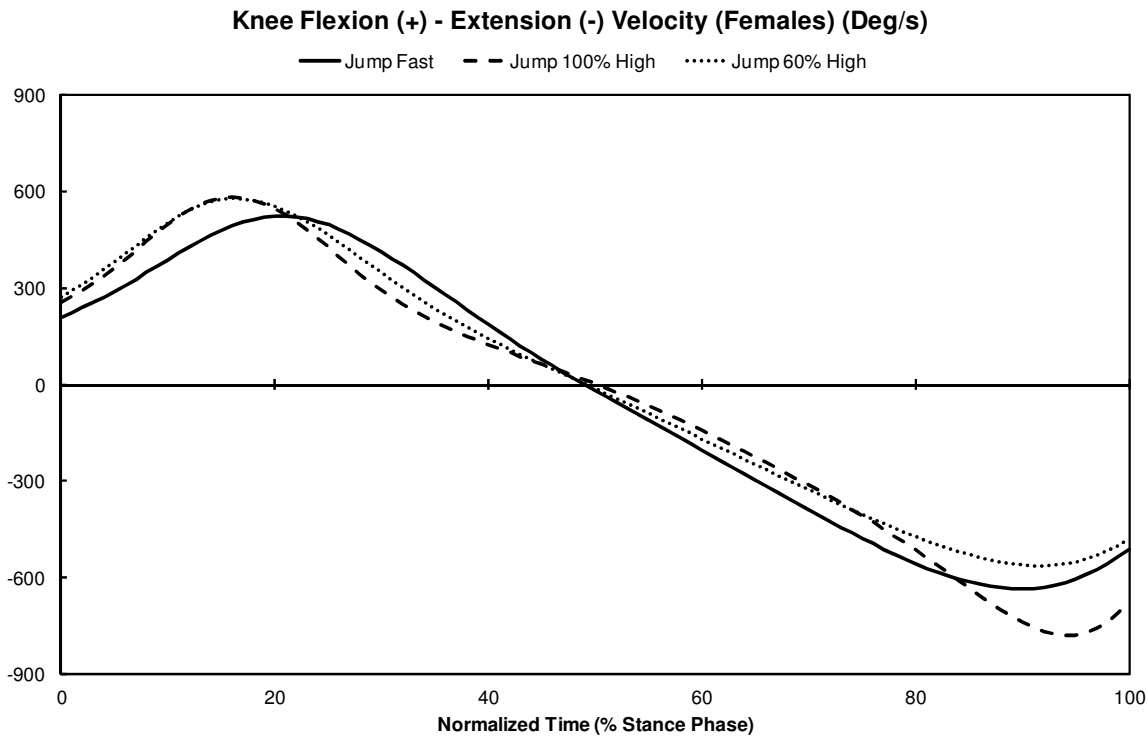

FIGURE B.8. Knee flexion (+) - extension (-) velocity during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

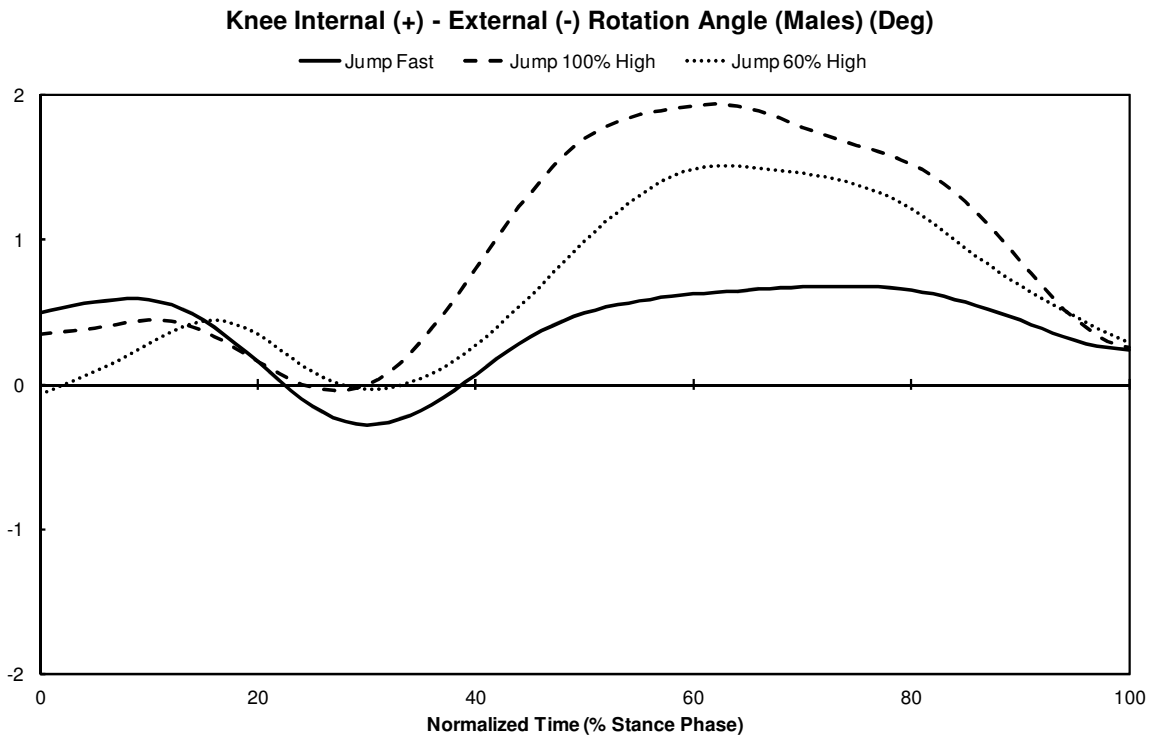

FIGURE B.9. Knee internal (+) - external (-) rotation angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

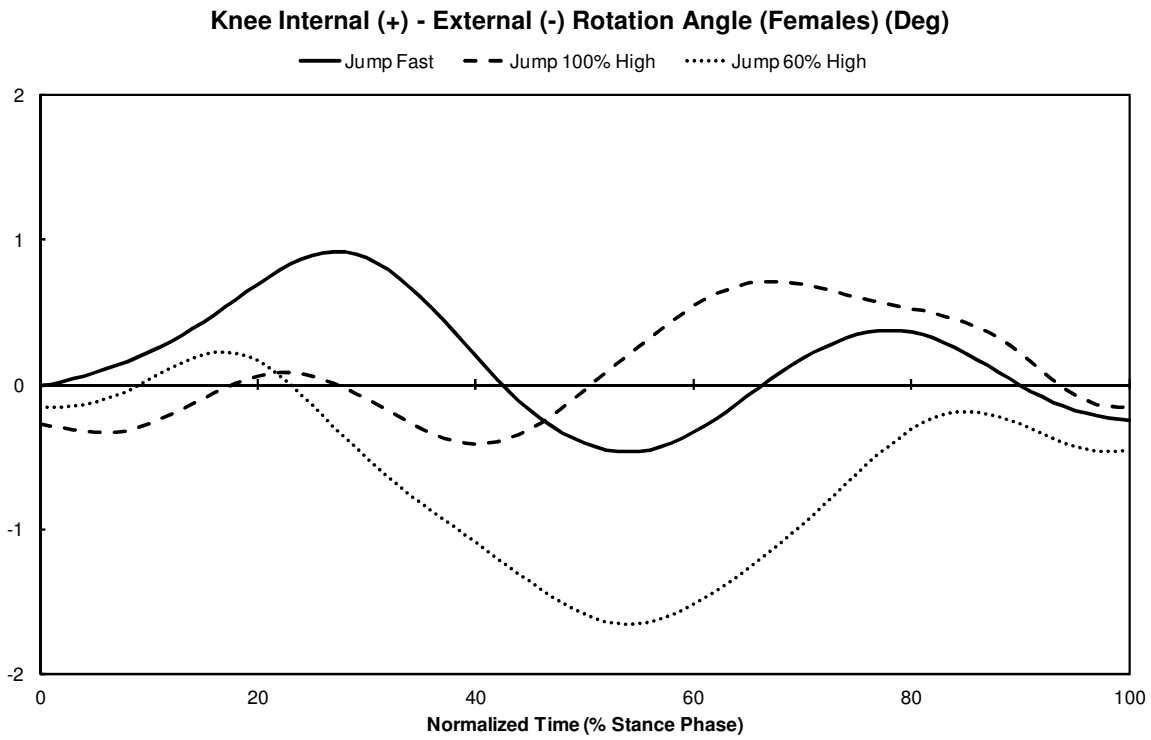

FIGURE B.10. Knee internal (+) - external (-) rotation angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

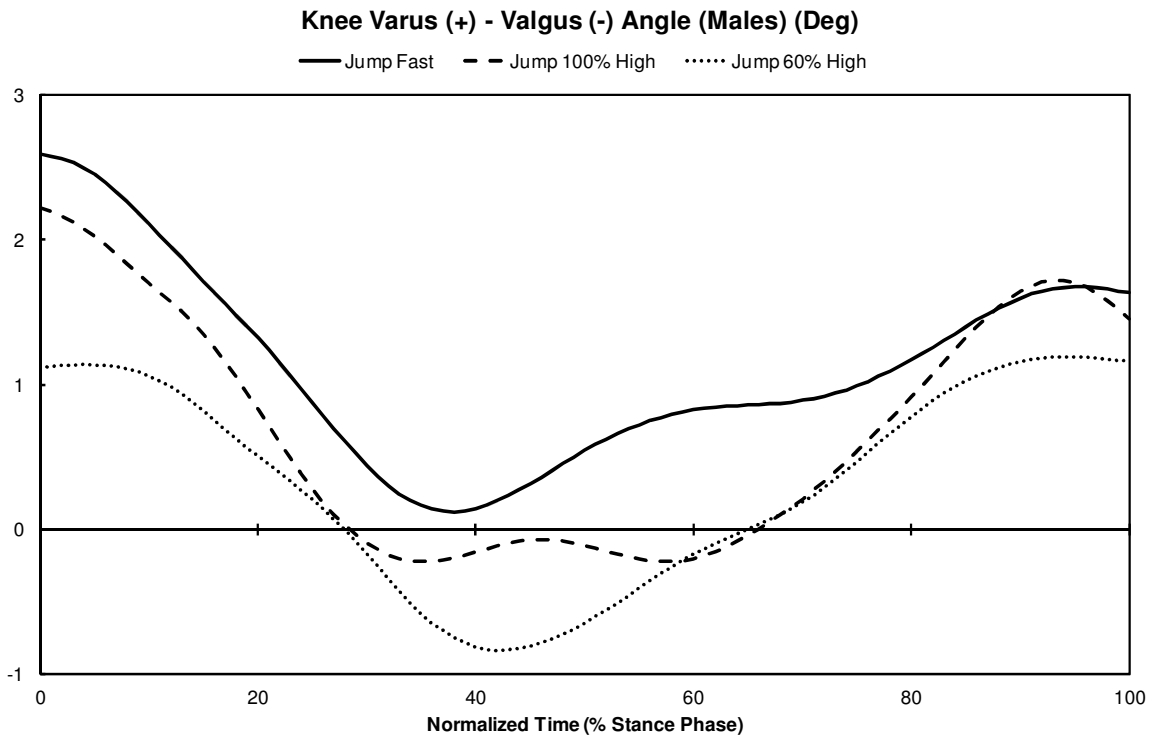

FIGURE B.11. Knee varus (+) - valgus (-) angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

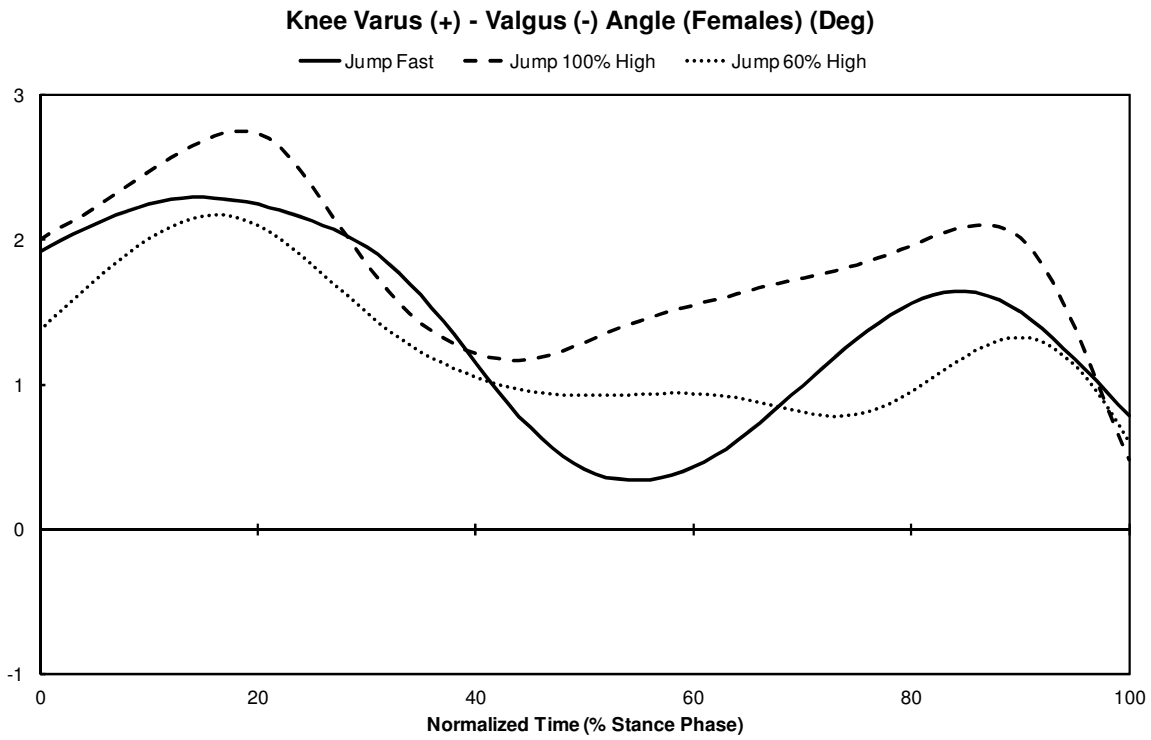

FIGURE B.12. Knee varus (+) - valgus (-) angle during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

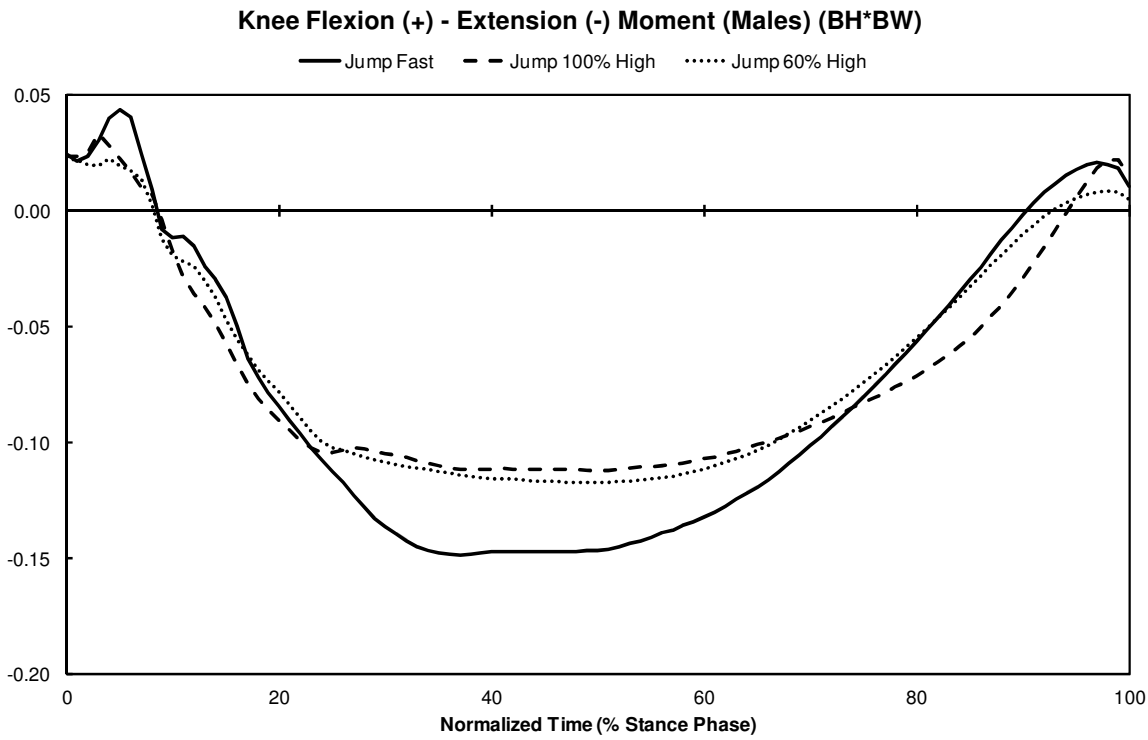

FIGURE B.13. Knee flexion (+) - extension (-) moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

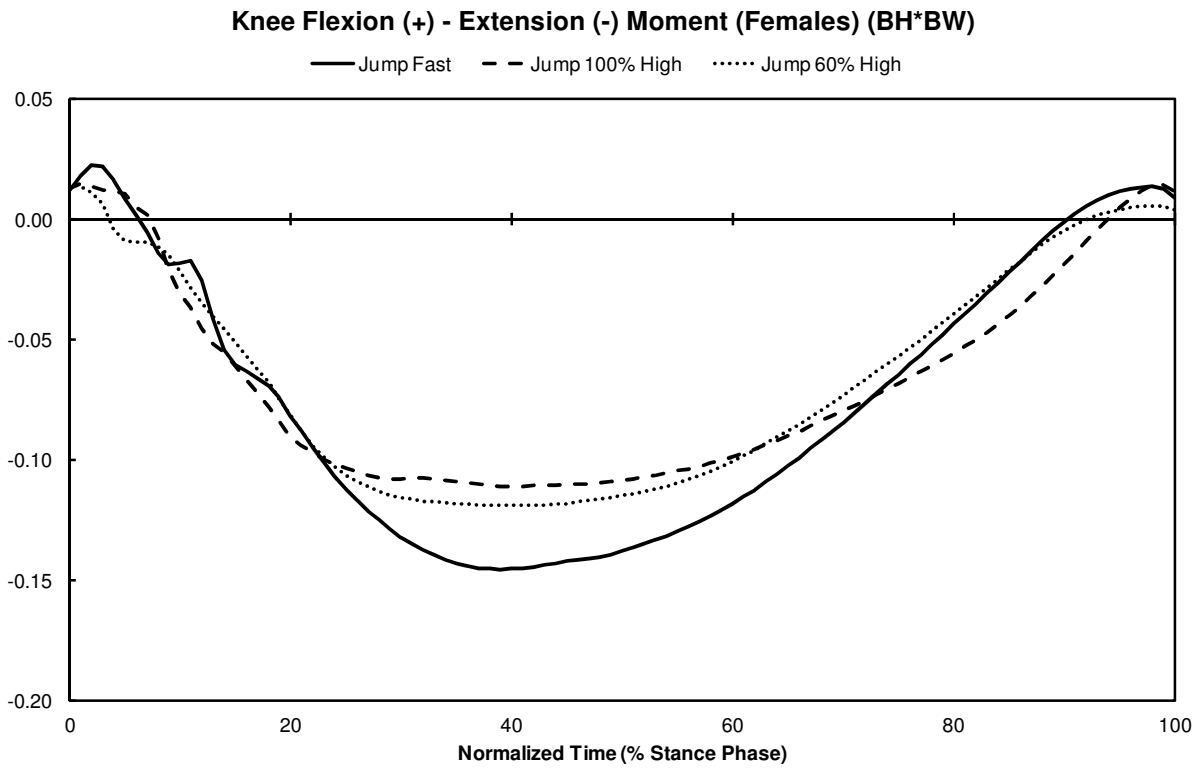

FIGURE B.14. Knee flexion (+) - extension (-) moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

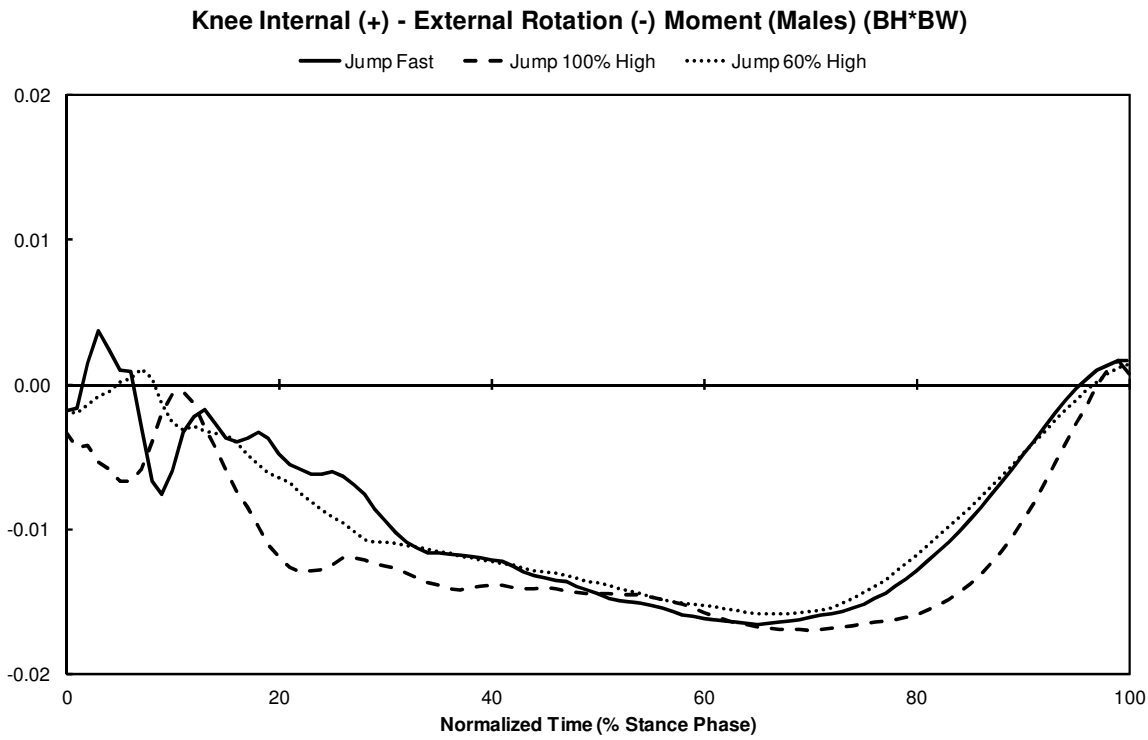

FIGURE B.15. Knee internal (+) - external (-) rotation moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

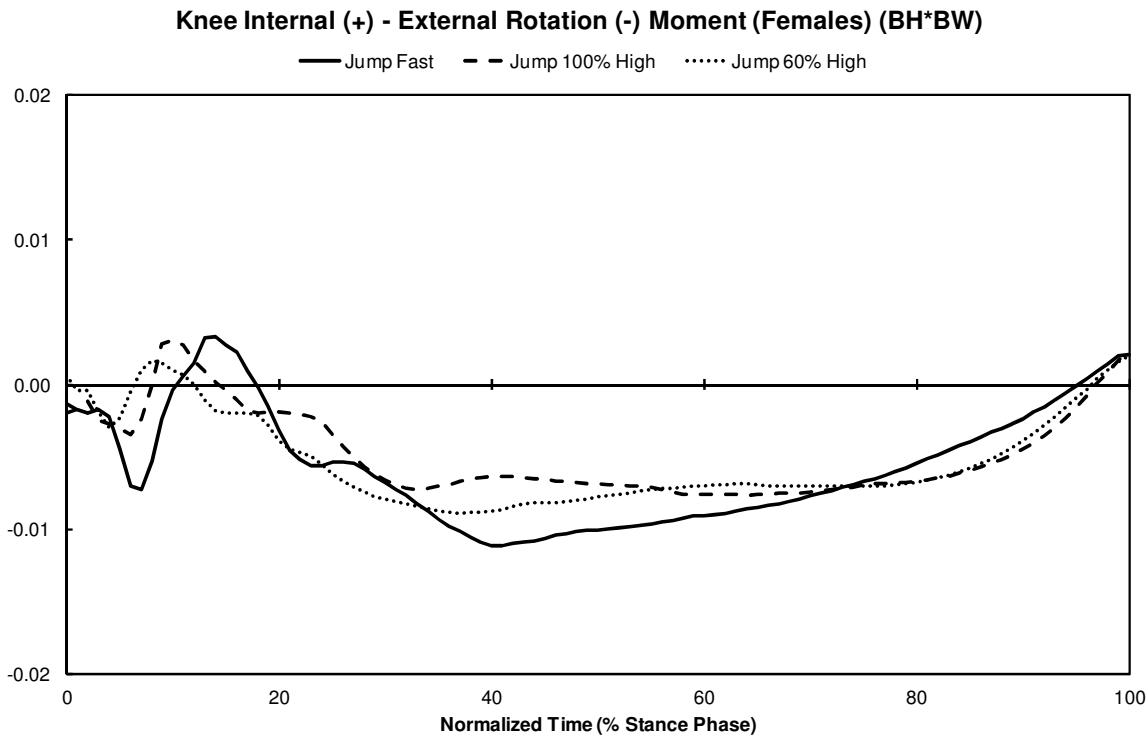

FIGURE B.16. Knee internal (+) - external (-) rotation moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

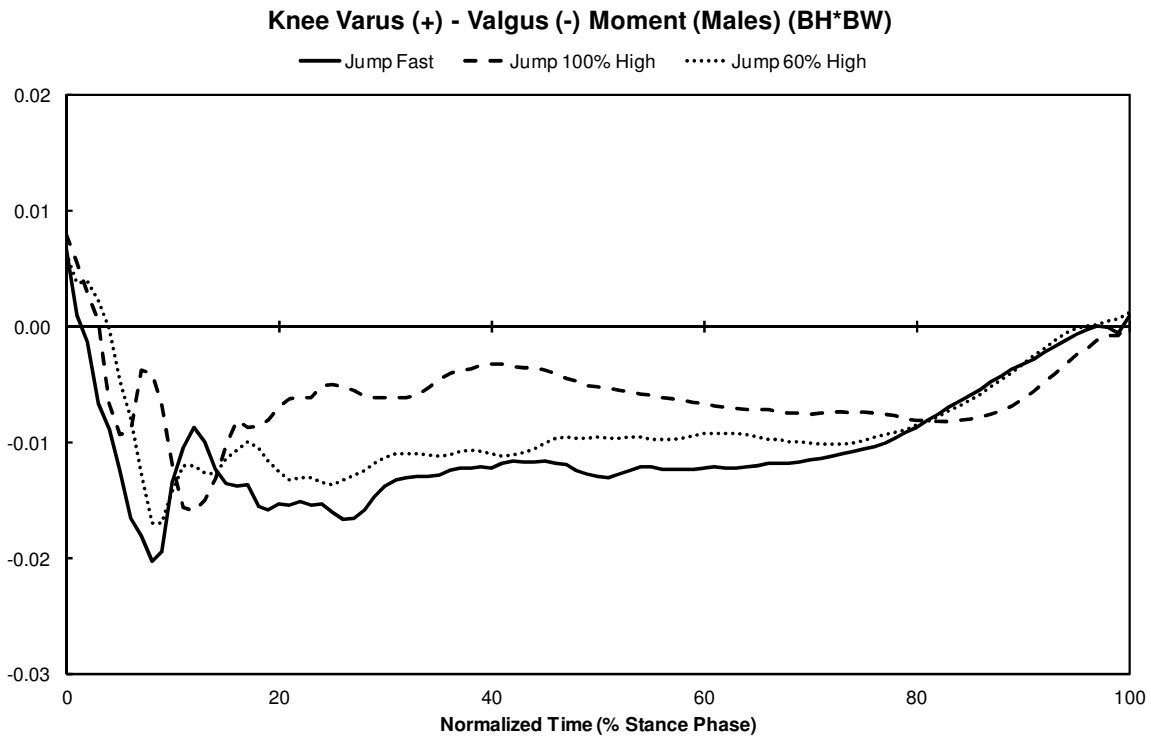

FIGURE B.17. Knee varus (+) - valgus (-) moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for males

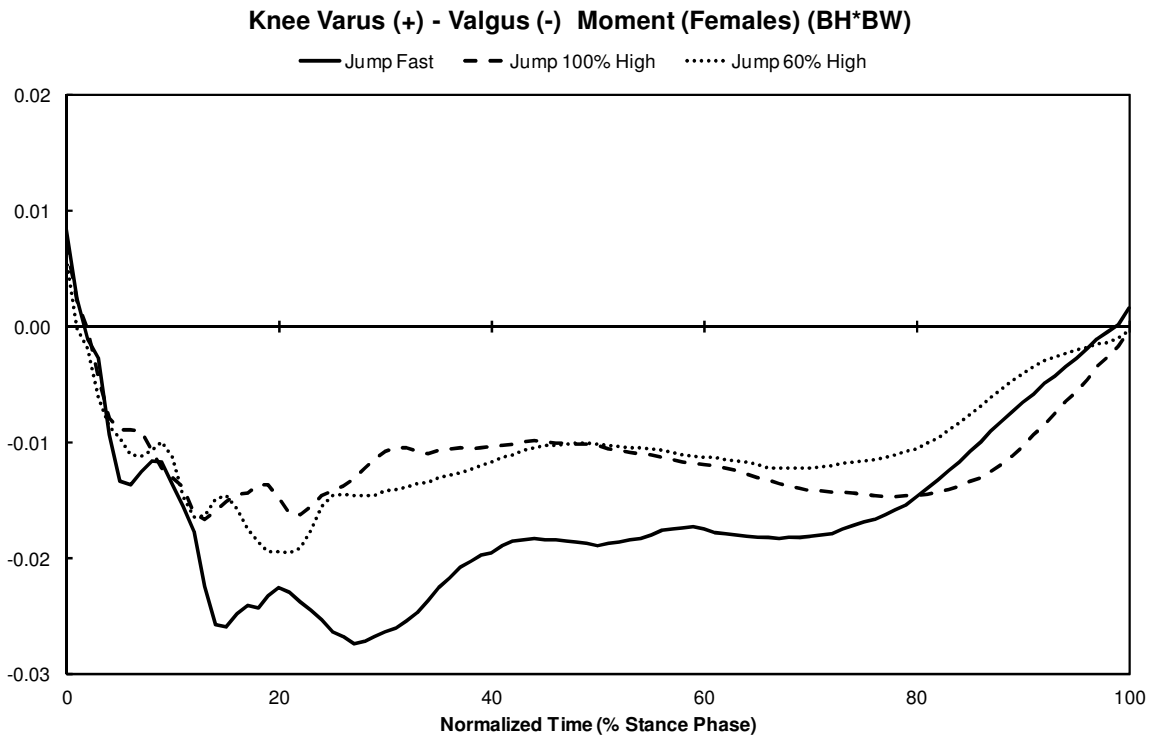

FIGURE B.18. Knee varus (+) - valgus (-) moment during jumping fast, jumping for maximum height, and jumping for 60% of maximum height conditions for females

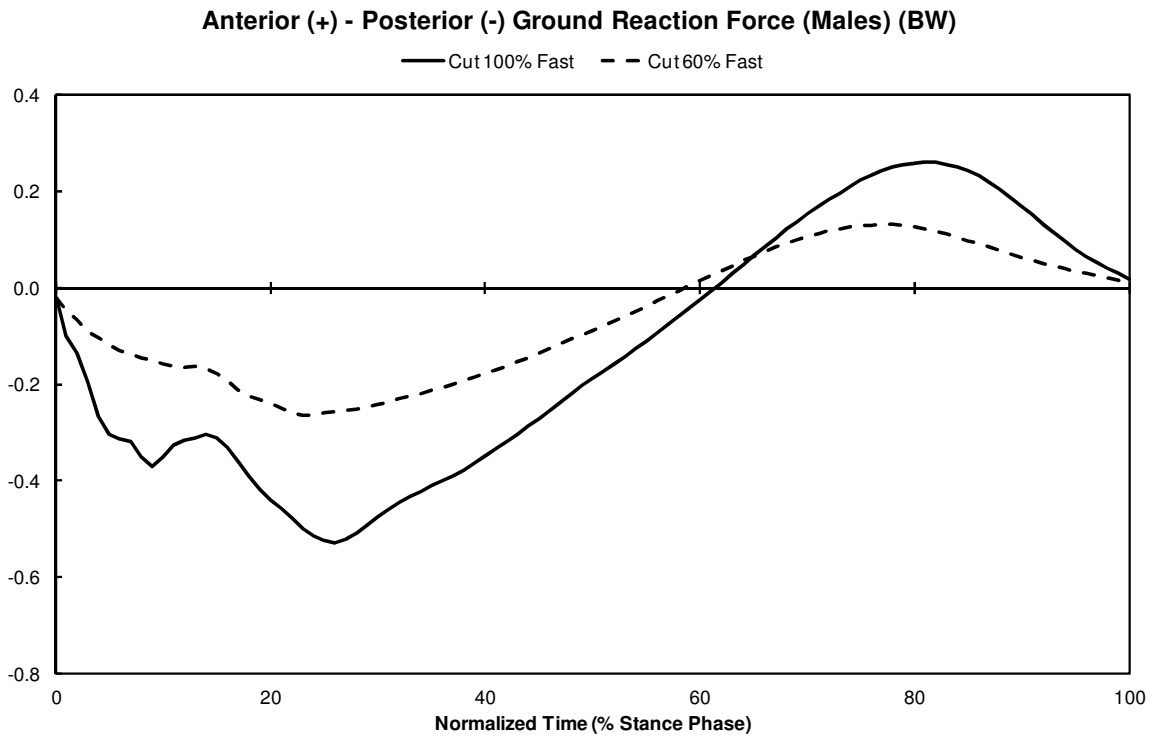

FIGURE B.19. Anterior (+) - posterior (-) ground reaction force during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

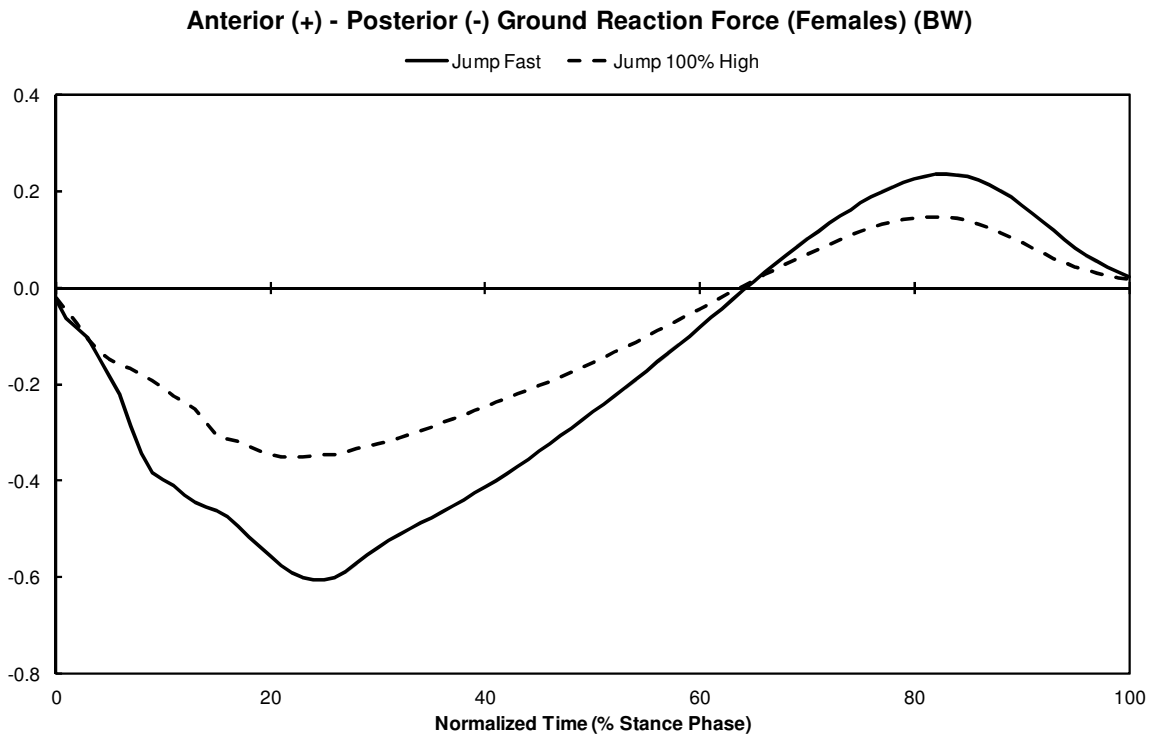

FIGURE B.20. Anterior (+) - posterior (-) ground reaction force during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

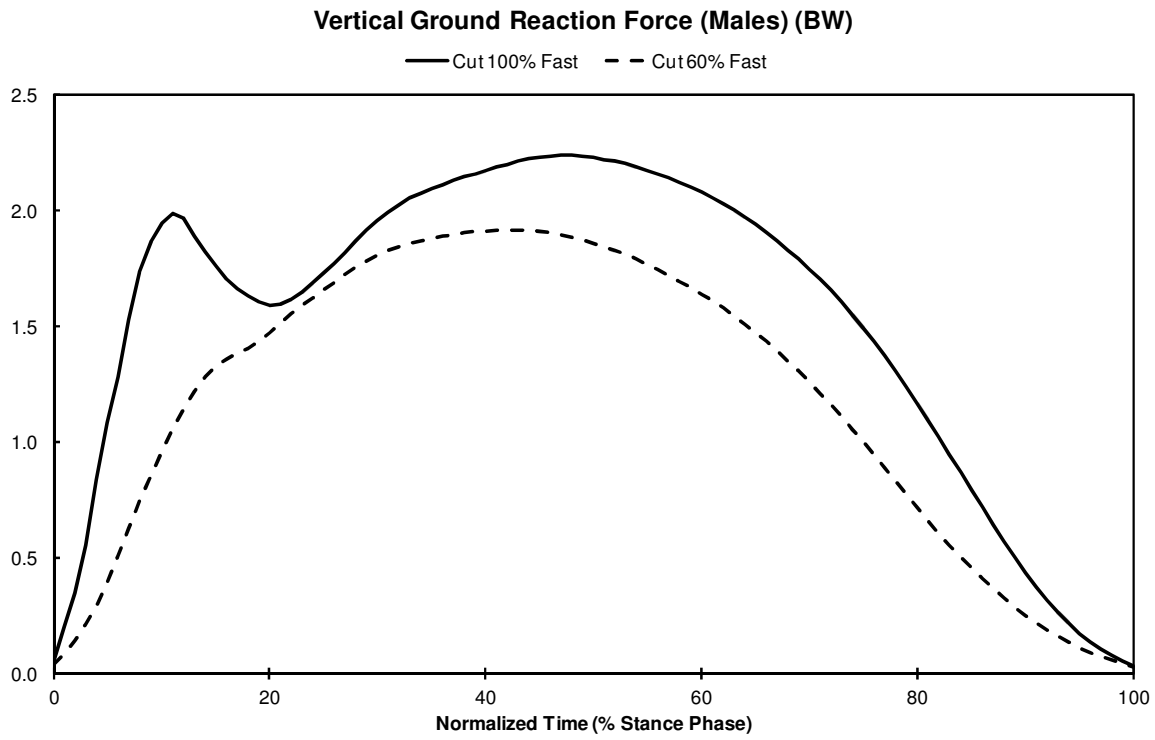

FIGURE B.21. Vertical ground reaction force during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

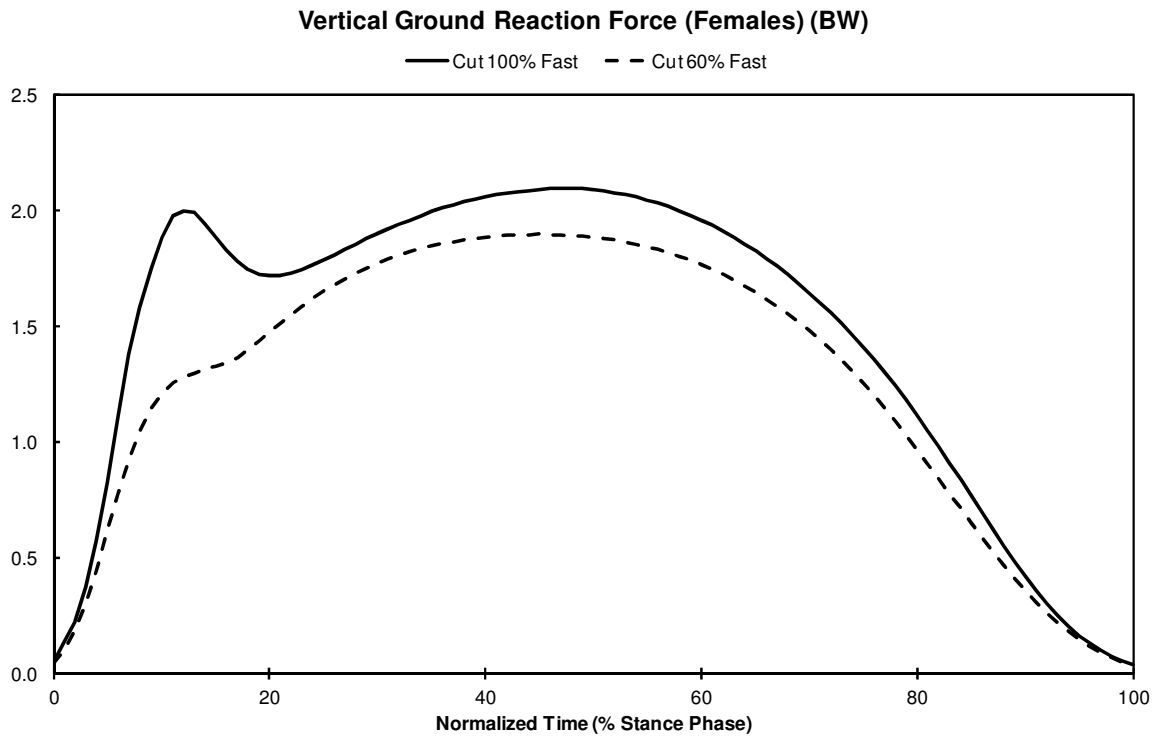

FIGURE B.22. Vertical ground reaction force during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

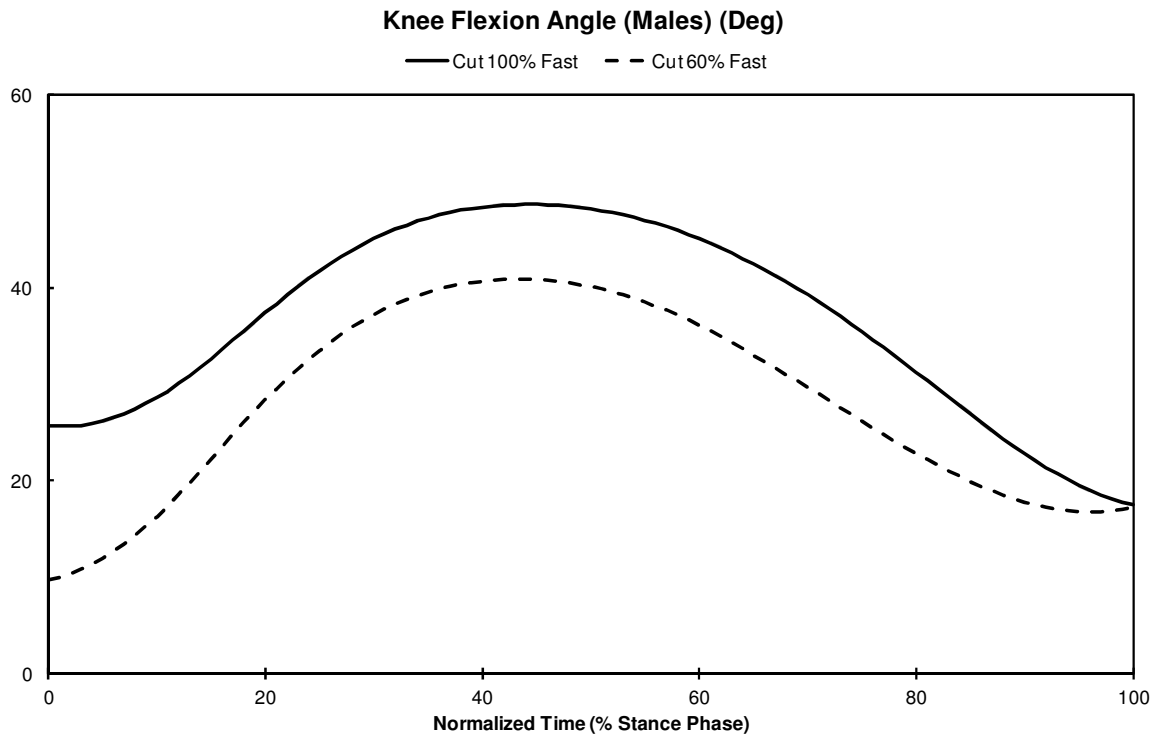

FIGURE B.23. Knee flexion angle during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

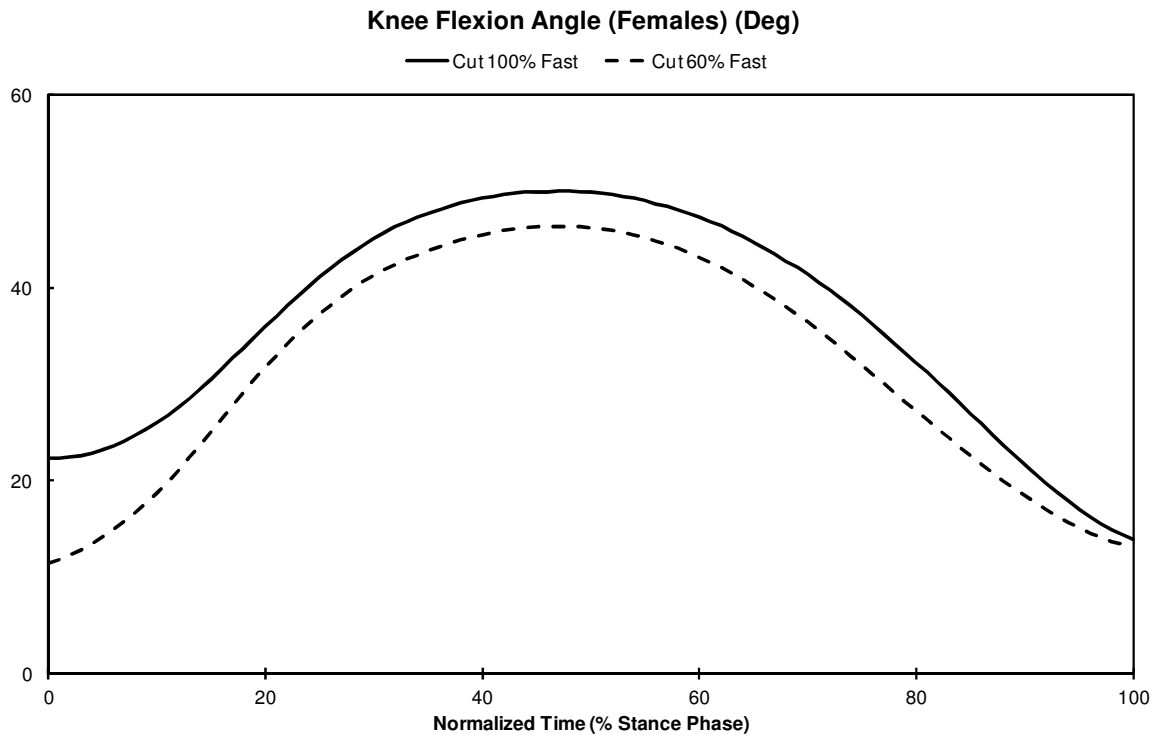

FIGURE B.24. Knee flexion angle during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

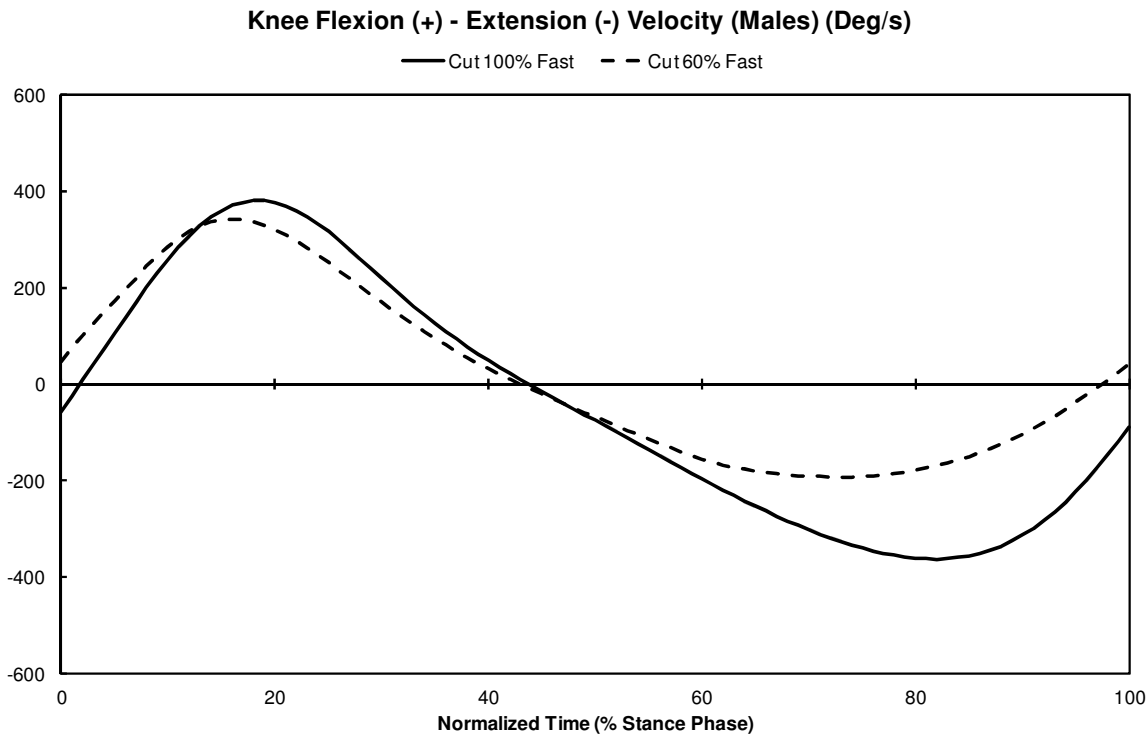

FIGURE B.25. Knee flexion (+) - extension (-) velocity during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

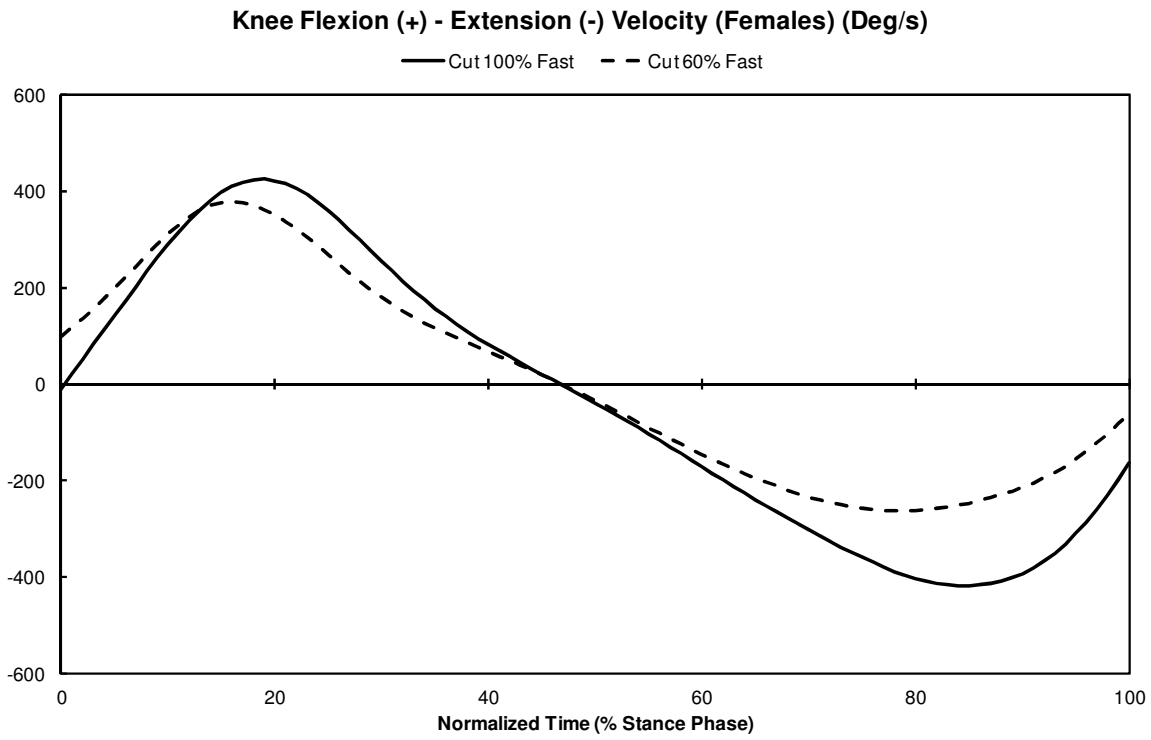

FIGURE B.26. Knee flexion (+) - extension (-) velocity during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

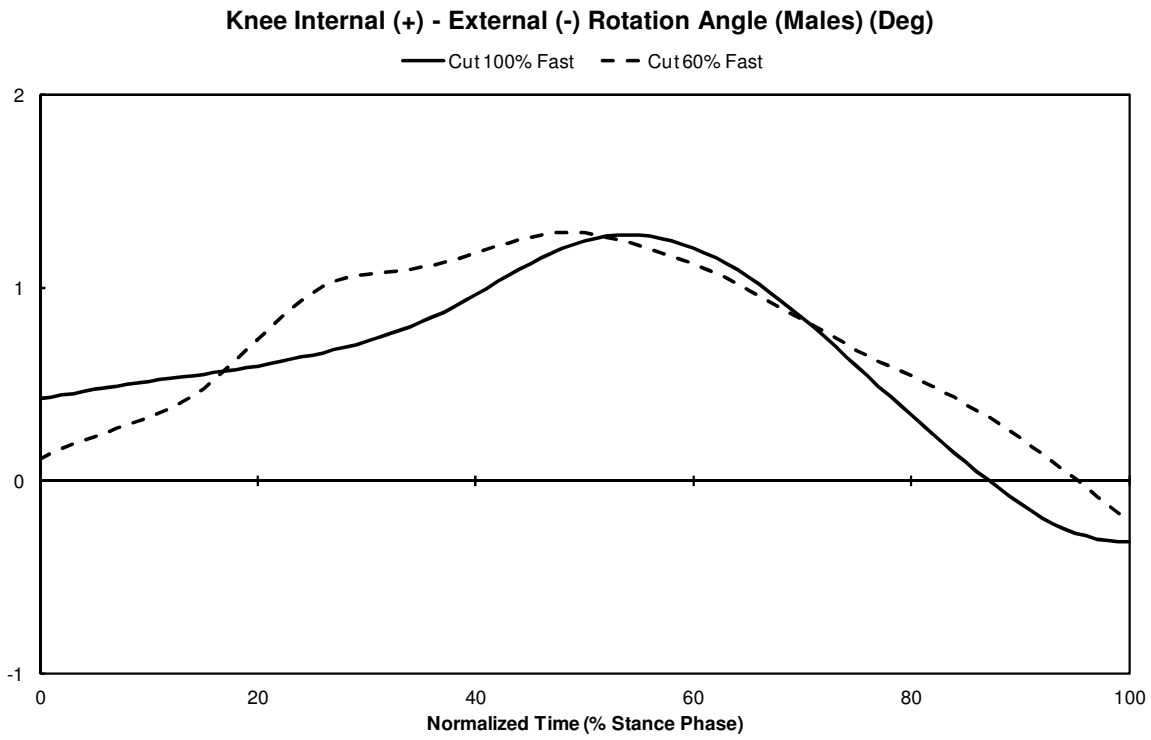

FIGURE B.27. Knee internal (+) - external (-) rotation angle during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

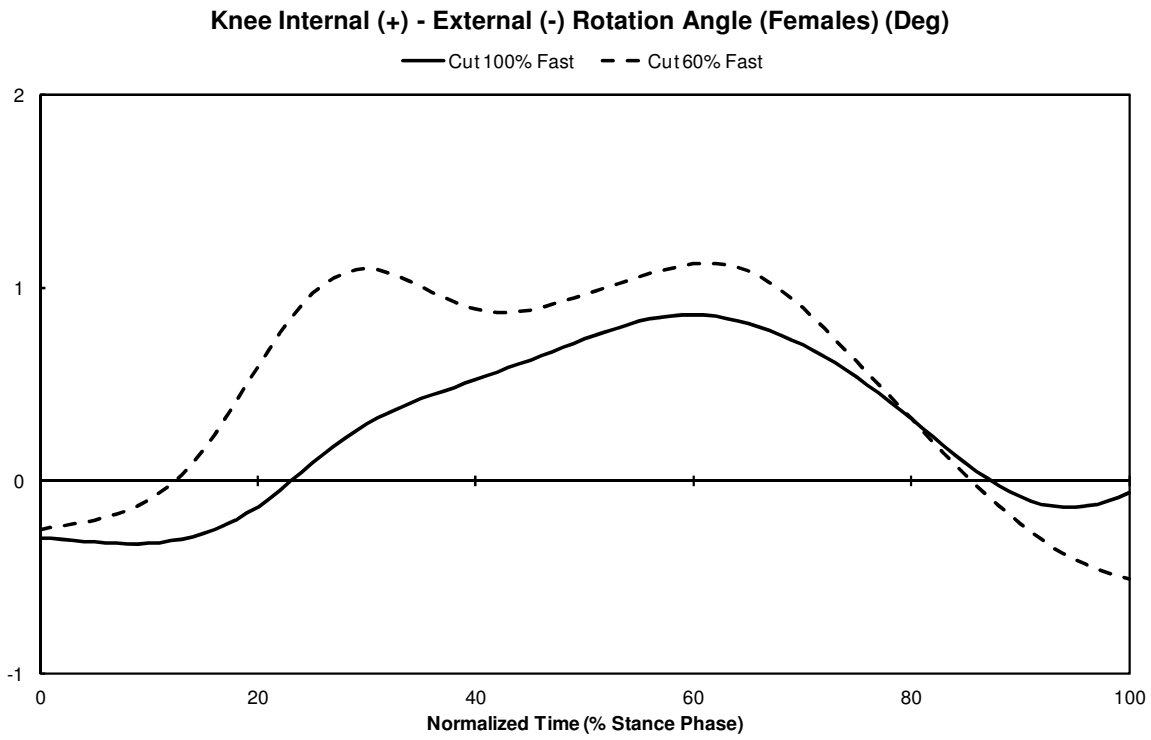

FIGURE B.28. Knee internal (+) - external (-) rotation angle during cutting with maximum speed and cutting with 60% of maximum speed for females

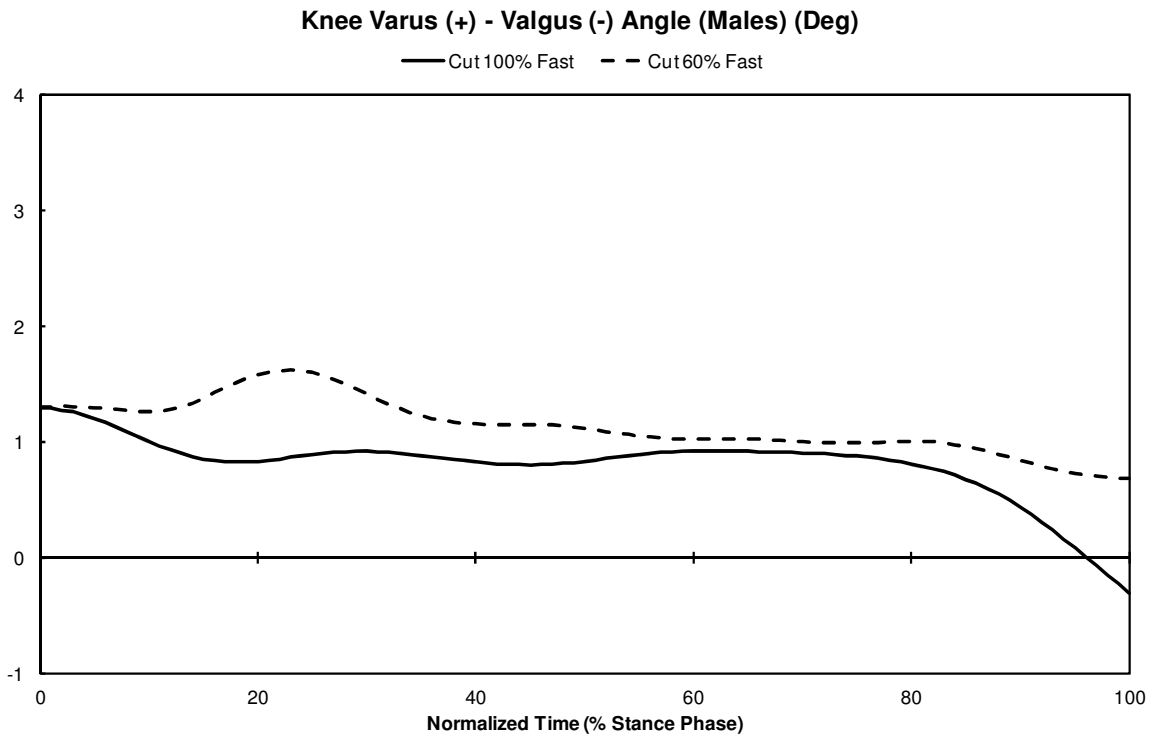

FIGURE B.29. Knee varus (+) - valgus (-) angle during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

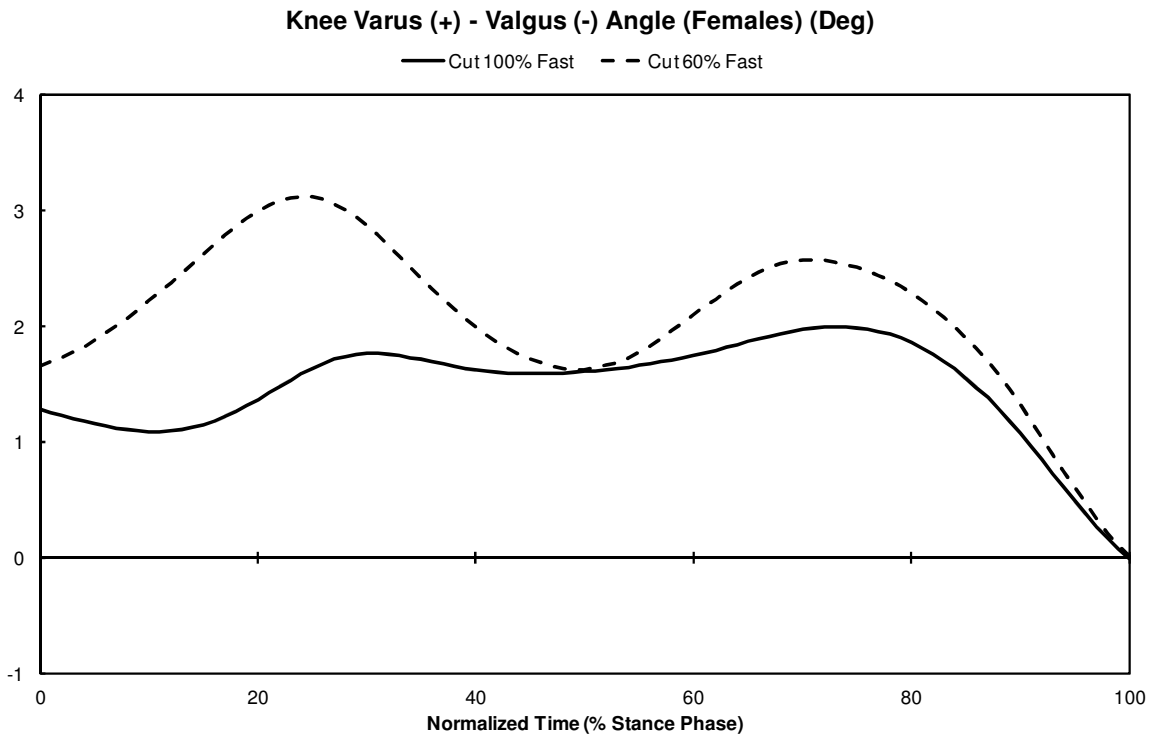

FIGURE B.30. Knee varus (+) - valgus (-) angle during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

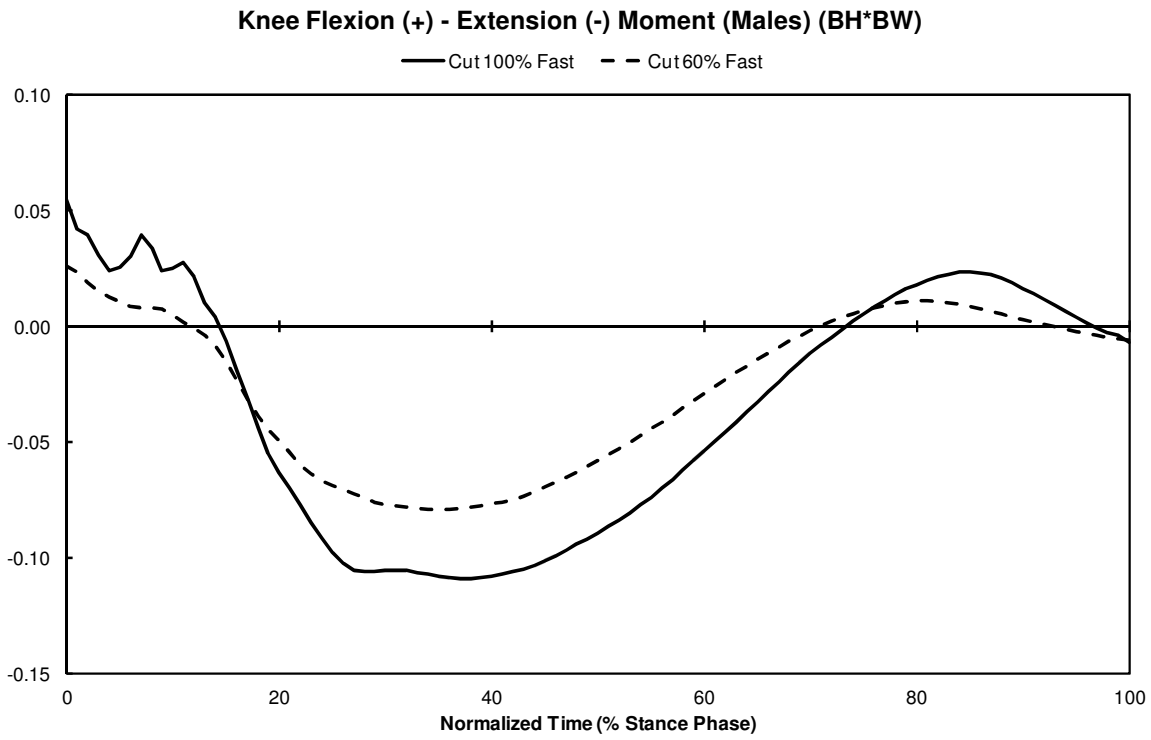

FIGURE B.31. Knee flexion (+) - extension (-) moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

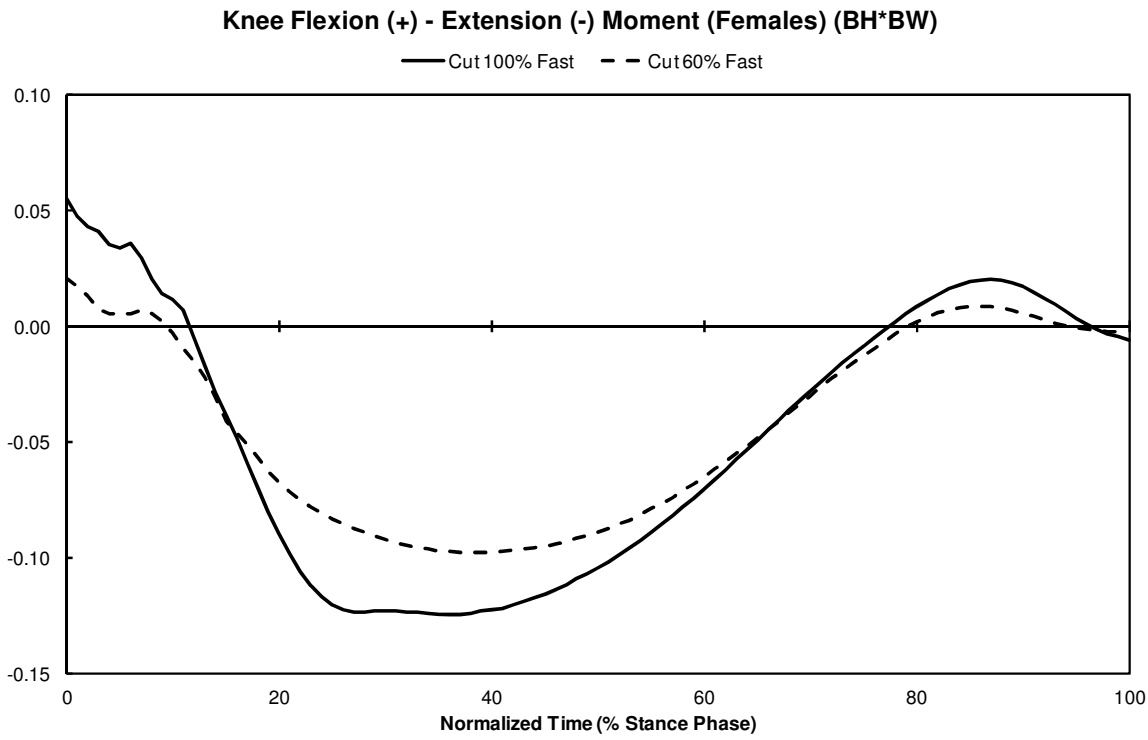

FIGURE B.32. Knee flexion (+) - extension (-) moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

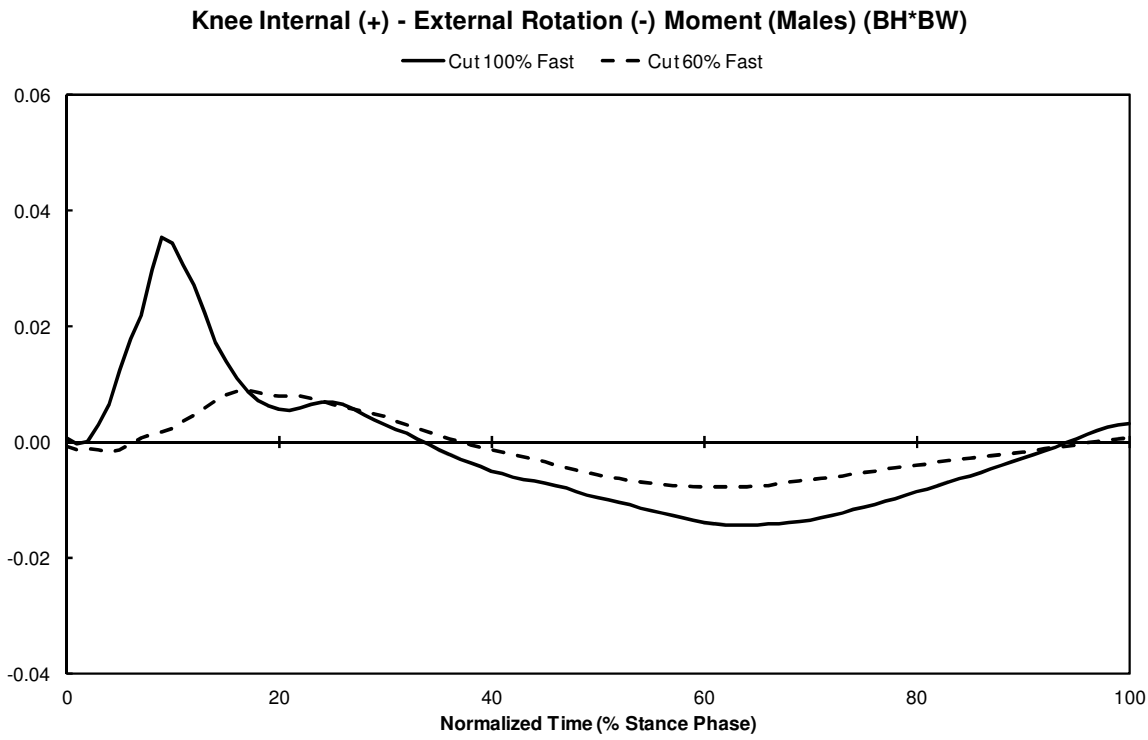

FIGURE B.33. Knee internal (+) - external (-) rotation moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

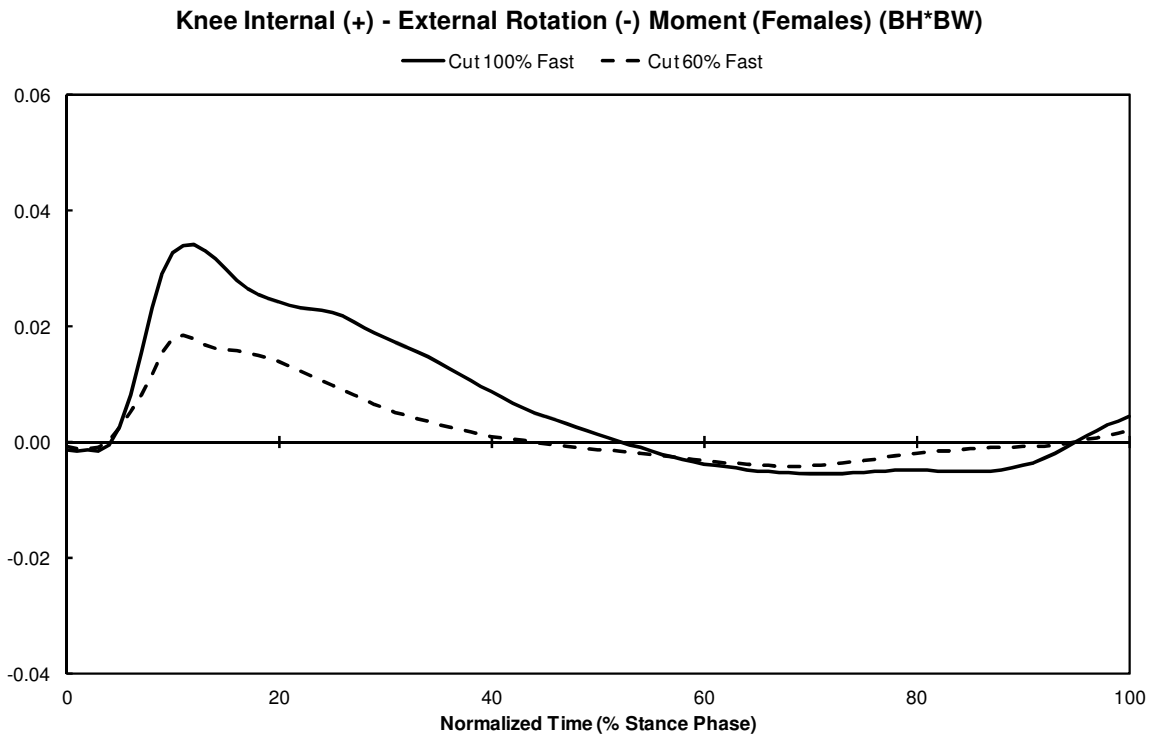

FIGURE B.34. Knee internal (+) - external (-) rotation moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

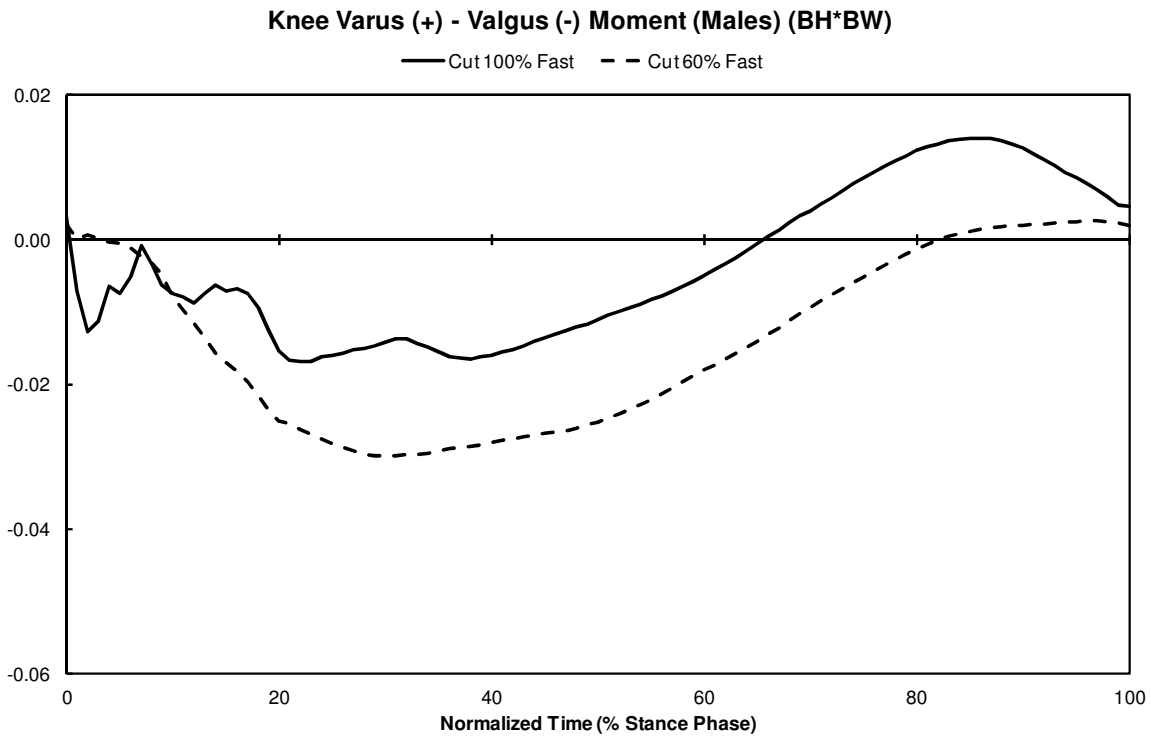

FIGURE B.35. Knee varus (+) - valgus (-) moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for males

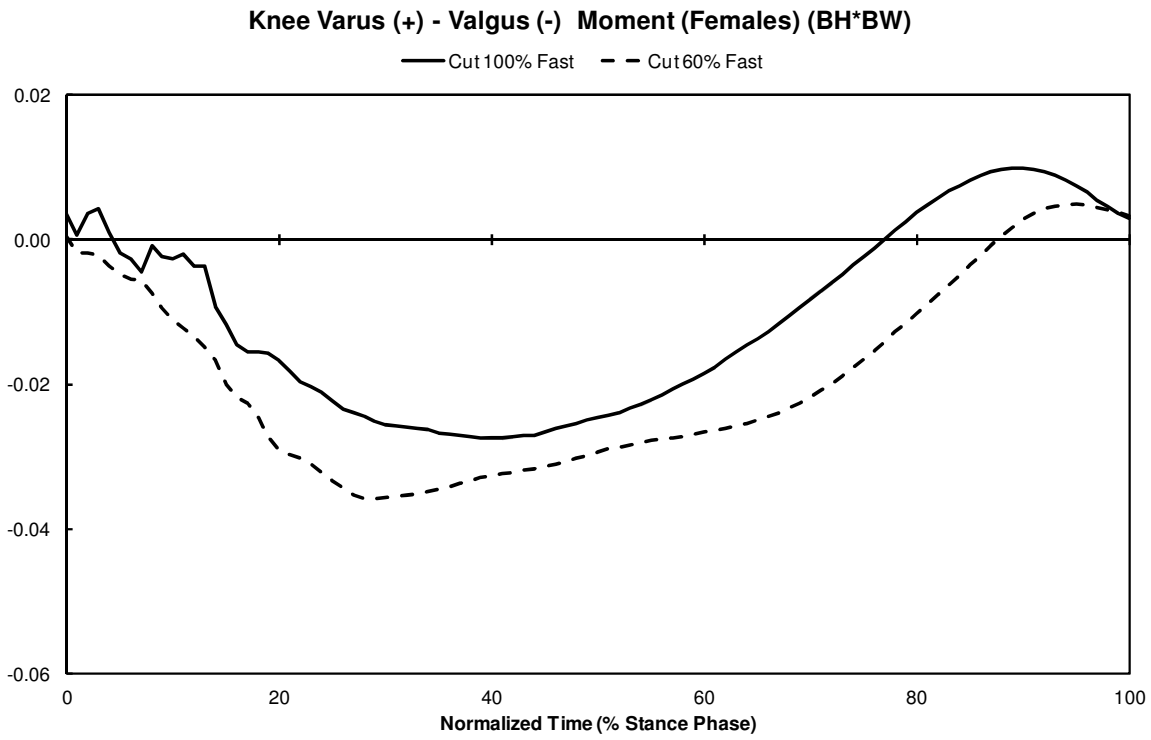

FIGURE B.36. Knee varus (+) - valgus (-) moment during cutting with maximum speed and cutting with 60% of maximum speed conditions for females

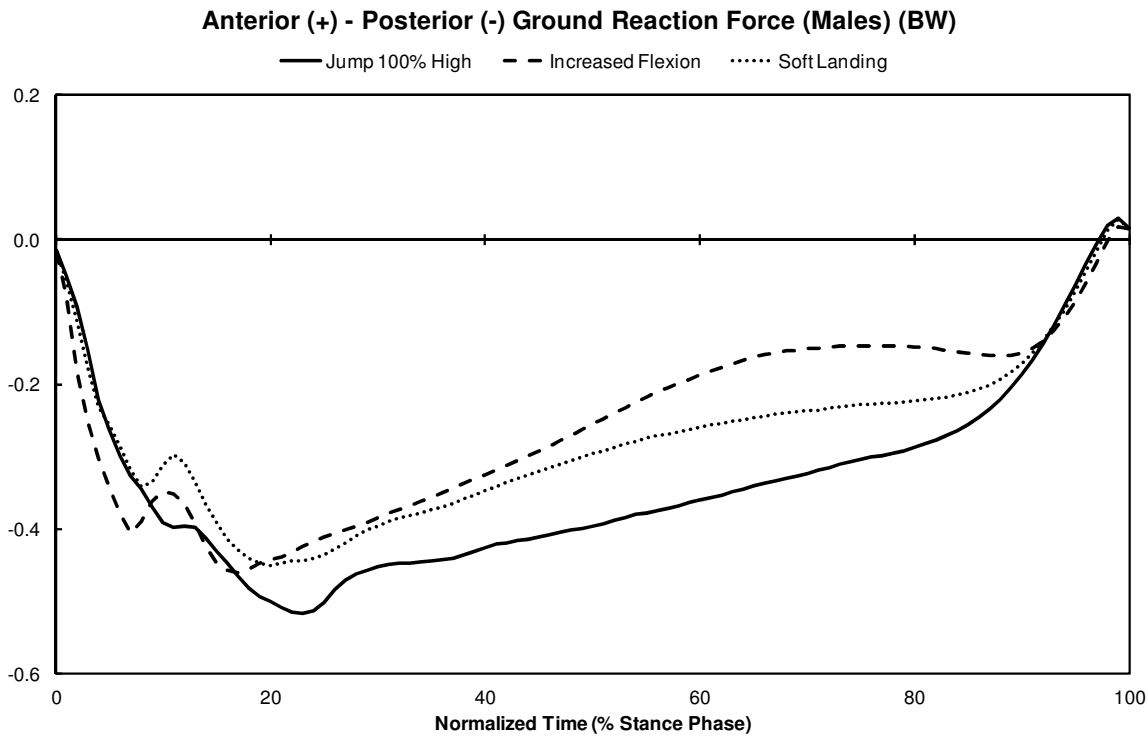

FIGURE B.37. Anterior (+) - posterior (-) ground reaction force during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

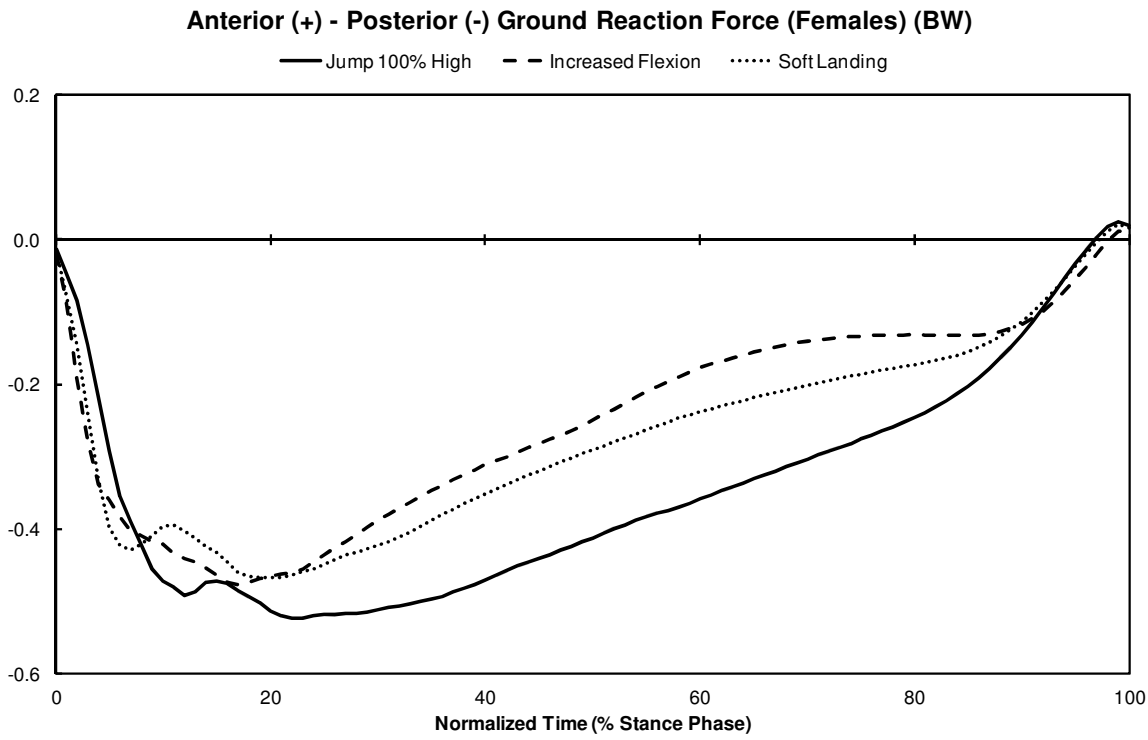

FIGURE B.38. Anterior (+) - posterior (-) ground reaction force during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

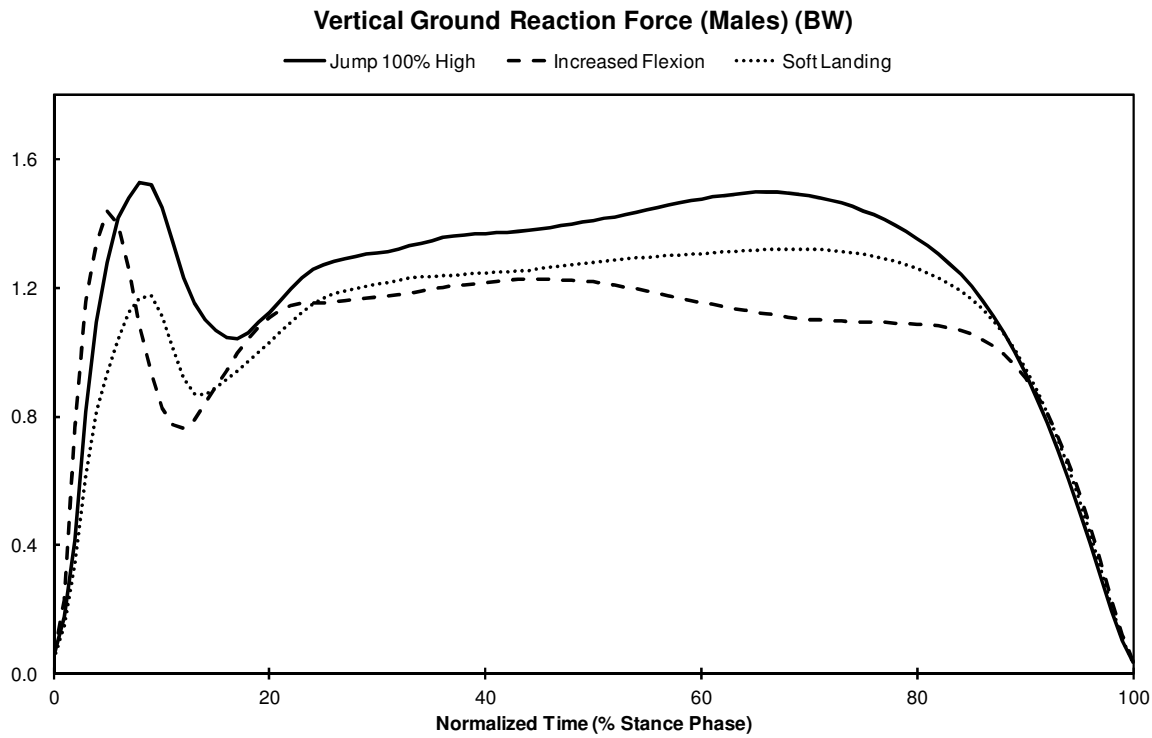

FIGURE B.39. Vertical ground reaction force during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

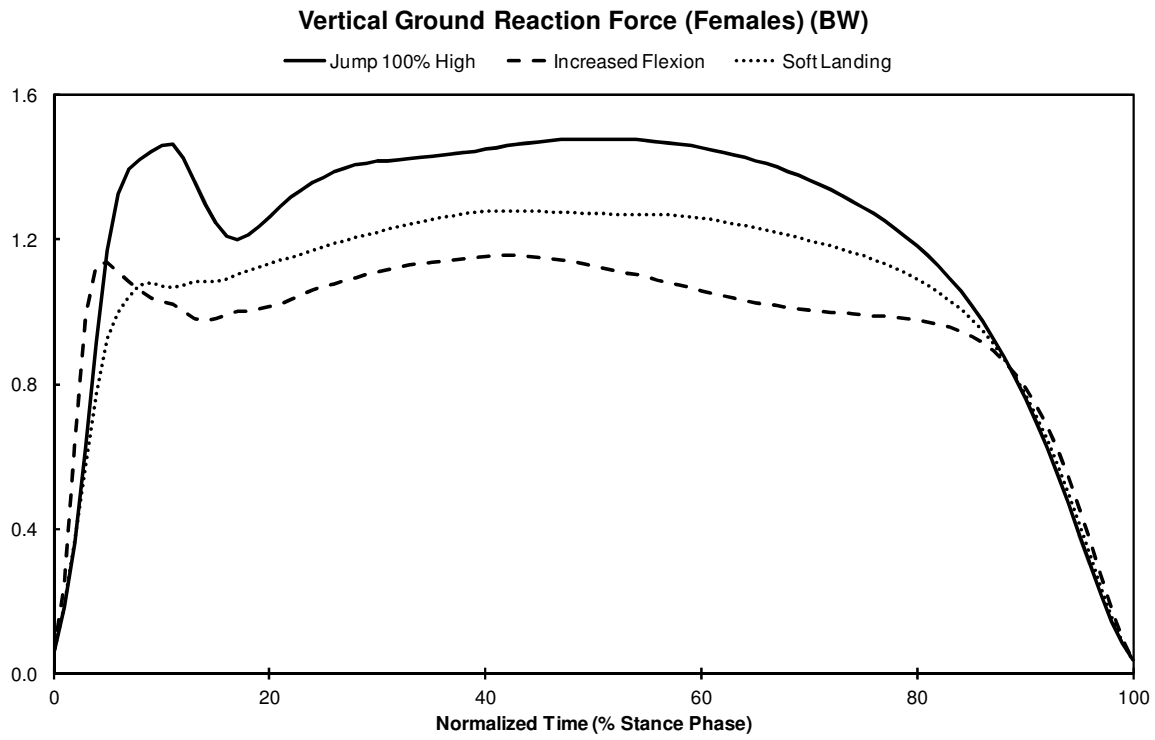

FIGURE B.40. Vertical ground reaction force during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

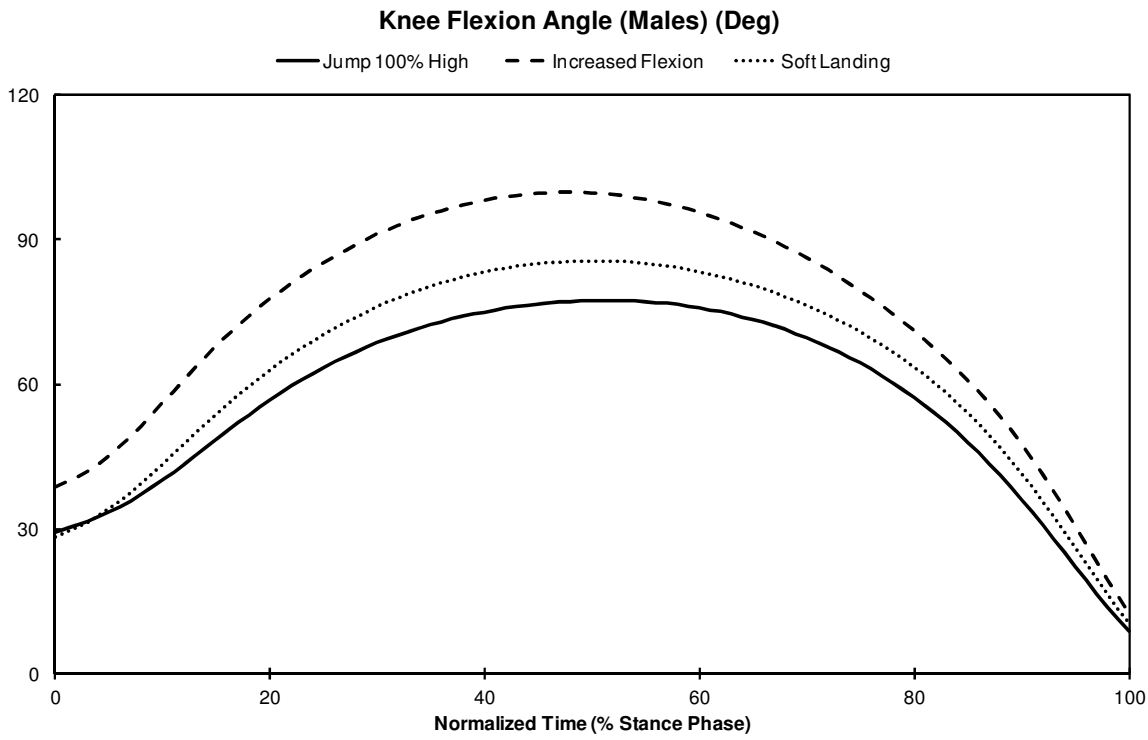

FIGURE B.41. Knee flexion angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

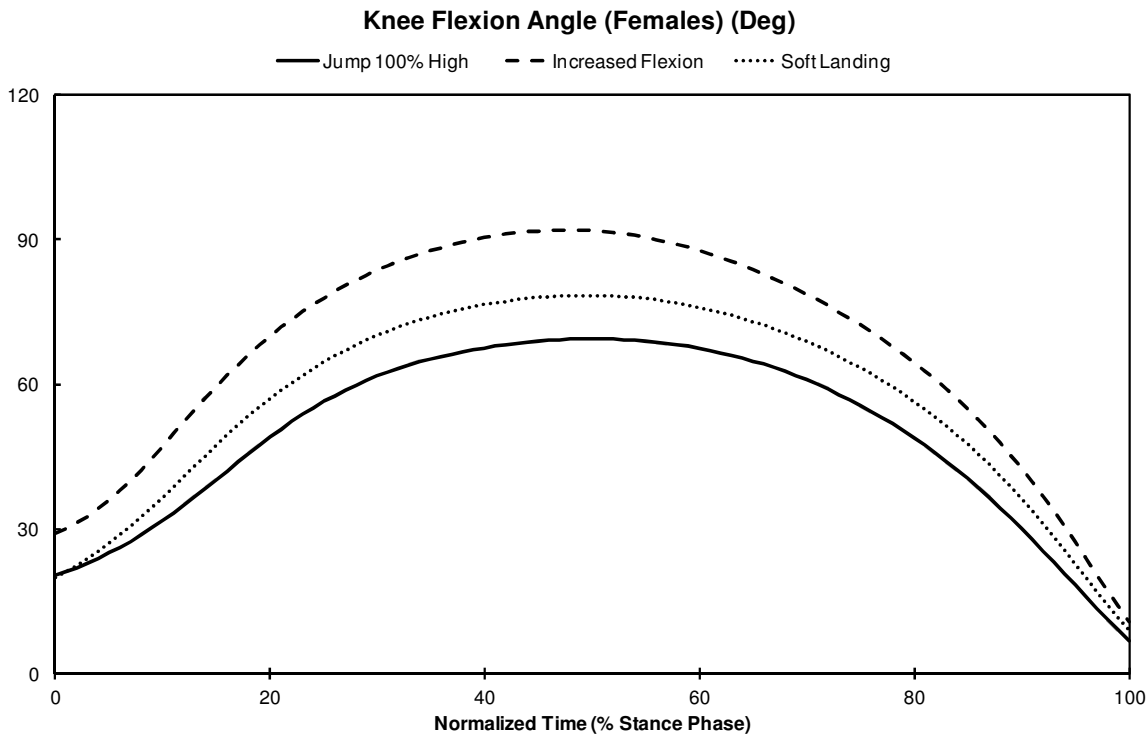

FIGURE B.42. Knee flexion angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

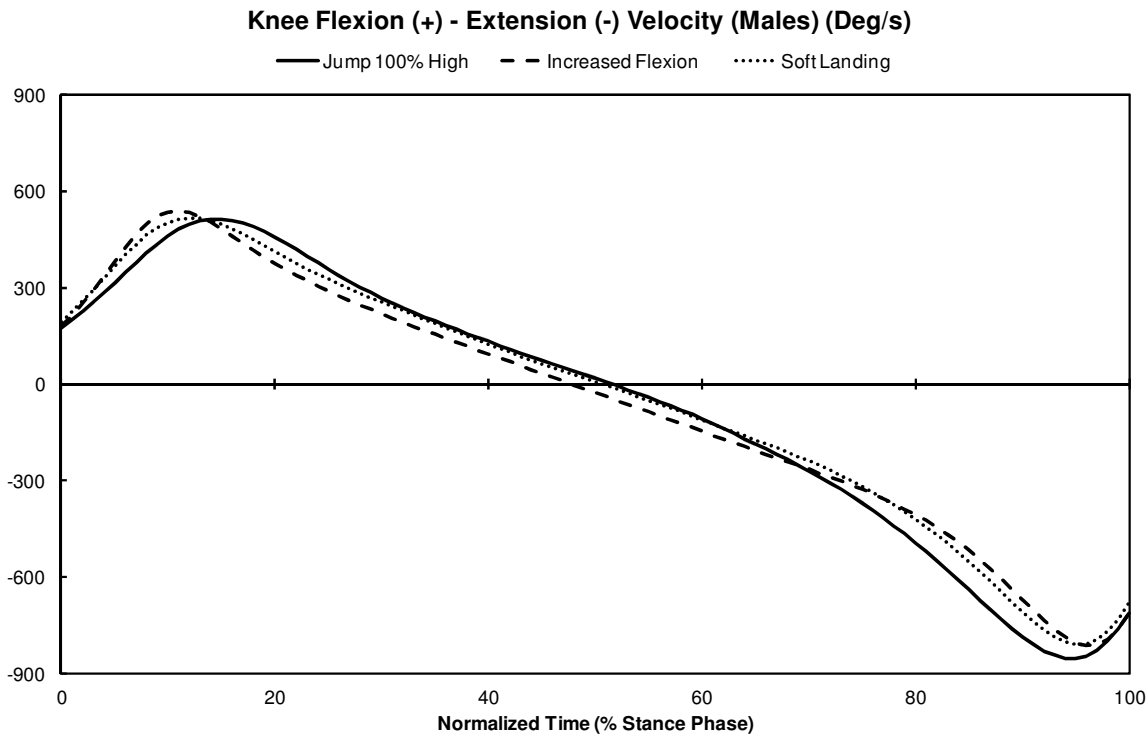

FIGURE B.43. Knee flexion (+) - extension (-) velocity during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

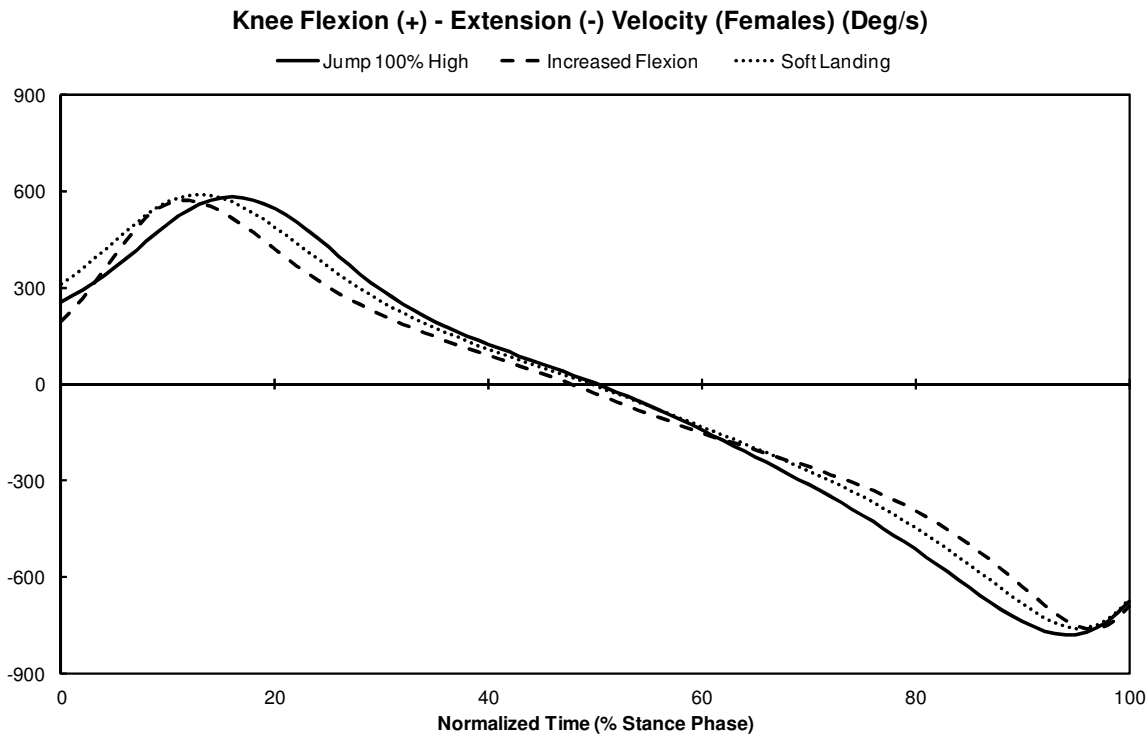

FIGURE B.44. Knee flexion (+) - extension (-) velocity during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

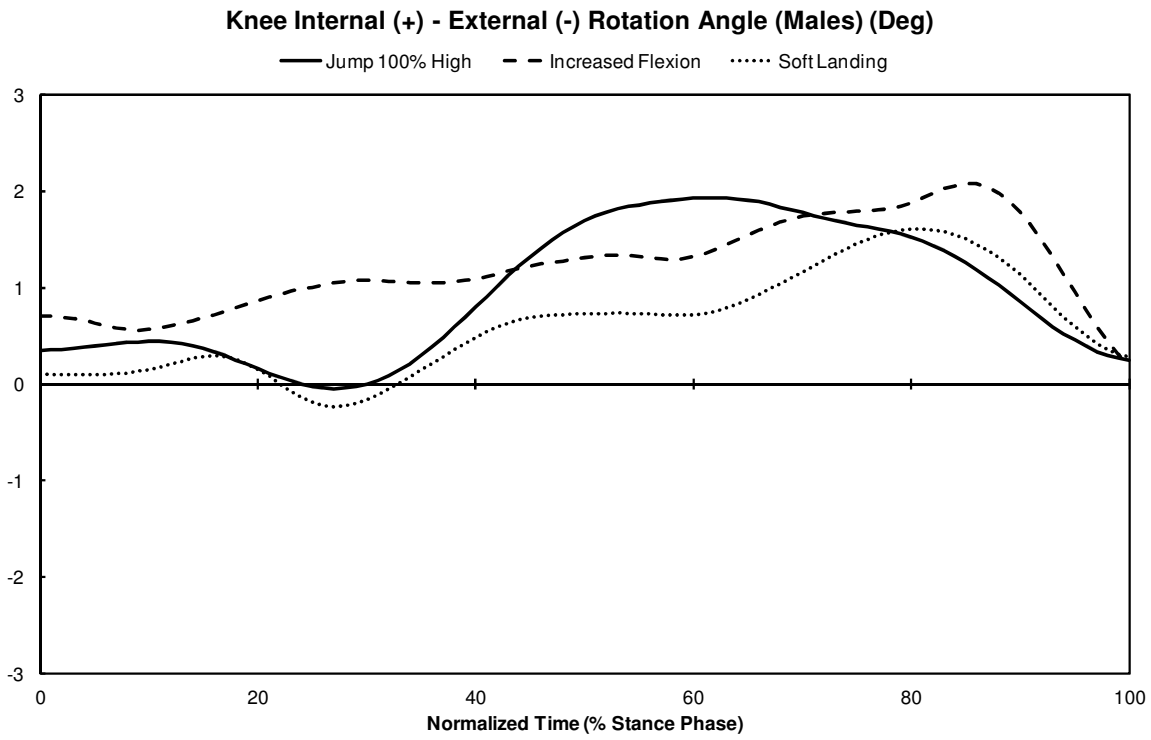

FIGURE B.45. Knee internal (+) - external (-) rotation angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

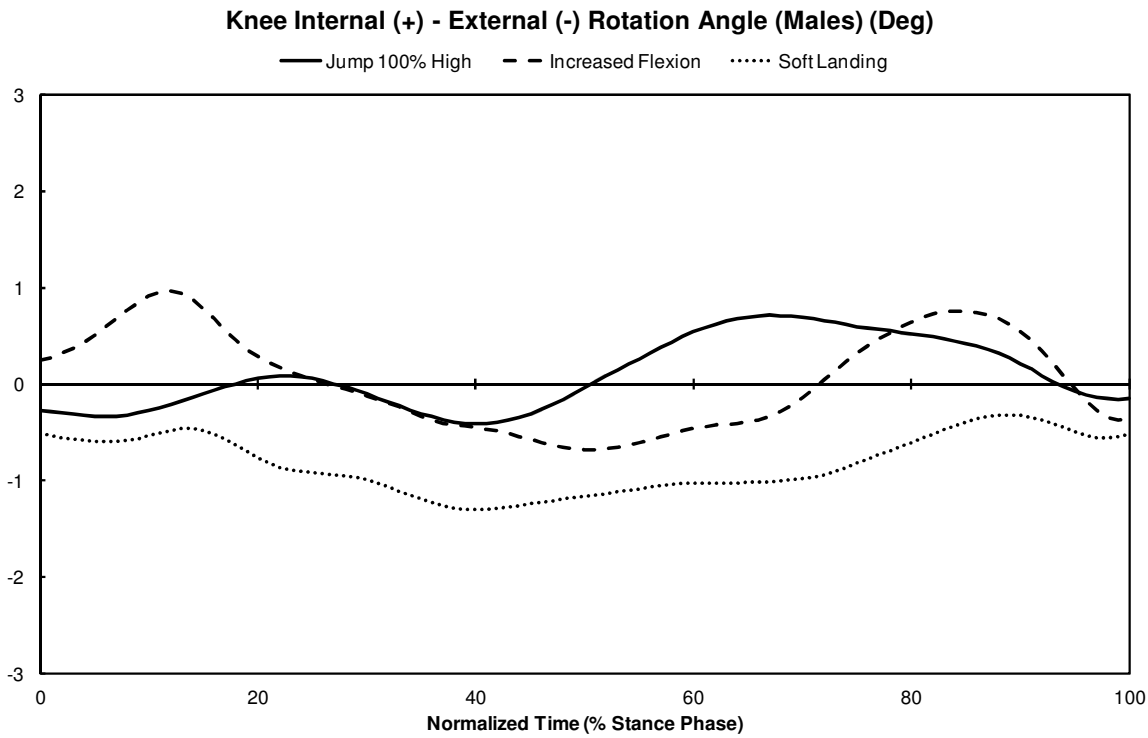

FIGURE B.46. Knee internal (+) - external (-) rotation angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

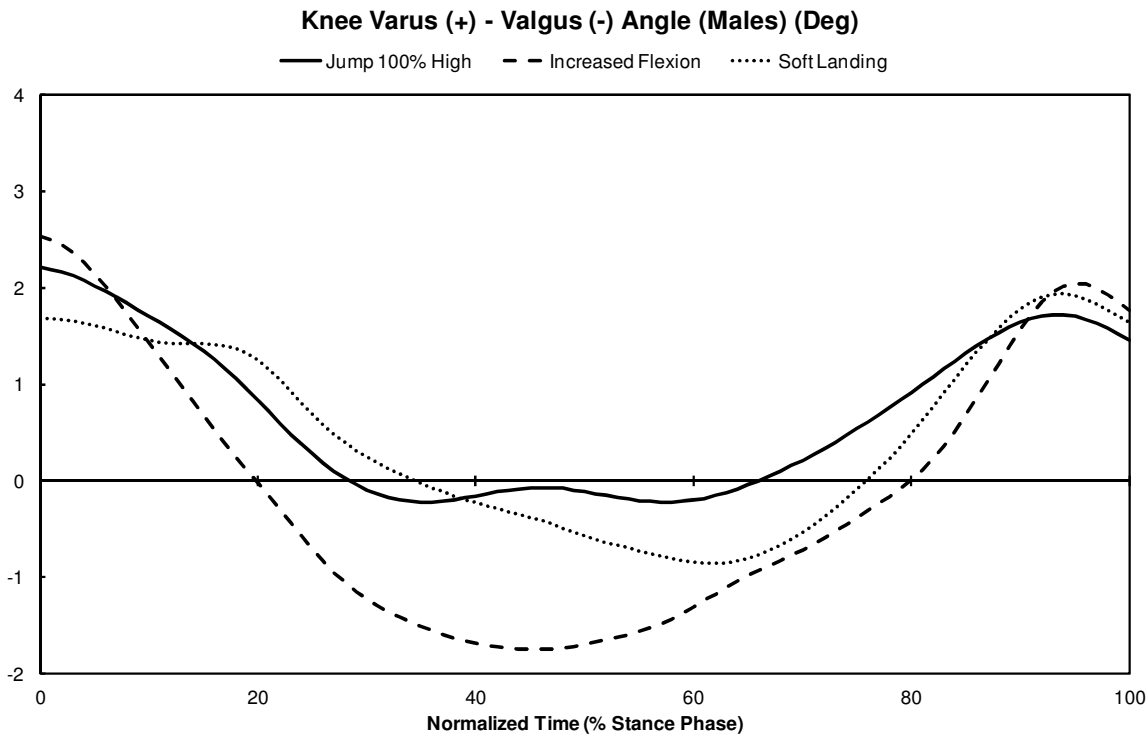

FIGURE B.47. Knee varus (+) - valgus (-) angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

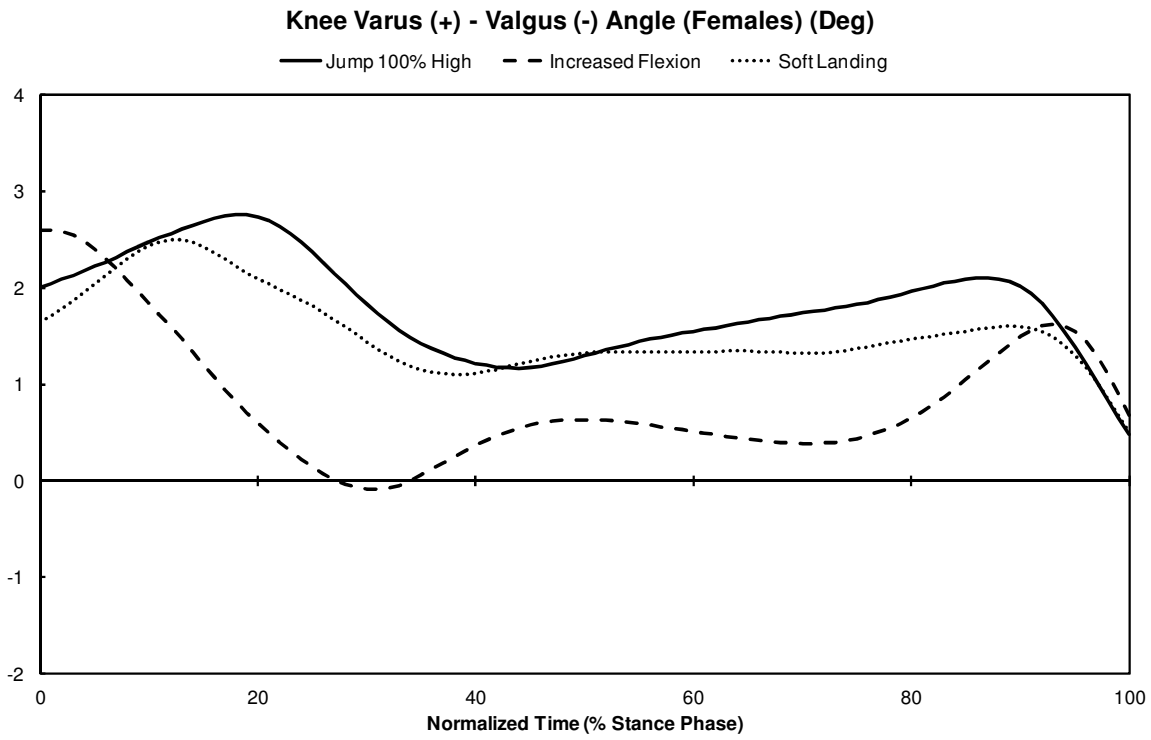

FIGURE B.48. Knee varus (+) - valgus (-) angle during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

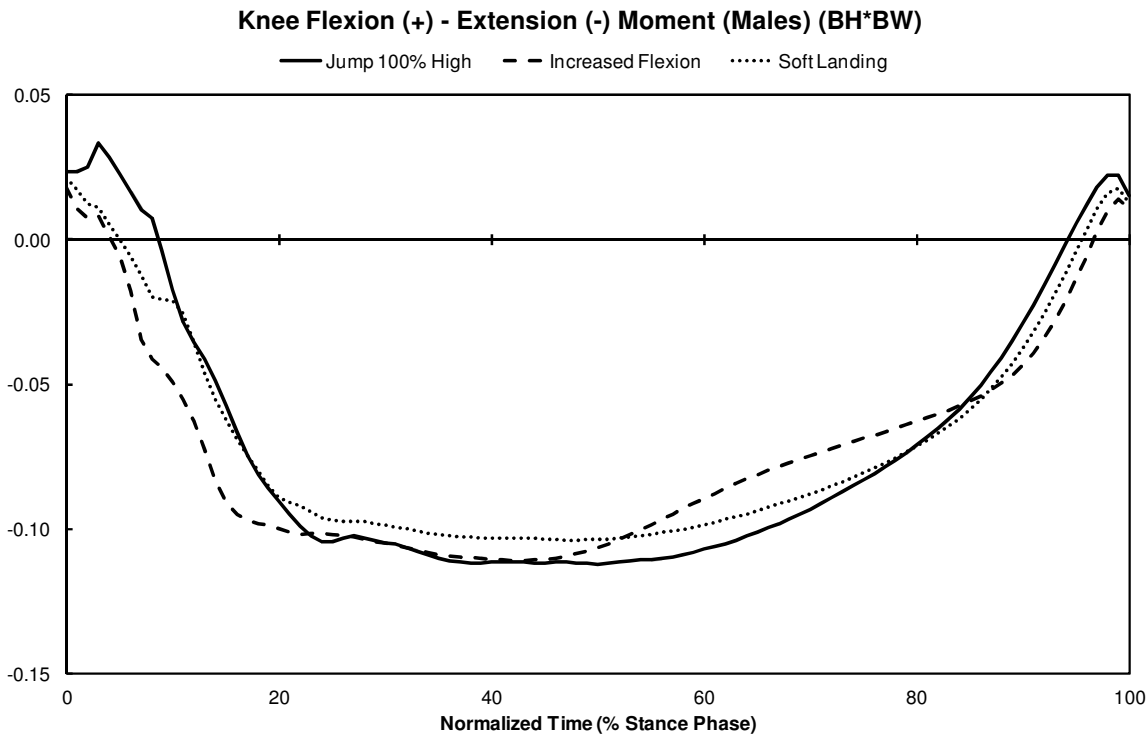

FIGURE B.49. Knee flexion (+) - extension (-) moment during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

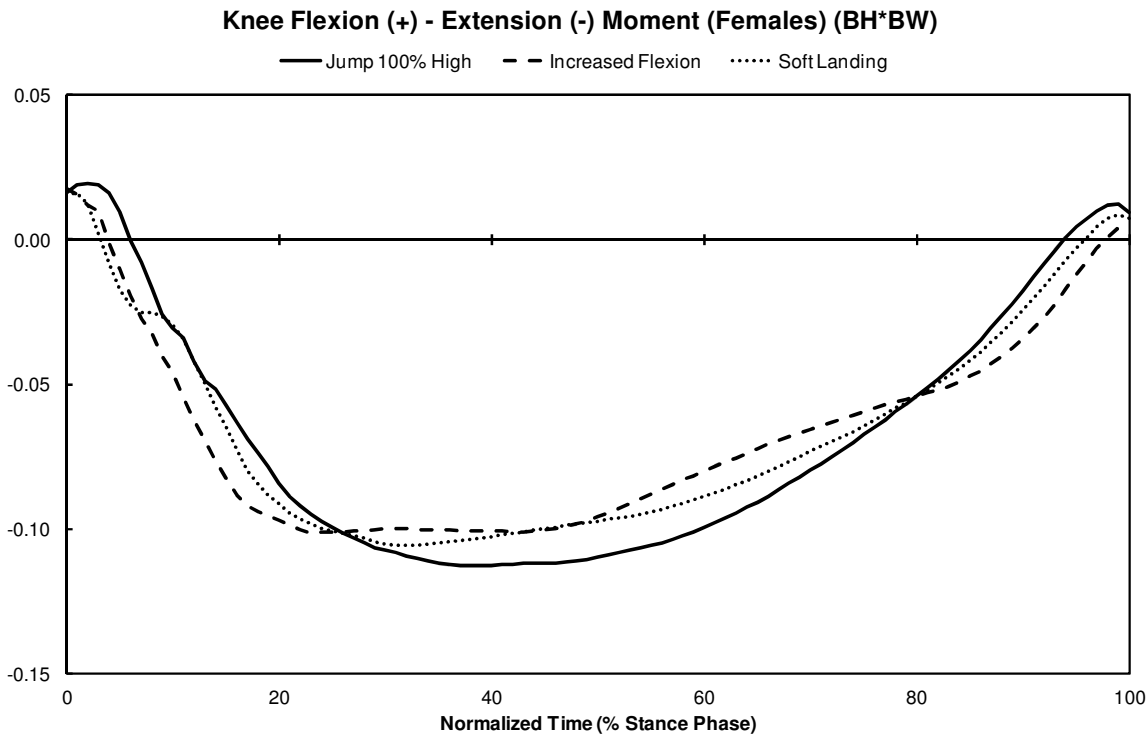

FIGURE B.50. Knee flexion (+) - extension (-) moment during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

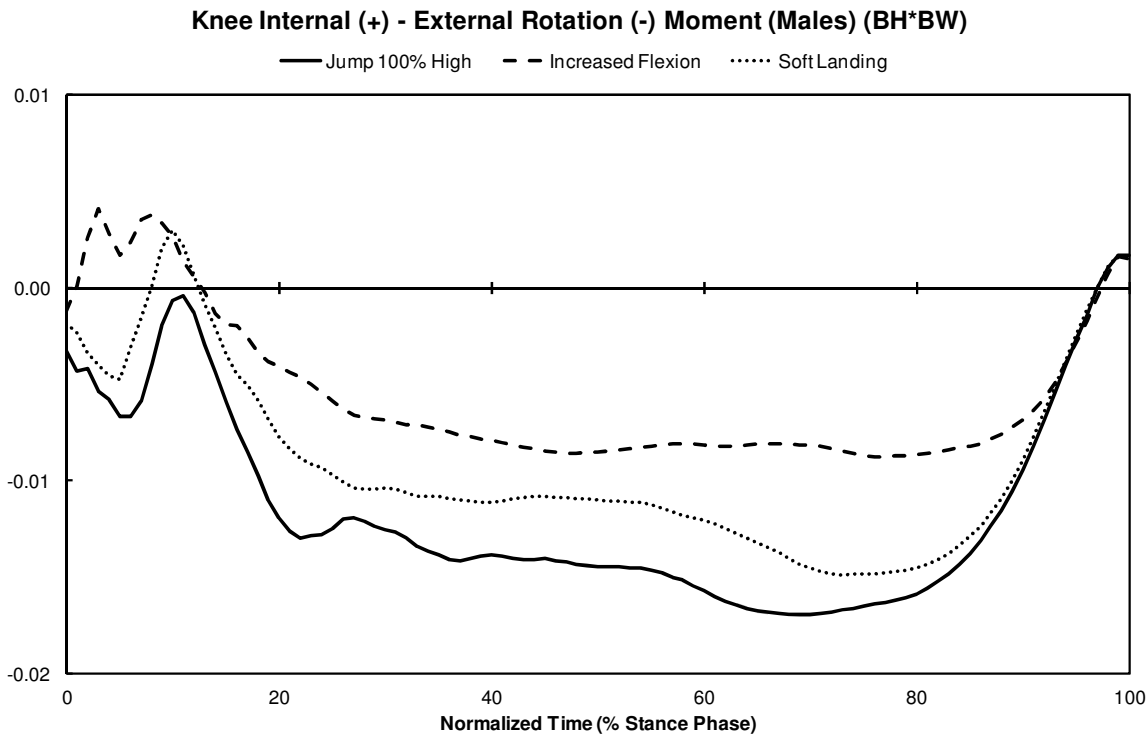

FIGURE B.51. Knee internal (+) - external (-) rotation moment during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

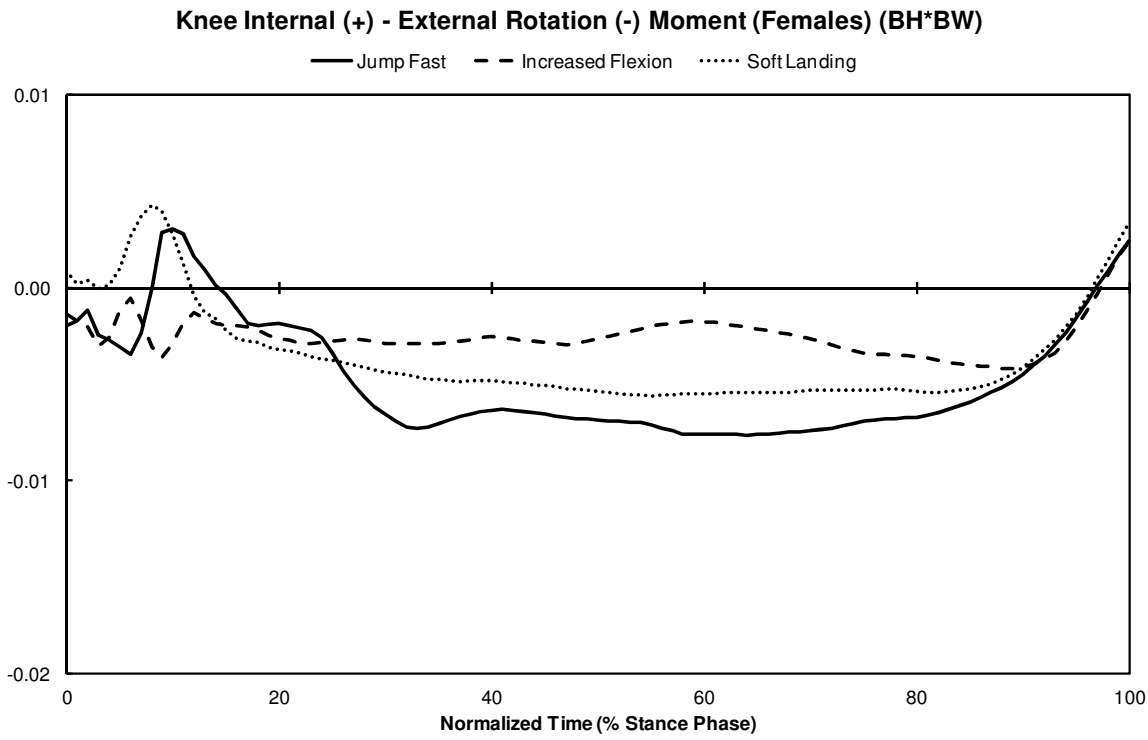

FIGURE B.52. Knee internal (+) - external (-) rotation moment during jumping for maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

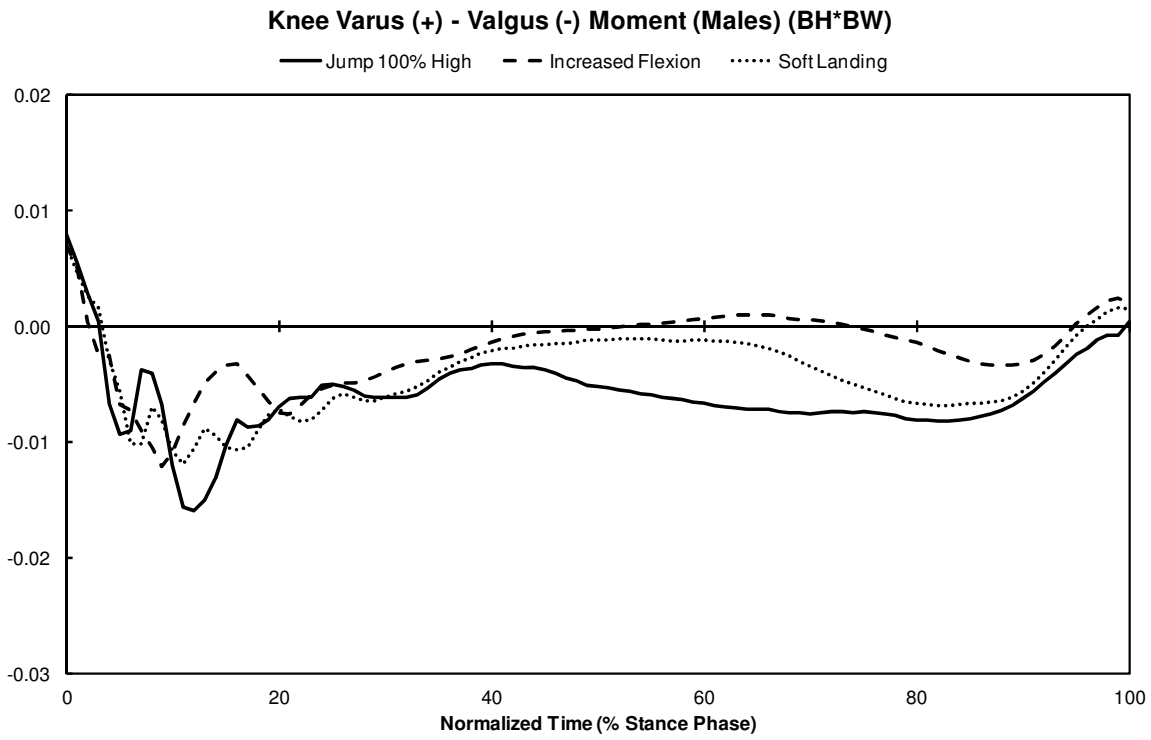

FIGURE B.53. Knee varus (+) - valgus (-) moment during jumping form maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for males

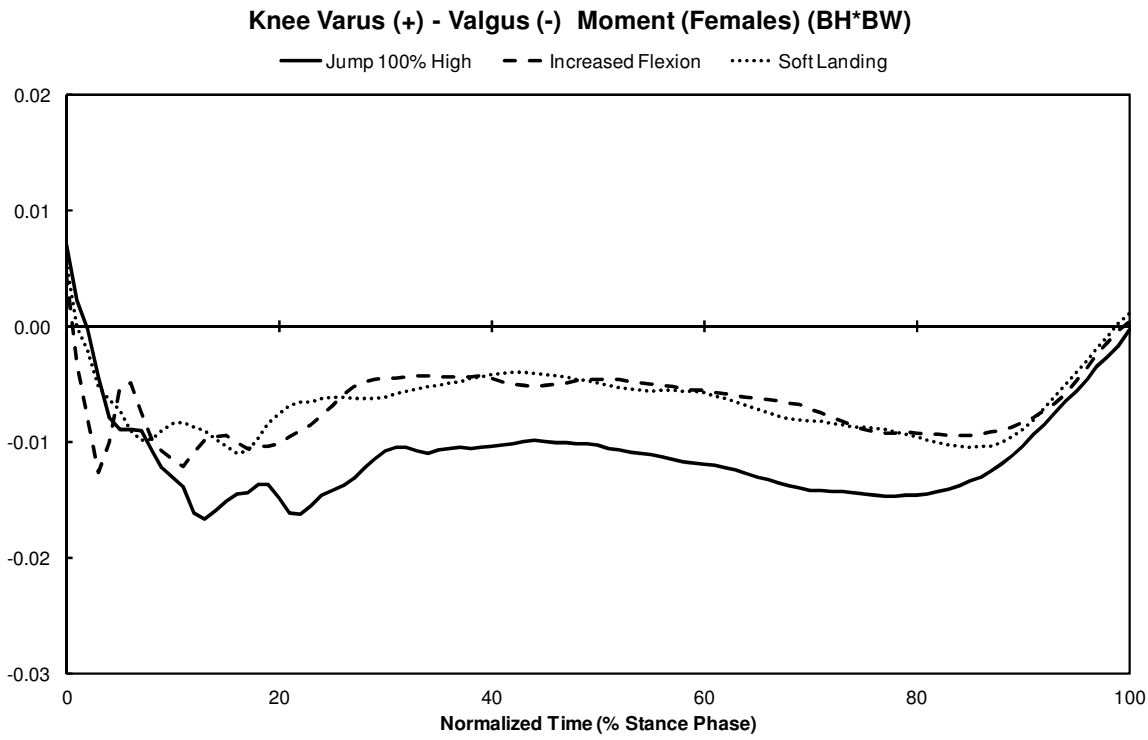

FIGURE B.54. Knee varus (+) - valgus (-) moment during jumping form maximum height, jumping with increased initial knee flexion landing, and jumping with soft landing conditions for females

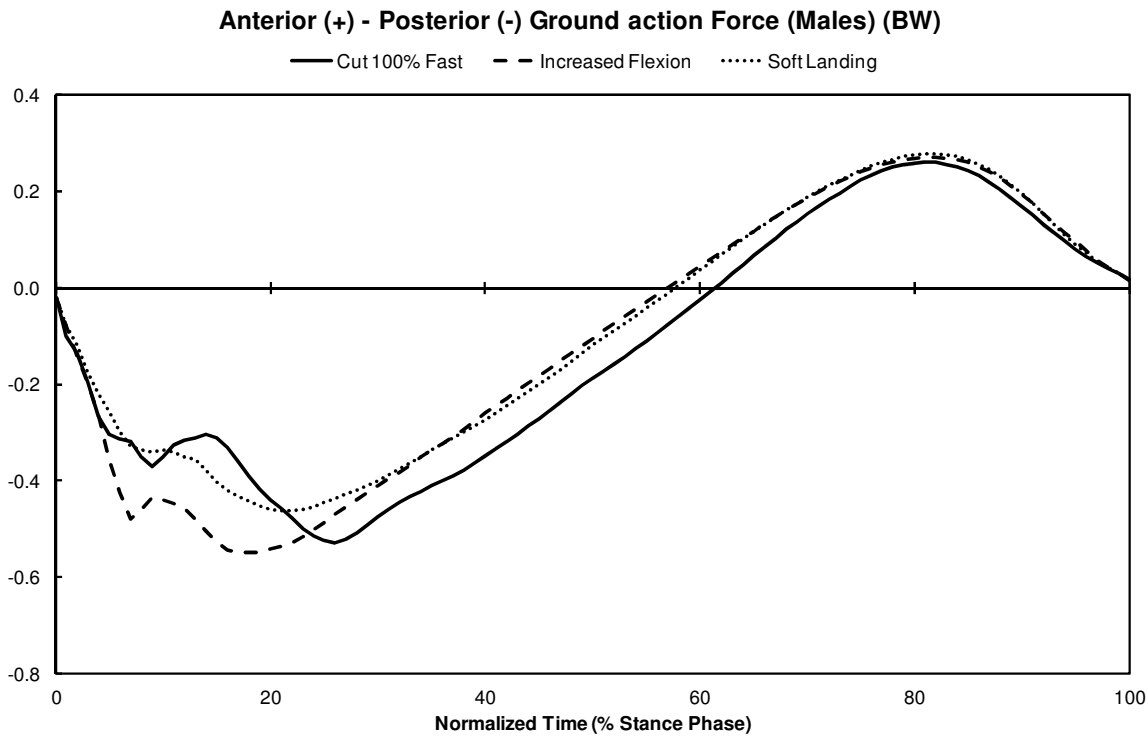

FIGURE B.55. Anterior (+) - posterior (-) ground reaction force during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

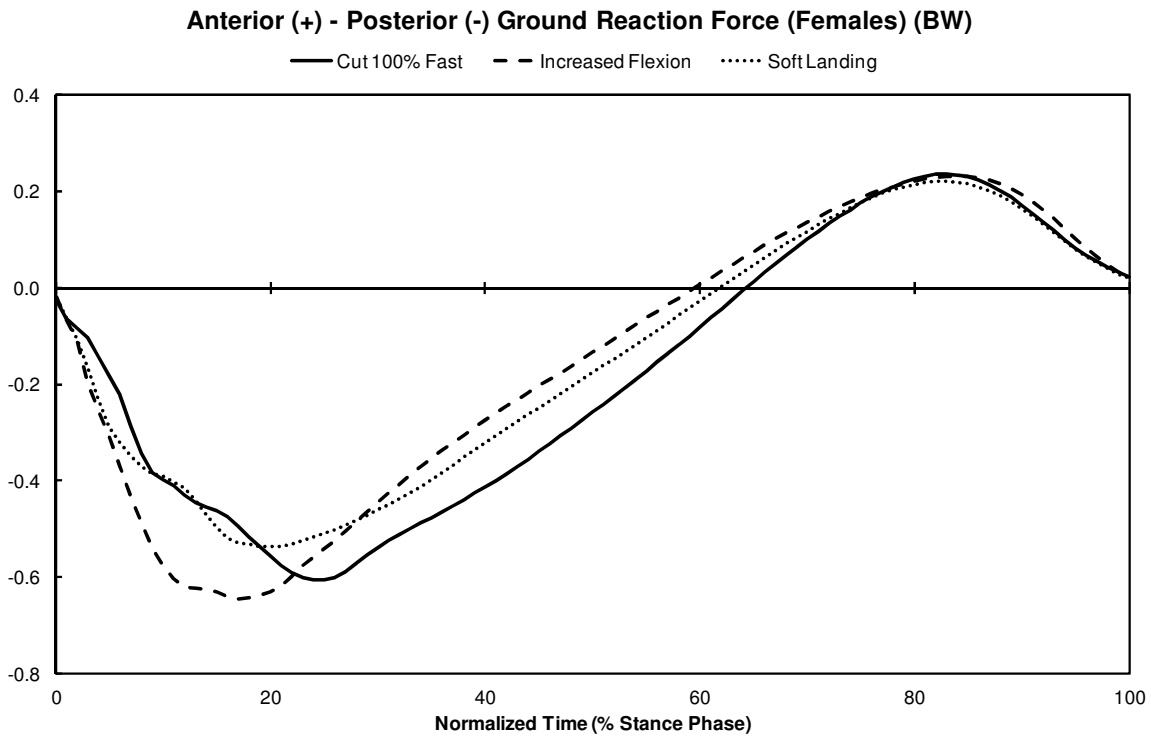

FIGURE B.56. Anterior (+) - posterior (-) ground reaction force during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

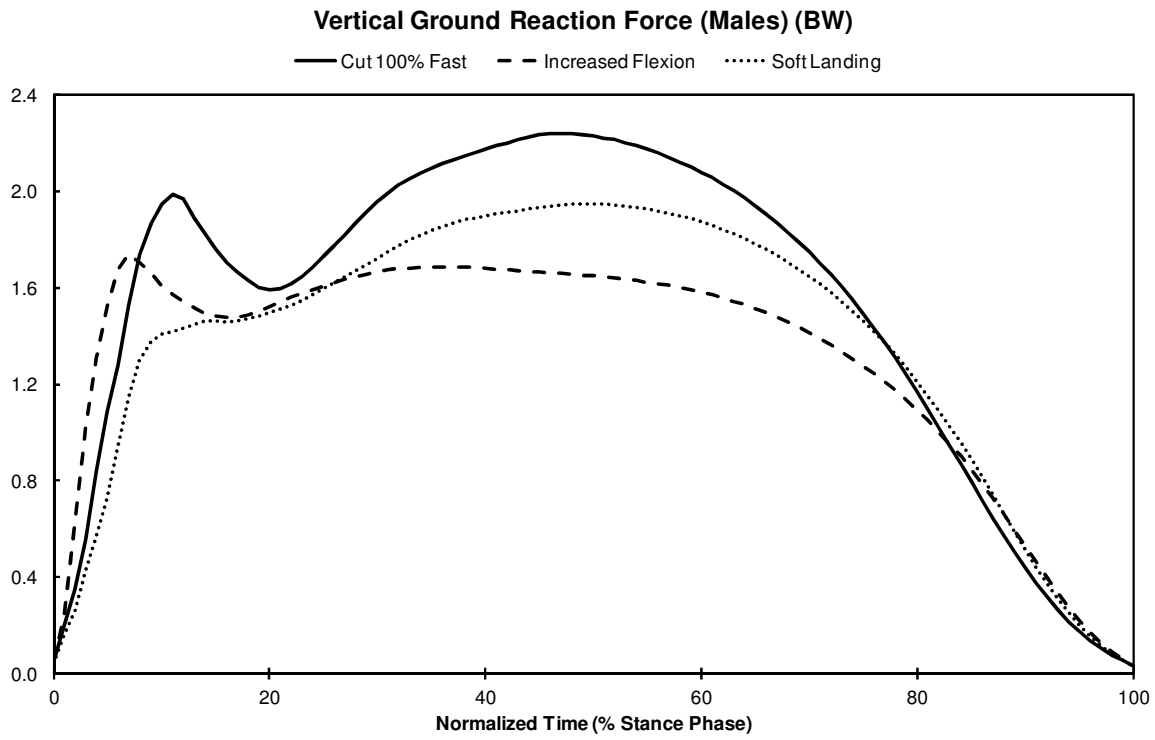

FIGURE B.57. Vertical ground reaction force during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

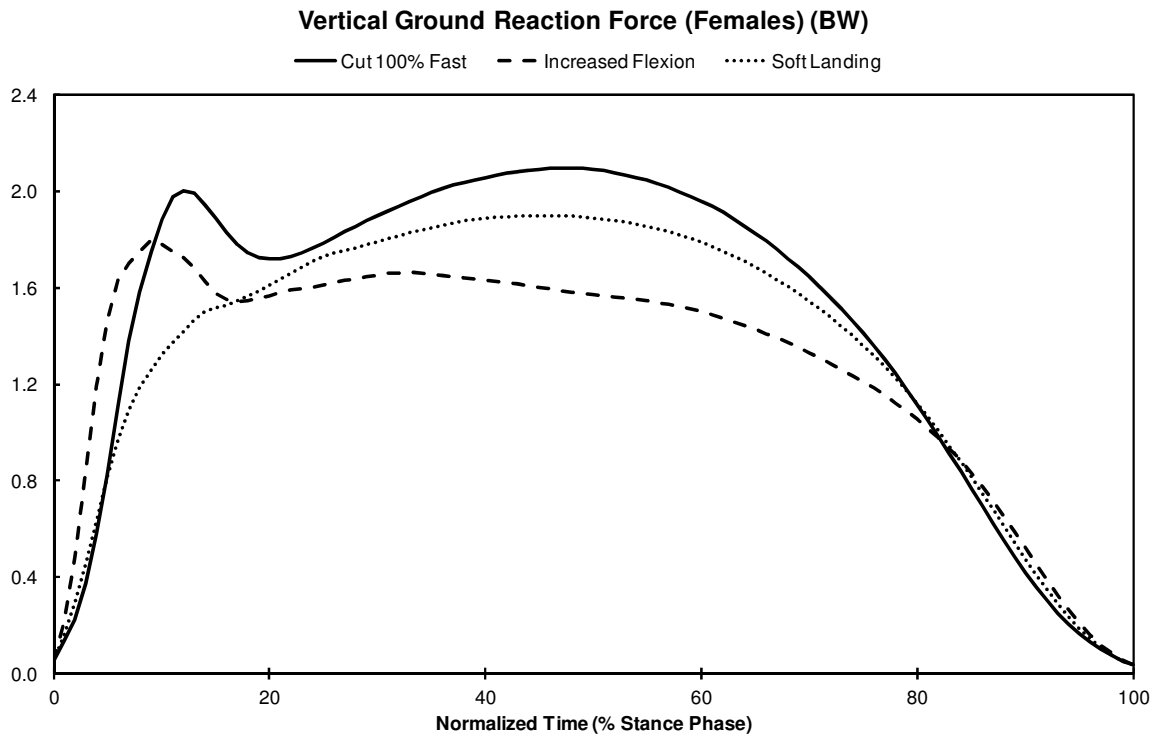

FIGURE B.58. Vertical ground reaction force during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

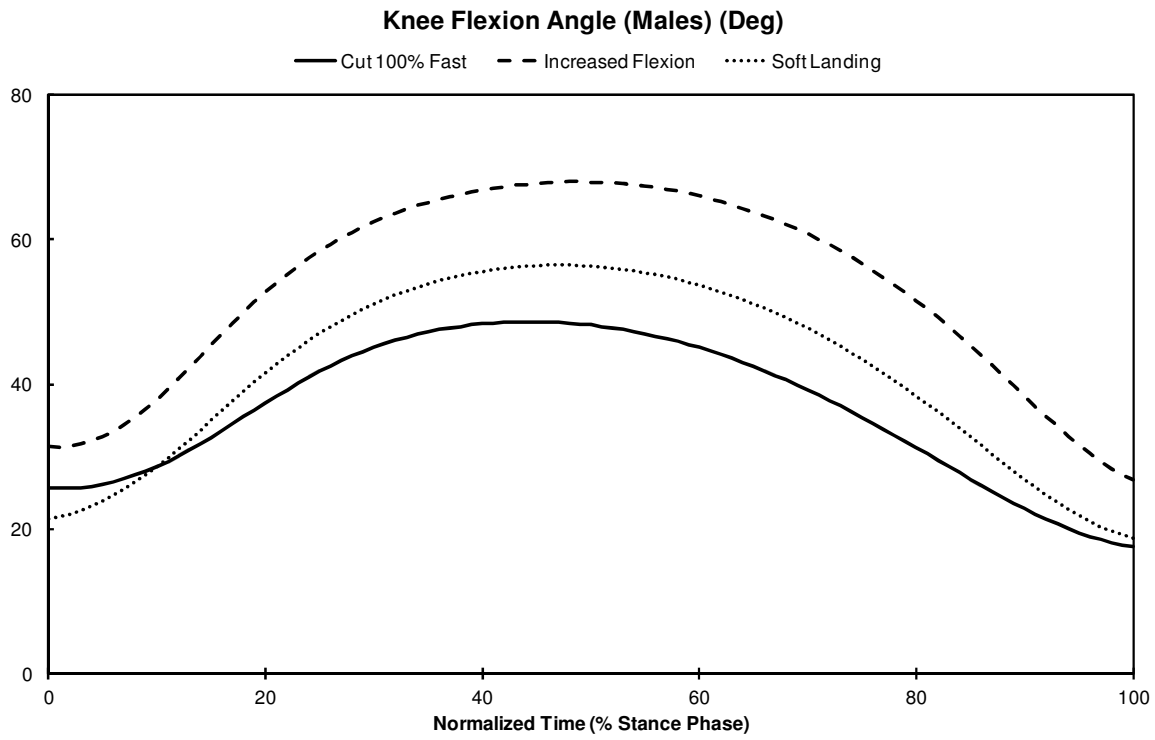

FIGURE B.59. Knee flexion angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

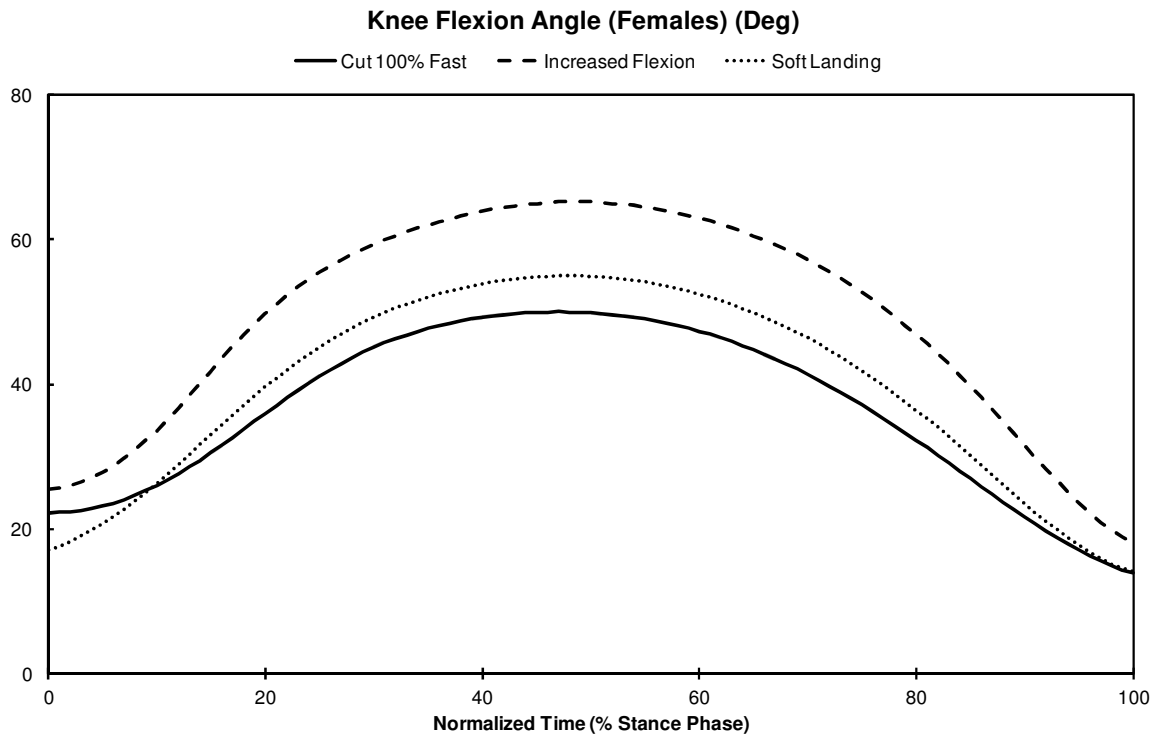

FIGURE B.60. Knee flexion angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

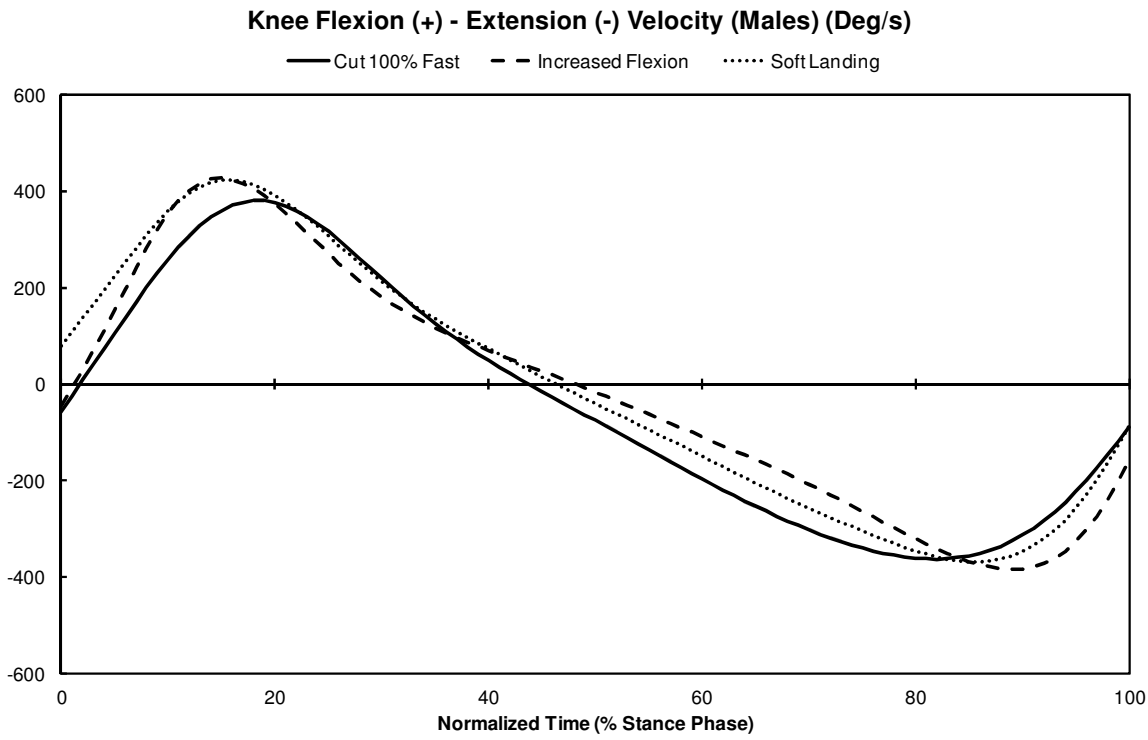

FIGURE B.61. Knee flexion (+) - extension (-) velocity during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

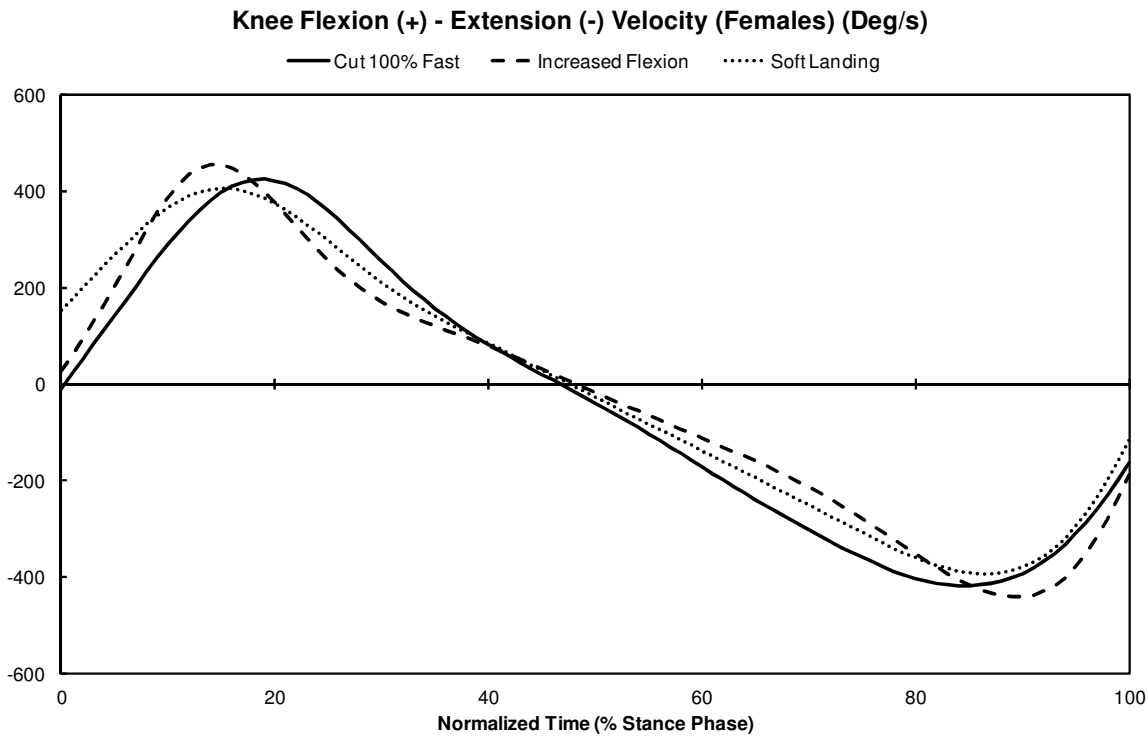

FIGURE B.62. Knee flexion (+) - extension (-) velocity during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

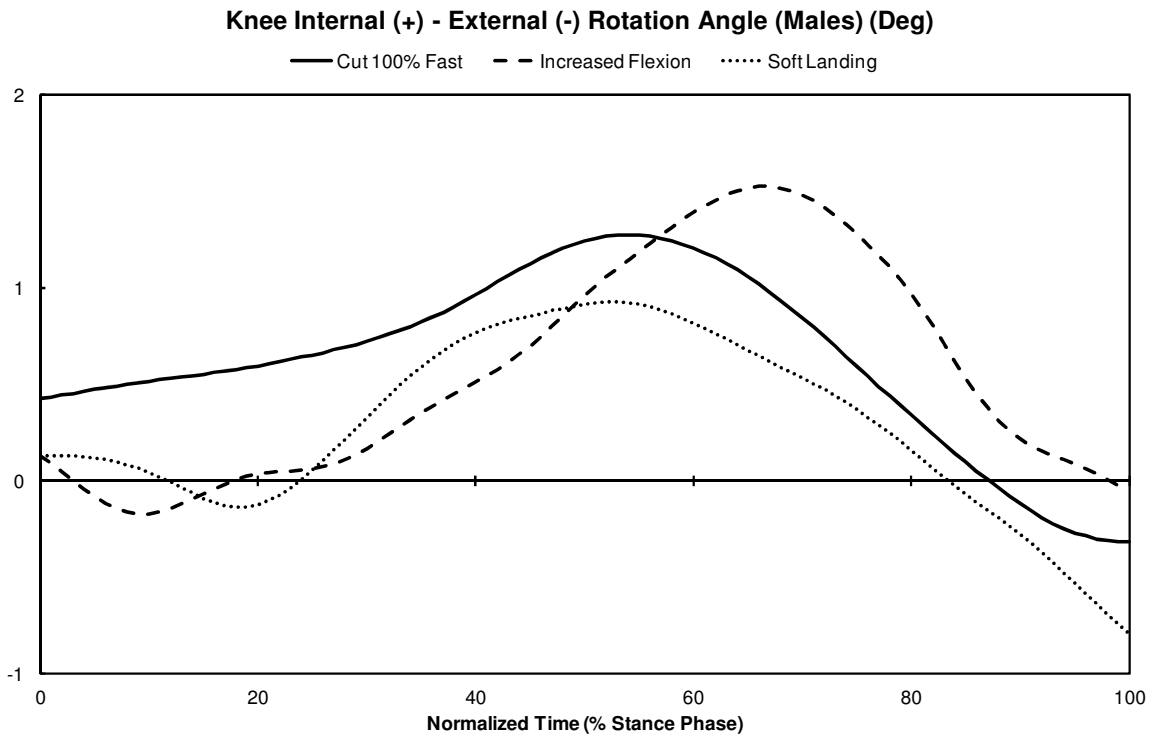

FIGURE B.63. Knee internal (+) - external (-) rotation angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

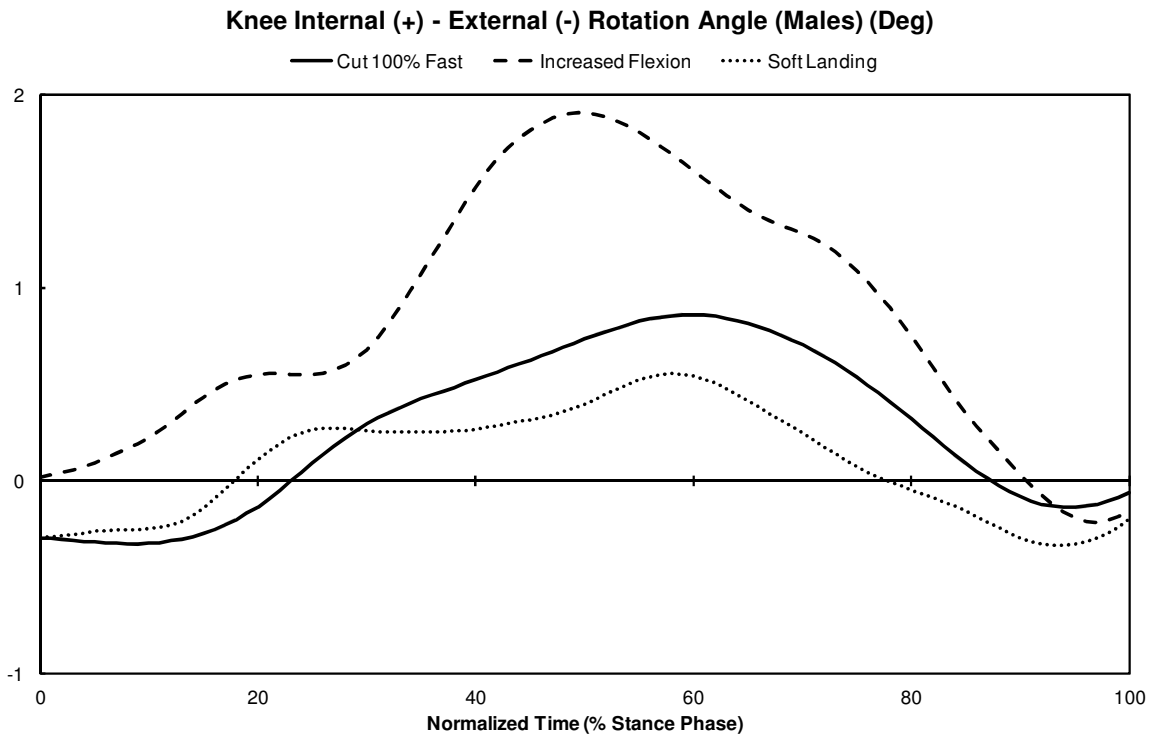

FIGURE B.64. Knee internal (+) - external (-) rotation angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

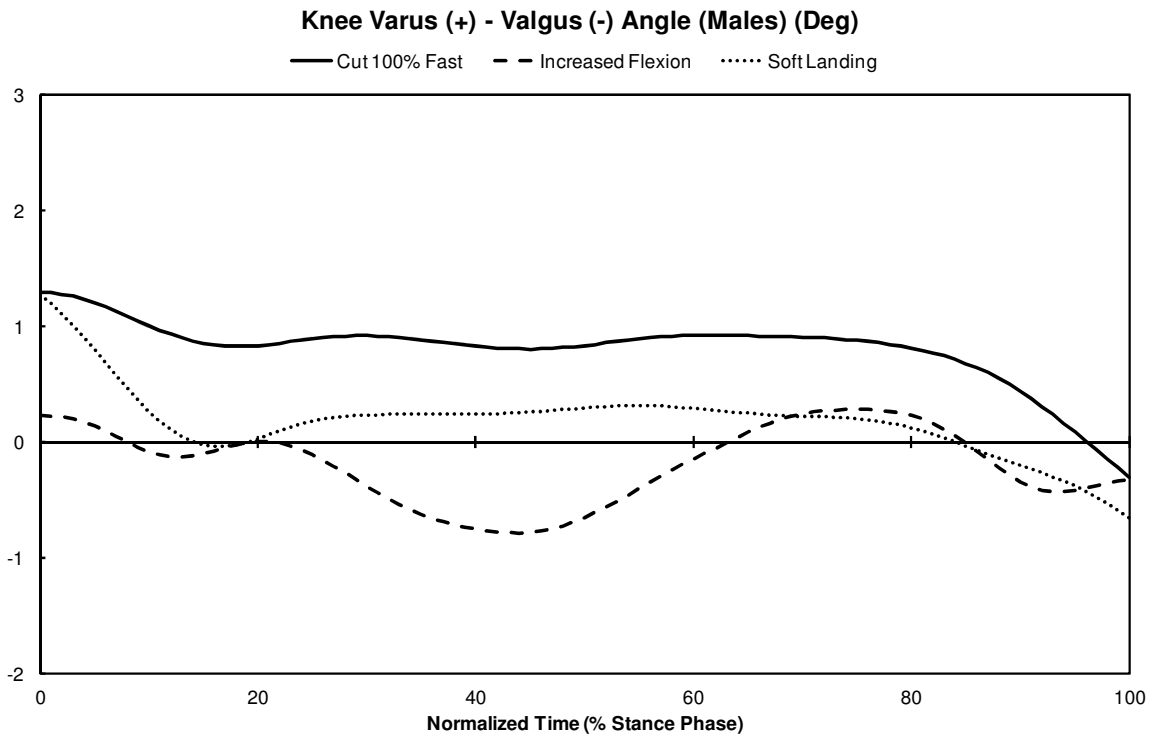

FIGURE B.65. Knee varus (+) - valgus (-) angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

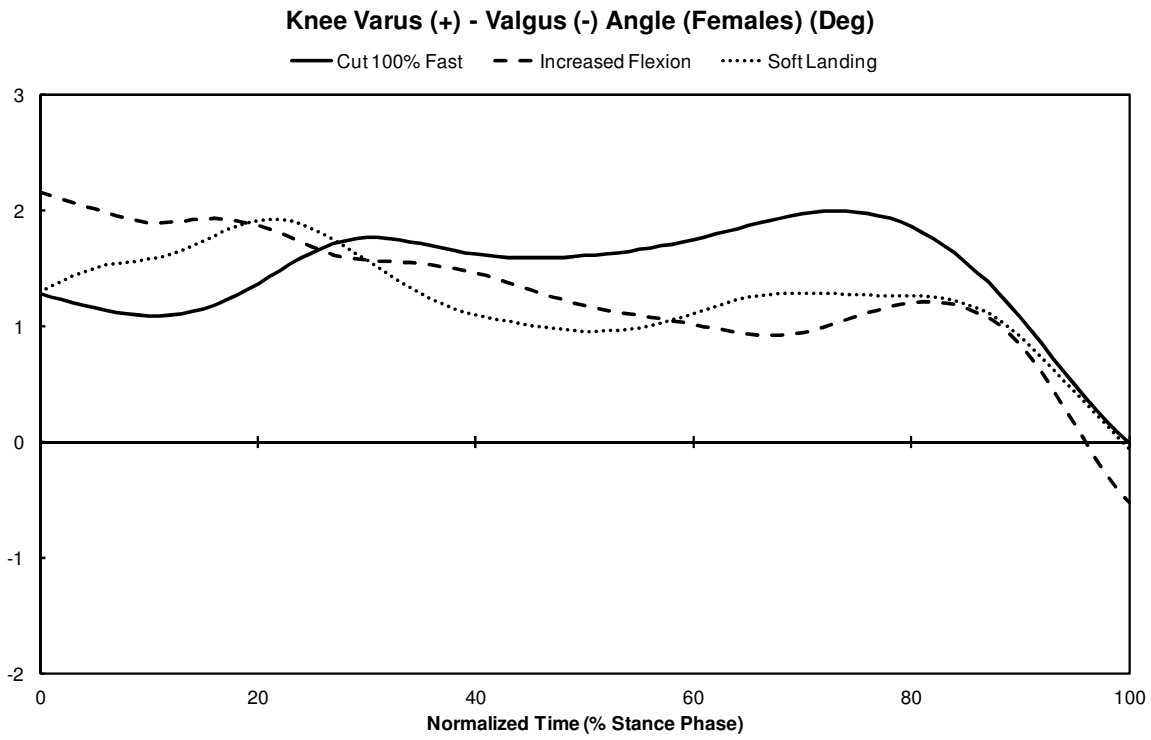

FIGURE B.66. Knee varus (+) - valgus (-) angle during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

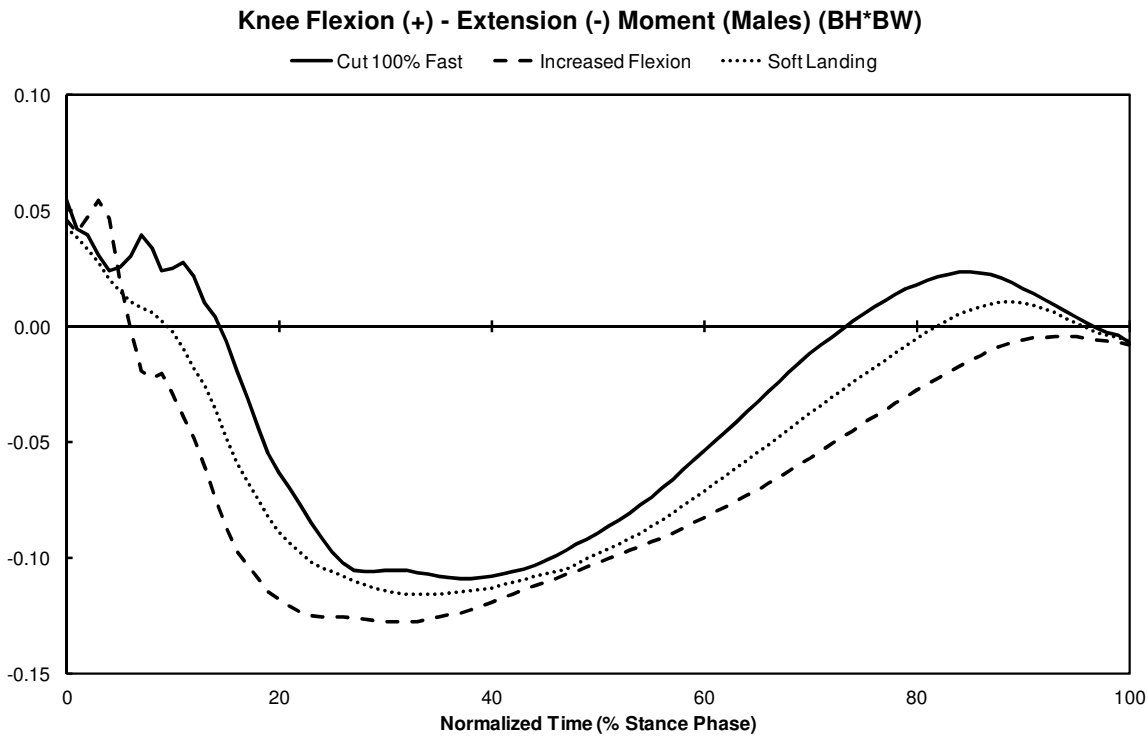

FIGURE B.67. Knee flexion (+) - extension (-) moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

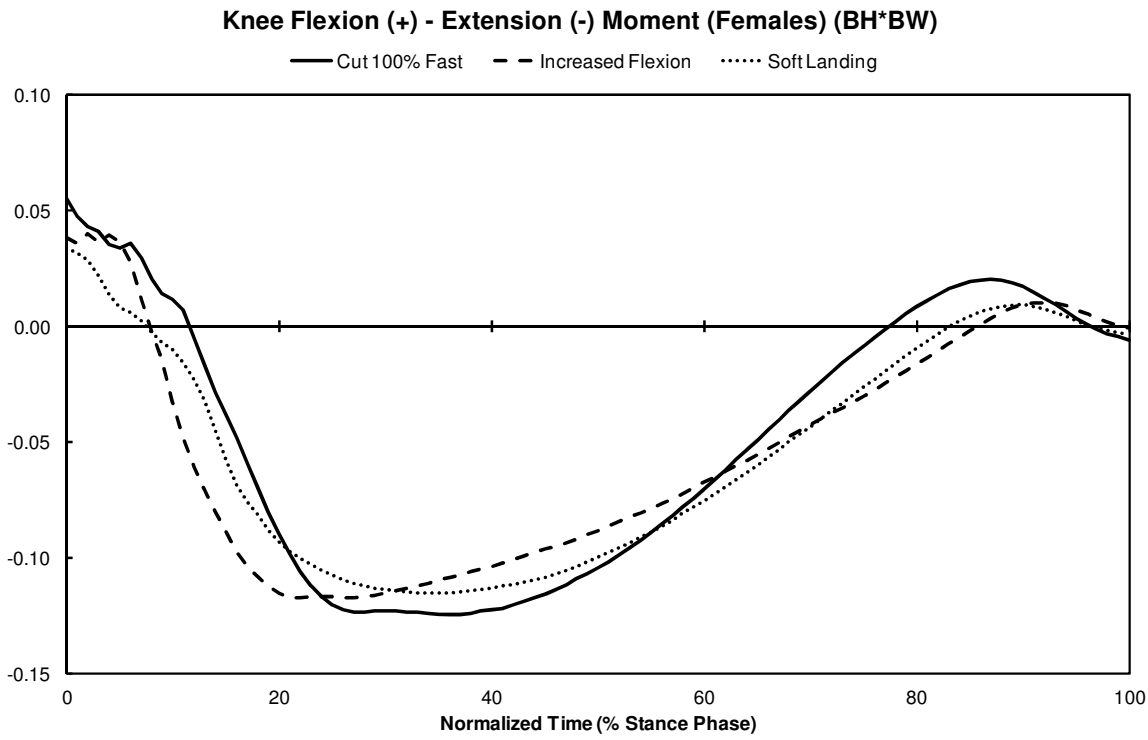

FIGURE B.68. Knee flexion (+) - extension (-) moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

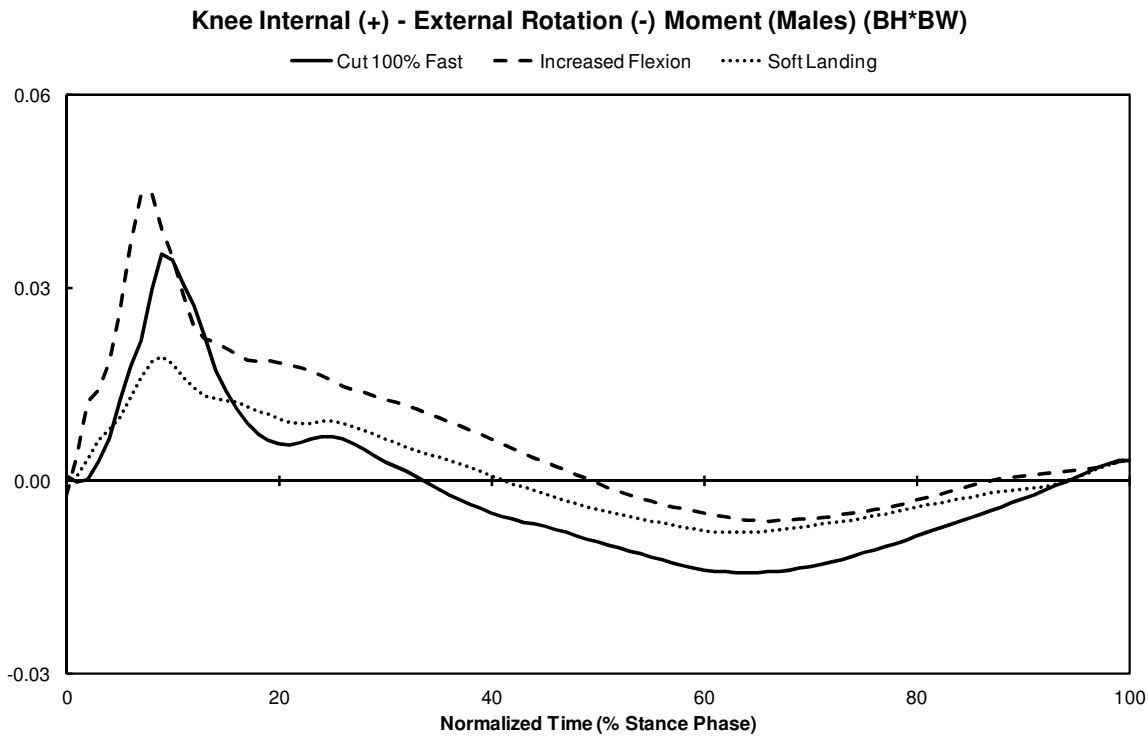

FIGURE B.69. Knee internal (+) - external (-) rotation moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

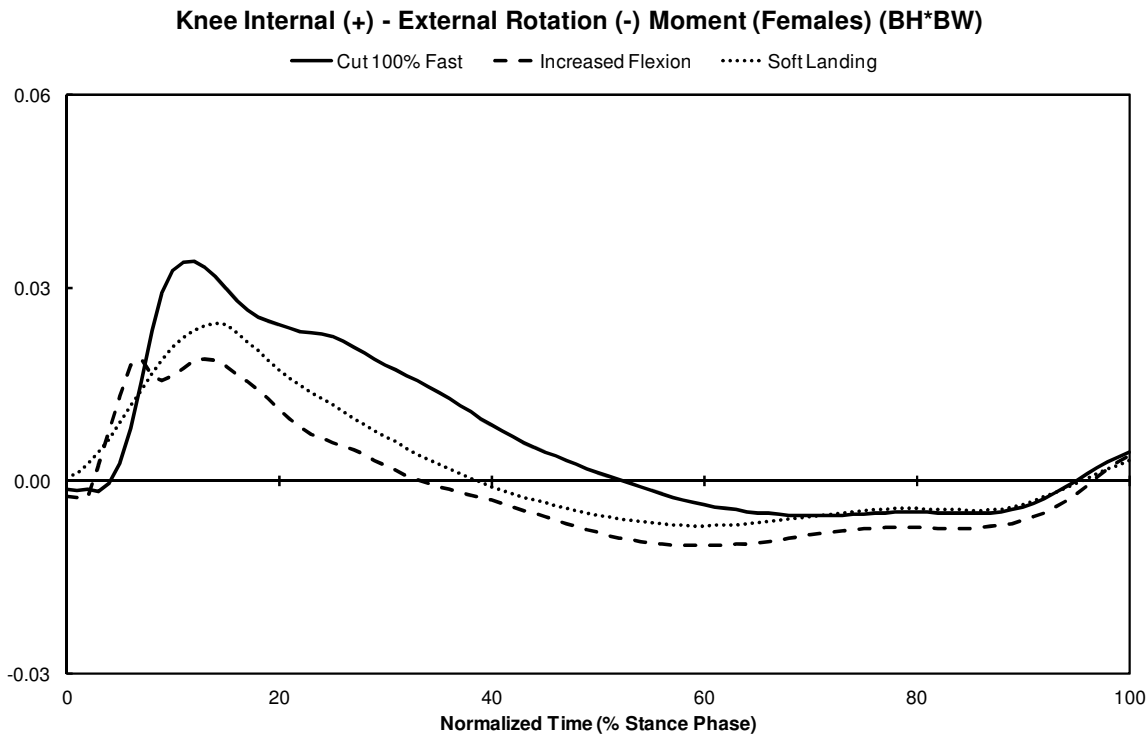

FIGURE B.70. Knee internal (+) - external (-) rotation moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

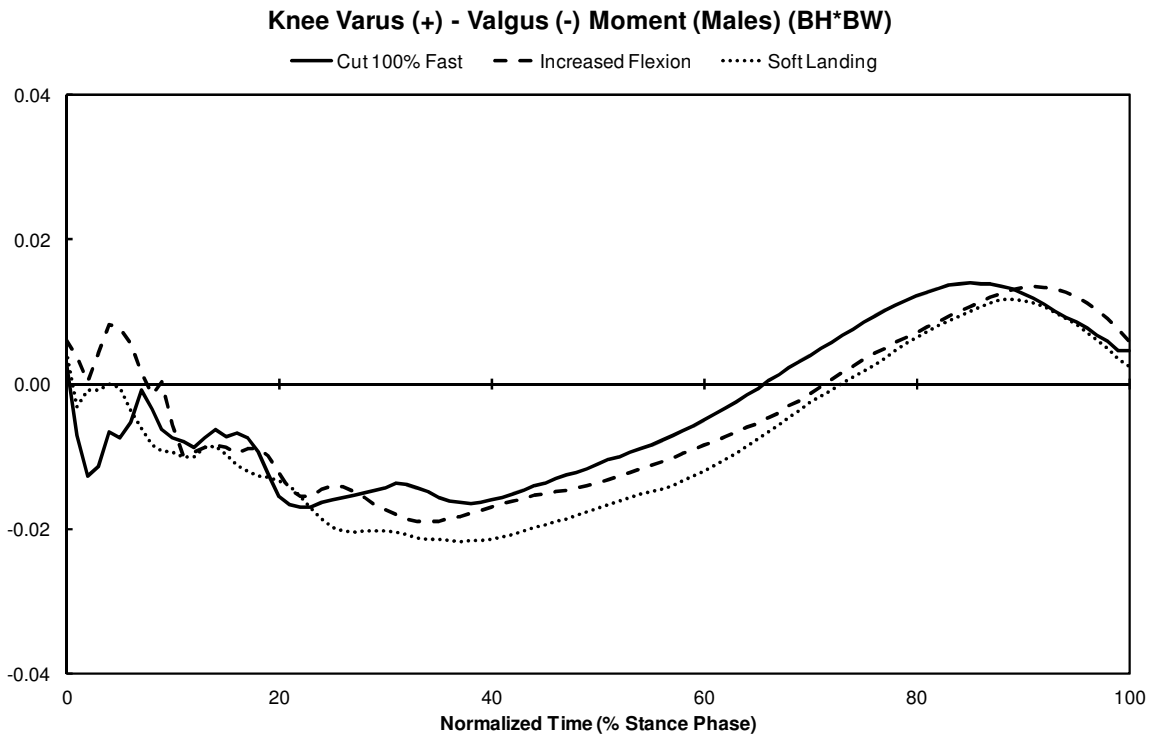

FIGURE B.71. Knee varus (+) - valgus (-) moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for males

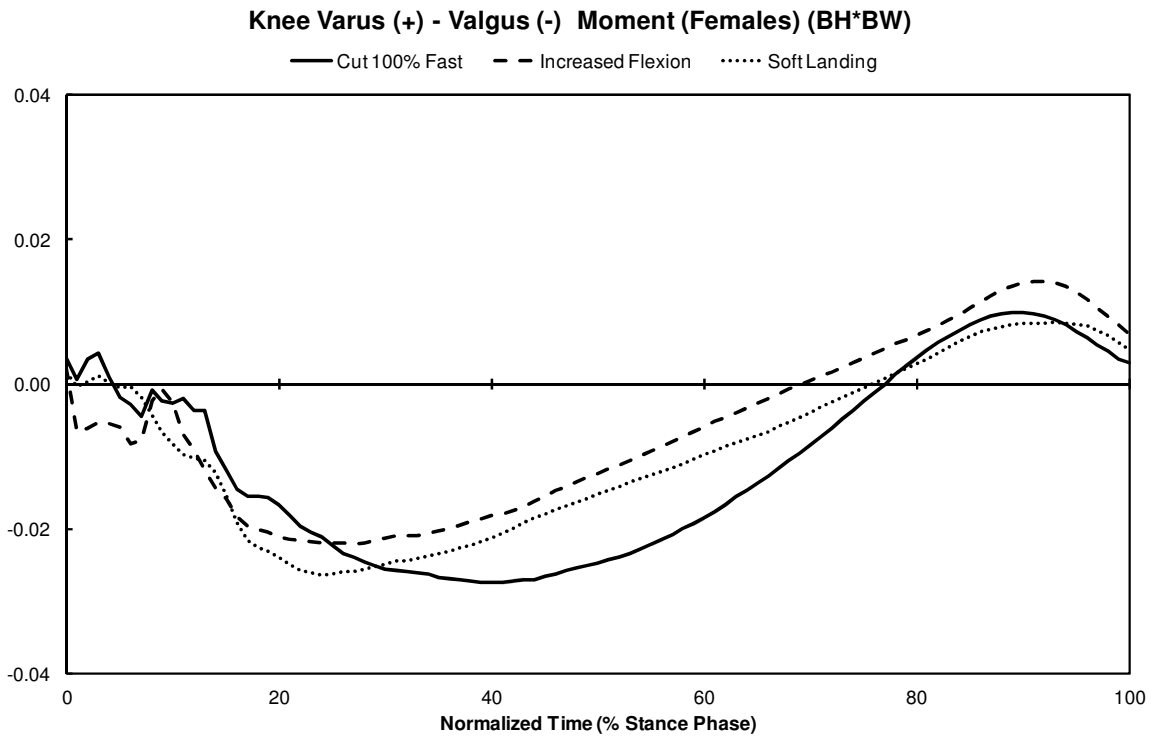

FIGURE B.72. Knee varus (+) - valgus (-) moment during cutting with maximum speed, cutting with increased initial knee flexion, and cutting with soft landing conditions for females

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