PREDICTION OF KNEE KINEMATICS DURING A STOP JUMP-CUT MANEUVER USING TRUNK NEUROMUSCULAR CHARACTERISTICS AND KINEMATICS IN A HEALTHY, PHYSICALLY ACTIVE POPULATION

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Non-contact anterior cruciate ligament injuries (ACL) persist in athletic populations despite years of research to mitigate such injuries. Core stability is purported to be essential in injury prevention, including ACL injury, by healthcare and fitness professionals; however, there is little research investigating this relationship. Examining the relationship between neuromuscular characteristics of core stability and knee kinematics identified to predict or be related to ACL injury may provide insight. The purpose of this study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex could predict knee kinematics during a stop jump-cut maneuver (SJCM).

Fifty three healthy, physically active college-aged subjects participated (age: 22.0±2.1yrs; height: 172.5±8.4cm; weight: 71.6±10.4kg). Testing order for all subjects was: forward/lateral flexion trunk active joint position sense (AJPS), kinematic assessment during a SJCM, and isokinetic trunk extension/rotation strength. Dependent variables were knee valgus and flexion angles at initial contact (IC), total knee valgus excursion, and maximum knee flexion angle. Independent variables were: trunk extension and rotation average peak torque; trunk flexion and lateral flexion AJPS; trunk lateral displacement and trunk flexion angle at initial contact and maximum; and sex. Backwards stepwise linear regression was performed for each of the dependent variables with their respective, selected independent variables.

None of the independent variables were significant predictors for knee valgus angle at IC or maximum knee flexion angle. Trunk rotation strength towards the direction of the cut and sex were found to be significant predictors of total knee valgus excursion (R^2=0.259, p=0.001), with lower trunk rotation strength and female sex predicting greater total knee valgus excursion. Trunk extension strength and sex were found to be significant predictors of knee flexion at IC (R^2=0.282, p<0.001), with lower trunk extension strength and female sex predicting lower knee flexion at IC.

The results indicate that deficits in trunk strength and female sex induce risky knee kinematics that may be associated with ACL injury risk. Future research should investigate if deficits in trunk strength predict non-contact ACL injury and determine if targeted programs to increase trunk extension and rotation strength decrease non-contact ACL injury risk.

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PREFACE

I would like to pause for a moment to thank the many people who have made the completion of this research project and degree possible. I would like to thank Dr. Scott Lephart for the opportunity to work in the Neuromuscular Research Laboratory while pursuing my degree at the University of Pittsburgh. I would like to extend a special thank you to my dissertation chair, Dr. Timothy Sell, for his patience and guidance through this process. To Dr. John Abt, thank you for always being a source of calm in the storm. Thank you to Dr. Mita Lovalekar for your statistical guidance on this project and unique perspective over the years. Thank you to Dr. David Stone, whose sharp wit and humor always keeps me on my toes. To all of the remaining NMRL faculty and staff, thank you for making this journey a great experience.

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1.0 INTRODUCTION

Despite years of research investigating risk factors of and prevention strategies for anterior cruciate ligament (ACL) injury, these injuries continue to persist in a young athletic population. Although accounting for a relatively small percent of athletic injuries overall, ACL injury poses a significant burden on both the healthcare system and the individual. Core stability is advocated by healthcare and fitness professionals to be essential for injury prevention; however, the relationship between core stability and injury, including ACL injury, is not clearly understood. One way to better understand how core stability may be related to ACL injury is to determine how neuromuscular characteristics responsible for maintaining functional stability of the trunk are related to knee kinematics. The purpose of the current study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex could predict knee kinematics during a stop jump-cut maneuver (SJCM). Knee kinematics included knee valgus and knee flexion angles at initial contact, total knee valgus excursion, and maximum knee flexion angle. It was hypothesized that greater knee valgus angle at initial contact would be predicted by lower trunk extension muscular strength, higher (worse) trunk lateral flexion proprioception towards the dominant leg, greater lateral trunk displacement in the frontal plane at initial contact, and female sex. It was hypothesized that greater total knee valgus excursion would be predicted by lower trunk rotation muscular strength towards the cutting direction, higher (worse) trunk lateral flexion proprioception towards the dominant leg, greater maximum lateral trunk displacement, and female sex. Further, greater knee flexion angle at initial contact would be predicted by greater trunk extension muscular strength, lower (better) trunk flexion proprioception, greater trunk flexion angle at initial contact, and male sex. It also was hypothesized that greater maximum knee flexion angle would be predicted by greater trunk extension muscular strength, lower (better) trunk flexion proprioception, greater maximum trunk flexion angle, and male sex. Identifying which trunk neuromuscular and kinematic characteristics predict knee kinematics will enable selection of appropriate core exercises with the intent of reducing the risk of noncontact ACL injury.

1.1 ANTERIOR CRUCIATE LIGAMENT INJURY

More than half of all sports-related injuries are lower extremity injuries, with injury to the ankle and knee occurring most frequently.[1,2] Although male and female athletes have comparable overall injury rates,[2] female athletes continue to be at greater risk of sustaining an ACL injury than their male counterparts.[2-4] Further, the majority of these injuries are noncontact,[4-7] indicating that intervention strategies may be able to target modifiable risk factors and decrease the risk of ACL injury. Despite years of research and the development of intervention strategies, a statistically significant increase in the rate of ACL injury has occurred at the collegiate level, with an average annual increase of 1.3% over a 16 year period.[1] One possible explanation is that previous research has, for the most part, ignored the potential role of the trunk or core stability in noncontact ACL injury. It is believed that the ability to control the trunk is crucial during dynamic movement, as this provides the foundation for the distal segments to function effectively.[8-12] Further, from an injury prevention perspective, the position of the trunk relative to its base of support may be more noticeable, making it easier for coaches, athletic trainers, and athletes to identify and correct improper or "risky" positioning.

1.1.1 Consequences of Anterior Cruciate Ligament Injury

Although ACL injuries account for less than 3% of all injuries,[1-3,6] ACL injury still poses a considerable burden. Immediate consequences include significant time loss for athletes,[1,7,13-15] loss of potential scholarship or salary, and alterations in emotion and behavior,[16-18] potentially affecting physical recovery[19-22] as well as academic performance.[23] Although ACL reconstruction is relatively cost-effective in comparison to other medical treatments,[24,25] there are long-term consequences associated with ACL injury. While individuals who have undergone ACL reconstruction demonstrate significant short-term[26,27] and long-term[27,28] improvement in knee function survey scores, only 33-50% these individuals are able to return to sports.[20,28,29] In addition, while knee-related function is maintained, physical activity levels are significantly reduced.[26] Individuals who have sustained traumatic knee injuries are at increased risk of developing osteoarthritis (OA) later in life.[30,31] More specifically, 48-82% of athletes who had sustained ACL injury demonstrated radiographic evidence of osseous changes of the tibiofemoral joint at long-term follow-up, with 15-51% diagnosed with OA.[29,32-34] While some authors indicate that the development of OA is significantly greater in individuals who have undergone ACL reconstruction versus conservative treatment following ACL injury,[29,32,33] others have reported comparable rates.[34]

1.1.2 Noncontact Anterior Cruciate Ligament Injury

Based on the consensus statement from the Hunt Valley II meeting, risk factors for noncontact ACL injury can be classified as environmental, anatomical, hormonal, familial tendency, or neuromuscular.[35] Among these categories, the neuromuscular risk factors may be the most amenable to preventative intervention strategies. Neuromuscular risk factors can be further classified as those related to altered movement patterns, altered muscle activation patterns, and insufficient muscle stiffness.[35] While controlled laboratory studies have helped to lay the theoretical foundation for neuromuscular risk factors of noncontact ACL injury, there are a limited number of prospective studies that have been conducted to determine the association between these risk factors and risk of noncontact ACL injury.

Prospectively, it has been demonstrated that female athletes who sustained ACL injury possessed different lower extremity kinematics during a drop vertical jump landing than those who did not sustain ACL injury.[36] More specifically, those who went on to ACL injury demonstrated significantly greater knee abduction at initial contact and at maximum displacement as well as significantly less maximum knee flexion. Kinematic differences also have been identified between athletes who sustained a second ACL injury following ACL reconstruction (ACL-R) and return to sport.[37] Male and female athletes who incurred a second ACL injury demonstrated significantly greater total front plane (valgus) movement during a drop vertical jump landing than those who did not. Further, ACL-R athletes with increased total frontal plane movement were three times as likely to suffer a second ACL injury.

Video analysis of actual ACL injury events has provided further insight into the underlying injury mechanism of noncontact ACL injury.[38-44] At initial contact, the knee is in a relatively extended position (<30°) and in neutral valgus. It is believed that ACL rupture occurs approximately 30 to 40 milliseconds after initial contact and typically is accompanied by a rapid increase in knee valgus. However, in one study, valgus collapse was seen in only 20% of the male subjects during ACL injury (compared to 53% of the females) and mean knee flexion was significantly lower in male subjects.[42] These findings suggest that the mechanism underlying noncontact ACL injury may be different in males and females. It is possible that these differences may be explained by differences in neuromuscular control of the lumbopelvic-hip complex.

From a theoretical perspective, neuromuscular control of the hip can influence that of the knee and, therefore, abnormal hip mechanics can contribute to risk of knee injury. When the foot is fixed and the hip moves into an adducted and internally rotated position, dynamic knee valgus occurs as the knee abducts and foot pronates.[45] Dynamic knee valgus is related to ACL injury[36] as well as patellofemoral pain syndrome.[45,46] This relationship between the hip and knee is supported by both prospective[47] and retrospective[10,48] research that has demonstrated that weakness of the hip musculature is related to knee injury. More recently, the role that neuromuscular control of the trunk may play in knee kinematics[49,50] and in lower extremity injury[10,41,51,52] has been investigated.

1.2 FUNCTIONAL JOINT STABILITY

Functional joint stability is defined as the ability to preserve or revert back to proper positioning or alignment[53] and is modulated through the sensorimotor system. Effective and efficient functioning of the sensorimotor system is reliant on the sensory, motor, and central integration and processing components.[54] Functional joint stability is dependent on the interaction of static and dynamic components.[55-59] The static components of functional joint stability include ligaments, joint capsule, cartilage, bony geometry of the joint, and friction between the joint articulating surfaces whereas dynamic components include the musculotendinous structures that transverse a joint and their associated neural pathways.[56]

Underlying the ability to maintain functional joint stability are feedforward and feedback control.[60-62] Feedforward control is activated in anticipation of a destabilizing event, thereby regulating muscle stiffness in preparation for movement, joint loading, or perturbation.[59] In contrast, feedback control is initiated after sensory detection of a destabilizing event and is affected by previous experiences. Both mechanisms of control rely on somatosensory, visual, and vestibular information; however, feedforward mechanisms are utilized sporadically until feedback mechanisms begin.[61] Since feedback control relies on continuous integration and processing of afferent information, adjustments can be made in response to the changing environment.

One method by which afferent information regarding the changing environment is obtained is through conscious proprioception. Proprioception is a subcomponent of the sensorimotor system and refers to afferent information arising from stimulation of articular, musculotendinous, and cutaneous mechanoreceptors, which influences postural control, functional joint stability, and conscious sensation.[53,63] The modern submodalities of proprioception are kinesthesia, joint position sense, and force sense.[53] Kinesthesia[55,60,64,65] refers to the ability to sense movement, whereas joint position sense[60,64-66] and force sense[58,59] are the ability to identify or replicate limb/body position and force, respectively. Although functional joint stability and its underlying constructs have predominantly been applied to the extremities, these same principles can be applied to the core.

1.3 THE CORE AND CORE STABILITY

1.3.1 Definition of the Core

The core is referred to as the lumbopelvic region[67-70] or the lumbopelvic-hip complex.[9,11,71,72] Frequently described as a “box,” the musculature of the lumbopelvic-hip complex is used to delineate its boundaries. First described by Richardson et al.[73] and then adopted by other authors,[74,75] the anterior border of the box is formed by the abdominals, the posterior by the paraspinals and the gluteal muscles, the roof by the diaphragm, and the floor by the muscles of the pelvic floor and the hip girdle. This basic description has been modified by others to separate out the pelvic floor muscles and the hip girdle musculature, specifically the hip abductors and rotators, as forming the inferior and lateral borders, respectively.[69] These 29 pairs of muscles work in concert to stabilize the spine and pelvis as well as the kinetic chain during functional movement,[69,75,76] thereby maintaining functional stability.[74] In addition to the lumbopelvic-hip musculature, the core includes the neural structures associated with these muscles as well as the osseoligamentous structures of the lumbar spine, pelvis, and hip.[11,51]

1.3.2 Core Stability and Elements of Core Stability

Core stability is defined as the ability to maintain and control both the position and motion of the trunk over its base (pelvis and leg) during movement, thereby allowing for optimization of the production, transfer, and control of force and motion to the distal segments of the kinetic chain.[9] Core stability arises from the ability to maintain motor control as well as the muscular capacity (e.g., strength and endurance) of the lumbopelvic-hip complex.[10] Zazulak et al.[51] have further delineated that core stability is the ability to "maintain or resume a relative position of the trunk after perturbation." These definitions are in alignment with Riemann & Lephart,[53] who described functional joint stability as the ability of a joint (or segment) to remain in or promptly return to proper alignment through an equalization of forces and moments.

Originally defined by Panjabi,[77] core stability relies upon three subsystems: the passive musculoskeletal system, the active musculoskeletal system, and the neural and feedback system. These systems are synonymous with the components of the sensorimotor system that are used when discussing functional joint stability: static components, dynamic components, and central integration and processing.[54-59] The passive system is comprised of vertebrae, facet articulations, intervertebral discs, spinal ligaments, joint capsules, and the passive mechanical properties of the muscles. The active subsystem refers to the musculotendinous structures that surround the spinal column. The neural and feedback subsystem includes the cutaneous, articular, and musculotendinous mechanoreceptors as well as the neural control centers.

As with other joints or segments, core stability is reliant on the passive and active structures as well as motor control.[10,11,53,71,78] Since the passive structures of the lumbopelvic-hip complex primarily function to limit or restrict motion at the end ranges of motion,[78] it has been postulated that the passive components of the lumbopelvic-hip complex are not as critical to core stability and that core stability is the result of the active components and central integration and processing.[10,11] This, however, neglects to account for the mechanoreceptors that are present in the passive structures, which contribute to the feedback neural control over skeletal muscles in the region.[53,77]

These three systems, although separate systems, are interdependent and must function in concert to make the appropriate adjustments in response to changes in spinal posture as well as static and dynamic loads in order to maintain the intervertebral neutral zones within physiological limits.[77,79] This is in alignment with Riemann and Lephart's[53] description of how functional joint stability is maintained. In order to maintain proper alignment of a joint, the components of the sensorimotor system (sensory, motor, and central integration and processing) must function properly and work together.

Frequently core stability and core strength are used interchangeably in the literature. These terms, however, are not synonymous. Rather, core strength is an element of core stability.[80] According to Cowley and Swensen,[81] the key elements of core stability include muscular strength, endurance, and power as well as coordination of the lumbopelvic-hip complex. Although not specifically mentioned, underlying the coordination of the lumbopelvic-hip complex is proprioception. When assessing core stability it is important to include measures that can assess several of these neuromuscular characteristics to gain a more rounded perspective of core functional stability. One limitation in core stability research is that the majority of studies use only one of these key characteristics.

1.3.3 Core Stability and Lower Extremity Injury

Many describe the core as the "powerhouse"[69,74] or the center of the kinetic chain from which force is generated and transferred to the distal extremities, providing a stable foundation for movement of the extremities.[68,77,82] Adequate core stability has been advocated by health care professionals and performance specialists in order to decrease the risk of injury or re-injury. Possessing adequate core stability and/or core stability training have been promoted to decrease the risk of new or recurrent injury to the low back[83,84] as well as the extremities.[9,85] It has been theorized that deficits in core stability, particularly in muscular strength or endurance, result in the inability to effectively transfer energy through the core, thereby imparting greater stress on the tissue of the extremities.[68,75,76,86,87] If these deficits arise in an athletic population as a result of injury or deconditioning, then the risk of injury will increase if the athlete attempts to maintain the same level of performance.[68,76] As an extension of this belief, many athletic trainers, physical therapists, and strength and conditioning coaches prescribe core stability exercises or incorporate core stability training in order to decrease the risk of a first-time or recurrent injury.[68,75,87-92]

Few studies have explored the link between core stability and lower extremity injury. Leetun et al.[10] reported that athletes who sustained a lower extremity injury demonstrated significantly less hip abduction and external rotation strength; however, no significant differences were demonstrated in core endurance measures. In another study, female athletes who sustained a knee injury or a knee ligament/meniscal injury over a three year period exhibited significantly greater active repositioning error (proprioception) of the trunk as compared to female athletes who did not sustain a knee injury.[51] No significant difference in proprioception was noted between injured and uninjured male athletes. Further analysis of this data revealed that trunk angular displacement following a sudden release was significantly greater in athletes who sustained knee, ligament, and ACL injury as compared to uninjured athletes.[52] For female athletes, the final regression model for predicting risk of knee ligament injury included trunk angular displacement, trunk proprioception, and history of low back pain; however, only history low back pain was a significant predictor for knee ligament injury risk in male athletes.

It is possible that weakness of the core musculature results in risky knee kinematics. During a static task, Willson et al.[50] demonstrated significant weak to moderate positive correlations between knee frontal plane projection angle (FPPA) and trunk lateral flexion, hip external rotation, and knee flexion strength. Further, a significant difference in knee FPPA was found between males and females, with females presenting with a greater (increased valgus) knee FPPA than males.

According to Myer et al.,[93] the core provides the foundation upon which the muscles of the lower extremity produce or resist force. Since several muscles of the lumbopelvic-hip complex (or the core) cross both the hip and knee, poor conditioning of these muscles may result in faulty landing mechanics (e.g., increased knee valgus). These faulty landing mechanics may increase the risk of injury to the ACL. Hewett and Myer[12] have indicated that, in terms of evaluating ACL injury risk or prevention, positioning of the trunk should be considered.

Similarly, Zazulak et al.[51] stressed the importance of neuromuscular control of the trunk to minimize the risk of knee injury. One method to better understand how core stability may be related to lower extremity injury is to examine the relationship between neuromuscular characteristics of the trunk and knee kinematics that have identified in prospective studies[36,37] or in video analysis[38-44] to predict or be related to ACL injury.

1.3.4 Core Stability and Trunk and Lower Extremity Biomechanics

Deficits in neuromuscular control of the trunk may be related to increased risk of injury to the ACL.[12] As the lateral displacement of the trunk increases, there is a concomitant increase in the load on the knee, hip adduction torque, and knee abduction moment. While emphasizing that the focus of biomechanical contributors to ACL injury risk must extend beyond the knee, Hewett and Myer[12] proposed that addressing deficits in trunk neuromuscular control or focusing on control and positioning of the trunk may be a more practical approach to knee injury prevention since trunk position may be more noticeable and easier to explain, address, and correct to athletes and athletic trainers.

There have been a few studies that have investigated trunk and knee kinematics simultaneously. Video analysis of ACL injury events revealed that female athletes who sustained an ACL injury demonstrated significantly greater lateral trunk motion and knee abduction than males who sustained ACL injury.[41] Further, ACL-injured females demonstrated a progressive increase in knee abduction and significantly less forward trunk motion than controls. Kulas et al.[49] investigated the effect of kinematic trunk adaptation in response to trunk loading on knee anterior shear force and hamstring force estimated with biomechanical modeling during a double leg drop landing. Subjects who responded to the trunk load with greater trunk extension as compared to no load demonstrated a significant increase in peak and average knee anterior shear forces (17% and 35%, respectively). No significant increases were seen in the trunk flexion group. In addition, average hamstring force increased 13% in the trunk flexion group but decreased by 16% in the trunk extension group.

Blackburn and Padua[94] reported that trunk flexion angle can influence lower extremity kinematics. More specifically, intentionally increasing trunk flexion during a double leg drop landing resulted in significantly greater hip and knee flexion at initial contact as well as significantly greater maximum knee flexion angle. However, Farrokhi at al.[95] reported that in healthy subjects increasing trunk flexion angle during a forward lunge significantly decreases peak knee flexion angle as compared to a trunk-extended lunge and significantly increases peak hip flexion as compared to a trunk-extended and normal lunge. Further, the trunk-flexed lunge resulted in a significant increase in muscle activity of the hip extensors as assessed with electromyography. Identifying trunk kinematics and elements of core stability that contribute risky knee biomechanics will allow coaches and athletic trainers to target these trunk positions and elements, which may decrease the risk of injury and optimize athletic performance. One laboratory task that can be utilized to identify risky knee biomechanics is the stop jump-cut maneuver (SJCM).[96] This task mimics the movement that has been identified on video analysis to occur at the time of ACL injury as the SJCM requires an individual to land on a single leg and immediately perform a sidestep cut maneuver.

1.4 DEFINITION OF THE PROBLEM

For nearly 15 years, researchers have attempted to identify modifiable risk factors for noncontact ACL injury in order to develop and implement ACL injury prevention programs. Despite these efforts, ACL injuries persist as evidenced by a 1.3% annual increase in injury rate over a 16 year period at the collegiate level.[1] Although accounting for a relatively small percentage of overall injuries,[1-3,6] it is imperative to prevent ACL injuries due to the associated consequences such as significant time loss from athletics,[1,7,13-15] potential loss of athletic scholarship or salary, decreased academic performance,[23] and development of OA.[29,32-34] Limited research has been conducted investigating the relationship between trunk functional stability and lower extremity injury and knee kinematics. The available research has produced discordant results. One possible explanation is that most investigators typically examine only one neuromuscular characteristic of trunk functional joint stability as a proxy measure of core stability. Most often the core endurance tests developed by McGill et al.[97] are the tests employed; however, other neuromuscular characteristics such as trunk strength or proprioception may be a better proxy measure of trunk functional stability, particularly during dynamic, explosive tasks in a non-fatigued state. Further, simultaneous examination of multiple neuromuscular characteristics that underlie trunk functional stability will help to better delineate the role that each characteristic plays in potentially dangerous knee kinematics. More recently, trunk and lower extremity kinematics have been explored simultaneously but produced conflicting results.[41,49,94,95] Simultaneous analysis of trunk and knee kinematics during a dynamic task, such as a stop jump-cut maneuver, without instruction regarding trunk or knee positioning and without external loading will provide a better understanding of the innate relationship between trunk and knee kinematics.

1.5 PURPOSE

The purpose of this study was to examine the relationship between knee kinematics and neuromuscular characteristics of trunk functional stability as well as trunk kinematics in a healthy, physically active population. Trunk muscular strength and proprioception were assessed with isokinetic dynamometry and active joint position sense, respectively. Trunk and knee kinematics were assessed during a stop jump-cut maneuver using passive video-based motion analysis. The primary purpose of this study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex could predict knee valgus and flexion angles at initial contact as well as total knee valgus excursion and maximum knee flexion angle during a stop jump-cut maneuver.

1.6 SPECIFIC AIMS AND HYPOTHESES

Specific Aim 1: To determine if trunk muscular strength, proprioception, and kinematics as well as sex predict knee valgus angle at initial contact and total knee valgus excursion during a stop jump-cut maneuver (SJCM)

Hypothesis 1a: Trunk muscular strength, proprioception, kinematics, and sex will significantly predict knee valgus angle at initial contact during a SJCM. More specifically a greater knee valgus angle at initial contact will be predicted by:

1. Lower trunk extension muscular strength as measured by isokinetic testing
2. Higher (worse) trunk lateral flexion proprioception towards the dominant leg as assessed with active joint position sense

3. Greater lateral trunk displacement in the frontal plane at initial contact during the SJCM
4. Female sex

<u>Hypothesis 1b:</u> Trunk muscular strength, proprioception, kinematics, and sex will significantly predict total knee valgus excursion during a SJCM. More specifically a greater knee valgus excursion will be predicted by:

1. Lower trunk rotation muscular strength towards the cutting direction as measured by isokinetic testing
2. Higher (worse) trunk lateral flexion proprioception towards the dominant leg as assessed with active joint position sense
3. Greater maximum lateral trunk displacement in the frontal plane during the SJCM
4. Female sex

<u>Specific Aim 2:</u> To determine if trunk muscular strength, proprioception, and kinematics as well as sex predict knee flexion angle at initial contact and maximum knee flexion angle during a SJCM

<u>Hypothesis 2a:</u> Trunk muscular strength, proprioception, and kinematics as well as sex will significantly predict knee flexion angle at initial contact during a SJCM. More specifically a greater knee flexion angle at initial contact will be predicted by:

1. Greater trunk extension muscular strength as measured by isokinetic testing
2. Lower (better) trunk flexion proprioception as assessed with active joint position sense
3. Greater trunk flexion angle at initial contact during the SJCM
4. Male sex

Hypothesis 2b: Trunk muscular strength, proprioception, and kinematics as well as sex will significantly predict maximum knee flexion angle during a SJCM. More specifically a greater maximum knee flexion angle will be predicted by:

1. Greater trunk extension muscular strength as measured by isokinetic testing
2. Lower (better) trunk flexion proprioception as assessed with active joint position sense
3. Greater maximum trunk flexion angle during the SJCM
4. Male sex

1.7 STUDY SIGNIFICANCE

Although years of research have been dedicated to identifying modifiable risk factors for noncontact ACL injury and implementation of ACL injury prevention programs, the ACL injury rate has not declined and these injuries continue to pose a significant burden as demonstrated by the associated consequences such as immediate and long-term medical costs, potential loss of athletic scholarship or salary, decreases quality of life, and development of OA. Although core stability is frequently touted by healthcare and fitness professionals to be essential for injury prevention, there is limited research examining the relationship between the neuromuscular characteristics that underlie trunk functional stability and knee kinematics that have been identified through prospective studies and video analysis to be related to ACL injury. In addition, simultaneous study of trunk and knee kinematics without manipulation will provide better insight into how trunk kinematics influence knee kinematics. The knowledge gained from the current study will enable selection of appropriate core exercises when designing training programs with the intent of maximizing trunk functional stability to minimize the risk of ACL injury. If the relationship between trunk and knee kinematics can be determined, then coaches, athletic trainers, and athletes will be able to identify and correct potentially risky trunk position in an attempt to modify knee position and minimize risk of ACL injury.

2.0 REVIEW OF LITERATURE

2.1 KNEE INJURY

2.1.1 Epidemiology of Knee Injury

Injury to the lower extremity is a common occurrence in sports, accounting for almost half of all sports-related musculoskeletal injuries.[1,2] According to the National Collegiate Athletic Association (NCAA) Injury Surveillance System (ISS), the majority of these injuries occur at the knee or ankle.[1] The same is seen in adolescent sports, irrespective of sex.[2,98] Despite having similar overall injury rates, female adolescent soccer players are three to five times more likely to sustain a knee sprain than their male counterparts.[2,98] In adolescents, females who participated in sports at least four times per week are twice as likely to sustain ACL injury as compared to males with the same level of sports participation.[3]

Depending on the sport, the mechanism of ACL injury is most often noncontact.[1,4] In intercollegiate athletics, the greatest number of ACL injuries occur in football.[1] However, even though football is a collision/contact sport, greater than 40% of ACL injuries are still classified as noncontact.[6] These noncontact injuries occur without an external force applied to the knee; therefore, noncontact ACL injuries are potentially preventable through modulation of internal forces about the knee. As such, targeted intervention programs may be able to induce favorable changes in neuromuscular characteristics related to ACL injury risk. Research has demonstrated that intervention programs can increase strength,[99-101] optimize muscle activation,[99] and improve landing biomechanics.[99-105] Moreover, it has been demonstrated that some training programs can decrease the incidence of noncontact ACL injury in athletes[106-109] as well as overall injury rate.[110] Despite these efforts, there was a statistically significant increase in the rate of ACL injury from 1988 through 2004, with an average annual increase in injury rate of 1.3%.

2.1.2 Consequences of Anterior Cruciate Ligament Injury

Sports-related knee injuries are typically severe, with significant immediate and long-term consequences to the athlete such as time loss, depression resulting in delayed recovery and impaired academic performance, and development of OA later in life. In adolescent soccer, 30% of knee injuries resulted in time loss of greater than 7 days.[2] Similarly, in collegiate athletics, internal derangement of the knee was the first or second most common injury that resulted in greater than 10 days of time loss.[7,13-15,111-117] Although ACL injury accounts for only approximately 3% of all injuries, 88% of ACL injuries resulted in greater than 10 days of time loss[1] and greater than 59% of collegiate soccer or basketball players who sustained a knee injury required surgery.[4]

From a sports psychology perspective, significant injury can result in increased stress in an athlete, leading to alterations in emotion and behavior which, in turn, can affect physical recovery[19] as well as academic performance.[23] It has been reported that injured athletes experience depression,[16,18] anxiety,[17,18] and fear of re-injury or kinesiophobia.[17,20] Research has demonstrated that athletes who sustained an injury resulting in greater than one week of time loss are significantly more depressed than their uninjured counterparts[16] and also experience significant fluctuations in mood or emotion in the three weeks following injury.[18] Psychological stress[21] and anger expression[22] have been demonstrated to delay the healing process in the general population. In addition, in the student-athlete these moods alterations may lead to impaired academic performance. College-aged athletes who sustained an ACL injury demonstrated a significant drop in grade point average in the semester in which they were injured as compared uninjured students.[23]

Overall, individuals who have undergone ACL reconstruction demonstrate significant improvement in knee function survey scores; however, only 33-50% these individuals are able to return to sports.[20,28,29] Despite not returning to sports and having a significant reduction in physical activity,[26] many of these individuals will develop OA later in life. Epidemiological studies have reported that individuals who have sustained acute knee injury are 5 to 7.4 times more likely to develop OA later in life.[30,31] In terms of OA development following ACL injury, radiographic evidence of osseous change has been reported in 48-82% of athletes at long-term follow-up, with OA diagnosed in 15-51%.[29,32-34] In both male and female soccer players with history of ACL injury, osseous changes were evident in greater than 78% of athletes (males: 78%, females: 82%), with more than 41% (males: 41%, females: 51%) diagnosed with OA at 14 years and 12 years follow-up, respectively.[29,34] Further, the majority of subjects reported that knee symptoms impaired their quality of life.

2.1.3 Risk Factors of Noncontact Anterior Cruciate Ligament Injury

Risk factors for noncontact ACL injury can be classified as environmental, anatomical, hormonal, familial tendency, and neuromuscular.[35] While the majority of these classifications are non-modifiable, neuromuscular risk factors (i.e. movement patterns, muscle activation patterns, and insufficient muscle stiffness) can be modified through targeted intervention programs. The neuromuscular risk factors for noncontact ACL injury have been identified through prospective studies[36,37] as well as video analysis of ACL injury events.[38-44]

Prospective research has established that adolescent females who sustain noncontact ACL demonstrate significant kinematic and kinetic differences than those who remained injury-free. Hewett et al.[36] performed 3-dimensional biomechanical analysis during a drop vertical jump on 205 soccer, basketball, and volleyball players. Subjects were followed for two fall and one winter sports seasons for noncontact ACL ruptures. Those who sustained an ACL injury demonstrated greater knee abduction angle at initial contact and greater maximum knee abduction angle. Further, ACL-injured subjects demonstrated significantly less maximum knee flexion angle. Greater peak knee abduction moment, peak hip flexion moment, and vertical ground reaction force as well as shorter stance time were seen in ACL-injured subjects as compared to uninjured. It also was determined that knee abduction moment and angles at both initial contact and maximum were significant predictors of ACL injury.

Similar differences have been demonstrated prospectively in male and female ACL-R subjects who have returned to sports. Paterno et al.[37] assessed landing biomechanics during a drop vertical jump maneuver and postural stability in 56 ACL-R athletes who had returned to sport. Subjects were tracked for noncontact and indirect ACL injuries for 12 months after assessment. Significant kinematic and kinetic differences were demonstrated between the thirteen ACL-R subjects who sustained a second ACL injury and those who did not. Re-injured subjects demonstrated significantly greater knee valgus motion. Subjects with increased knee valgus motion were three times more likely to sustain an ACL re-injury than those with reduced valgus motion. In addition, hip moment impulse in the transverse plane was significantly different, with re-injured subjects demonstrating a net internal rotation moment and uninjured demonstrating a net external moment. Subjects with less hip external moment were over eight times more likely to sustain an ACL re-injury than those with greater hip external rotation moment.

2.1.4 Video Analysis of Noncontact Anterior Cruciate Ligament Injury Mechanism of Injury

Video analysis of actual ACL injury has provided additional insight into the mechanism of injury for noncontact ACL ruptures. Ebstrup and Bojsen-Moller[40] described three out of fifteen ACL injuries that occurred during a single season during indoor ball games (handball and basketball). Of these, one occurred during contact with another player. The remaining two injuries were sustained via a noncontact mechanism, occurring during plant-and-cut maneuvers. The descriptions of these injuries were similar, with the knee in valgus coupled with internal rotation of the thigh.

Similar results were seen on a larger scale when describing ACL injuries that occurred in female team handball players.[38] Twenty videos of ACL injuries were analyzed using standardized forms by three handball experts for activity at the time of injury and by three physicians with clinical and research experience on ACL injuries for lower extremity alignment at time of injury. Of the 20 injuries, 90% were classified as noncontact and 80% occurred during plant-and-cut maneuvers or single-leg landings. In addition, the knee was described in all cases as being only slightly flexed and in valgus accompanied by either internal or external tibial rotation.

Krosshaug et al.[42] used multiple raters to describe ACL injuries in male and female basketball players during high school, college, or professional games. Of the 39 injuries, 72% were categorized as noncontact. Of the 35 noncontact injuries, 30 videos were of high enough quality to analyze knee and hip motion. At both initial contact and 50ms after initial contact, the knee and hip flexion angles were significantly greater in females than in males. However, mean knee flexion at initial contact ranged between 8° and 15° across males and females, indicating a relatively extended knee position. No significant differences in knee valgus at initial contact were seen between the sexes; however, females had significantly greater valgus 50ms after initial contact. Further, the raters indicated that valgus collapse occurred in 53% of the females but only 20% of the males. The relative risk for knee valgus collapse at the time of ACL rupture was 5.3 times greater in female as compared to male basketball players.

Using video editing and computer software to measure lower extremity kinematics, Boden et al.[44] compared data of male and female subjects who sustained a noncontact ACL injury to uninjured athletes performing similar maneuvers. Lower extremity kinematics were calculated at initial contact and the ensuing five frames. Subjects who sustained an ACL injury demonstrated significantly less plantarflexion at initial contact as well as significantly greater hip flexion at initial contact and through the first three frames. No differences were seen in knee flexion at any time point. Although there were no differences in knee abduction at initial contact, ACL injured subjects demonstrated a progressive increase in knee abduction, with statistically greater knee abduction in frames three through five.

Hewett et al.[41] used video editing and computer software to measure knee and trunk kinematics during noncontact ACL injury at initial contact and 50ms, 100ms, 150ms, and 200ms post-initial contact in male and female athletes. Kinematics also were calculated in non-injured athletes performing similar tasks. Females who sustained an ACL injury demonstrated significantly greater knee abduction during landing as compared to male-injured subjects. No differences were demonstrated between females who sustain an ACL injury and female controls at initial contact. However, female ACL-injured subjects demonstrated significantly greater knee abduction than controls from 100ms through 200ms post-initial contact, indicative of dynamic knee valgus collapse. No differences were seen in knee flexion. Female ACL-injured subjects demonstrated significantly greater lateral trunk angle than male ACL-injured subjects across all time points as well as significantly less forward trunk lean than female controls.

Based on these prospective studies and video analyses, it appears that noncontact ACL injuries typically occur during single leg support of plant-and-cut or jump-landing maneuvers. Individuals who sustain a noncontact ACL injury demonstrate significantly greater knee abduction at initial contact and maximum. Knee abduction moment and angles at both initial contact and maximum are predictors of noncontact ACL injury. Further, at the time of injury, the knee is in a relatively extended position (<30°), regardless of sex. For females, dynamic valgus collapse typically occurs as well. In addition, trunk position may also play a role in noncontact ACL injuries, particularly in females, but has received little attention in the literature. The ability to sense and control the position of the trunk may influence the ability to sense and control the position of the knee, thereby reducing the risk of knee injury. The sensorimotor system is critical in modulating these risky positions in order to maintain functional joint stability and reduce the risk of injury.

2.2 FUNCTIONAL JOINT STABILITY

2.2.1 Sensorimotor System

In 1997, a comprehensive definition of the sensorimotor system, a subcomponent of the motor control system, was developed during the *Foundations of Sports Medicine Education and Research* workshop.[54] The components of the sensorimotor system include both afferent (sensory) and efferent (motor) pathways as well as the central integration and processing that occurs within the central nervous system. Proper functioning of the sensorimotor system is intimately related to the maintenance of functional joint stability or the ability to maintain joint homeostasis during human movement.[118] Functional joint stability refers to the preservation or prompt restoration of proper joint alignment[53] through an equalization of force and moments and is governed by both static and dynamic elements.[53,55-57,59]

2.2.2 Static and Dynamic Elements of Functional Joint Stability

The static components of functional joint stability include the joint capsule, ligaments, cartilage, and bony geometry of the joint.[56] In addition, friction within the joint is considered another static element that helps to maintain stability. Assessment of static stabilizers is fairly straightforward and typically occurs through clinical assessments such as ligament stress tests and joint arthrometry. The dynamic components of functional joint stability are more difficult to assess as the dynamic components encompass the musculotendinous structures that cross a given joint and the associated neural pathways associated with these tissues.[53,56] The ability of the musculotendinous structures and neural pathways to work effectively and efficiently is greatly impacted by the biomechanical and physical characteristics (e.g., range of motion, muscular strength, muscular endurance) of a particular joint.[53] Dynamic stabilization of a joint and postural control result from feedforward and feedback neuromotor control. Feedforward strategies are anticipatory, thereby enabling muscle tension to be modified in preparation for an event or movement.[59,61] Feedback controls are initiated after the sensory detection of a destabilizing event. Both are influenced by previous experience and integrate information from the visual, vestibular, and somatosensory systems.[61] Feedforward commands occur on an intermittent basis whereas feedback controls are regulated moment-to-moment.

2.2.3 Proprioception

Somatosensation refers to the acquisition of afferent sensory information arising from peripheral mechanoreceptors, thermoreceptors, and pain receptors. This sensory information can be categorized into four modalities: discriminative touch, proprioception, nociception, and temperature sense.[60] Proprioceptive signals arising from peripheral mechanoreceptors provide position and movement information to the central nervous system independent of visual input.[60] This information is then utilized to influence postural control, joint stability, and conscious sensation.[53,63]

According to Matthews,[119] conscious proprioceptive senses were first defined by Sherrington and include the following submodalities: posture, passive movement, active movement, and resistive movement. Currently, the conscious proprioceptive submodalities are considered to be joint position sense, kinesthesia, and force sense.[53,58,59] The ability to accurately reposition, identify, or replicate limb position either actively or passively is joint position sense.[55,60,64,65] Kinesthesia, or movement sense, refers to the capability to detect passive movement of a joint or segment.[60,64-66] Finally, force sense signifies the proficiency of a muscle group to reproduce a target torque or to identify a difference in weight (weight discrimination).[58,59]

Proprioceptive information is initiated by mechanical deformation of mechanoreceptors located in articular, cutaneous, and musculotendinous tissue.[120,121] This mechanical deformation results in the creation of an action potential that creates an afferent neural signal that is transmitted to the central nervous system.[59,60,120,122] It has been postulated that proprioceptive information may be weighted differently, with a greater reliance of proprioceptive information arising from muscle, tendon, ligament, and joint capsule rather than cutaneous or fascial sources.[123]

Mechanoreceptors can be categorized as either slow adapting or quick adapting, with slow adapting mechanoreceptors continuing to generate a neural signal with continued stimulation and quick adapting mechanoreceptors creating a neural signal within milliseconds of stimulus detection but then stopping.[59] Slow adapting mechanoreceptors, which include Ruffini endings and Golgi tendon-like organs, provide information regarding joint position, including joint angle, velocity, and intraarticular pressure.[124] Quick adapting mechanoreceptors include Pacinian corpuscles and relate information regarding joint motion and acceleration.[59,60,124]

Musculotendinous mechanoreceptors include muscle spindles and Golgi tendon organs. Muscle spindles, which lie in parallel with the extrafusal muscle fibers, are responsible for detecting muscle length as well as rate of change in muscle length[59,60,124-129] through the sensory organ that is located in the central third of the intrafusal muscle fiber.[127,129] The gamma motor neuron controls the contractile regions of the intrafusal muscle fiber, which is located in the outer two thirds of the fiber. The sensitivity of the muscle spindle is increased by stretching of the extrafusal muscle fibers as well as afferent signals from articular, cutaneous, and musculotendinous mechanoreceptors.[59] Muscle spindles are a vital component in joint position as change in muscle length is highly related to change in joint angle.[127] Golgi tendon organs, which are located at the musculotendinous junction, provide information regarding tension development within the musculotendinous structure during active contraction or force development.[60,124,125,129] Since the Golgi tendon organs are capable of detecting the development of both low and high levels of tension, these mechanoreceptors influence muscle tension to allow for fine and gross movements as well as protection of the musculotendinous unit from injury.[127,129]

Efficient functioning of the sensorimotor system, including both afferent (sensory) and efferent (motor) pathways as well as the central integration and processing, are crucial for the maintenance of functional joint stability. Previous research has investigated proprioception and neuromuscular characteristics of the lower extremity as related to ACL injury; however, limited research has investigated how functional joint stability of the core may be related to functional joint stability of the knee.

2.3 THE CORE AND CORE STABILITY

2.3.1 The Core and Core Musculature

The core is typically described as the lumbopelvic region[67-70] or the lumbopelvic-hip complex[9,11,71,72] and encompasses all of the muscles of the lumbar spine, pelvis, and hip as well as the osseoligamentous structures in these regions.[11,51] First described as a “box” by Richardson et al.,[73] the abdominal muscles form the anterior border, the paraspinals and gluteal muscles form the posterior border, the diaphragm the roof, and the muscles of the pelvic floor and hip girdle the floor.[74,75,130] This basic description has been modified by other researchers to include the obliques and the latissimus dorsi as forming the sides[131] and the hip abductors and rotators as forming the inferior and lateral borders, respectively.[69]

Bergmark[132] classified the musculature of the core as belonging to either the global or the local system. This classification scheme is based on the function of the muscles rather than location. The global system is comprised of the muscles that are responsible for load transfer between the thoracic cage and the pelvis whereas the local system is comprised of the muscles that act directly on the lumbar spine to maintain its curvature as well as to maintain mechanical stability of the lumbar spine. Although activation of the global system is a function of the line of action of the outer load and activation of the local system is a function of changes in posture of the lumbar spine, both systems are activated in response to magnitude changes of the outer load. The musculature of the global and local systems, as modified by Richardson et al.,[73] as well as the passive components of core stability are presented in Table 1.

The back and abdominal musculature work together to provide functional stability to the lumbopelvic-hip complex as well as to produce trunk movement.[133-135] The erector spinae, along with the multifidus and semispinalis thoracis, are the primary back extensors when contracted bilaterally. Trunk flexion occurs by bilateral contraction of the rectus abdominis and psoas major. Unilateral contraction of the quadratus lumborum, iliocostalis thoracis and lumborum, multifidus, and external and internal obliques produces lateral flexion. Rotation of the trunk occurs via unilateral contraction of the rotatores, multifidus, iliocostalis, longissimus, and the external oblique and the opposite internal oblique.

Table 1. Global vs. Local Systems of the Core Musculature

Passive Components	*Active Components*	
	Global System	*Local System*
Bones	Internal obliques	Multifidus
Cartilage	External obliques	Interspinalis
Ligaments	Rectus abdominus	Intertransversari
Tendons	Quadratus lumborum (lateral portion)	Quadratus lumborum (medial portion)
Fascia	Psoas major	Spinalis
	Latissimus dorsi	Transverse abdominis
	Erector spinae (thoracic portion of longissimus thoracis and iliocostalis lumborum)	Erector spinae (lumbar portion of longissimus thoracis and iliocostalis lumborum)
		Internal oblique (via insertion into thoracolumbar fascia)

The musculature of the back can be divided into three layers: superficial, intermediate, and deep. The more superficial muscles are primarily responsible for gross movements and tend to be longer, spanning multiple spinal segments.[133,134] The deep muscles deal almost exclusively with fine adjustments between vertebrae to maintain stability and span only one or two vertebrae.[133,134] In addition, these smaller deeper muscles have higher densities of muscle spindles, potentially allowing for greater proprioceptive feedback and spinal stability.[133] The abdominal muscles also are arranged in layers, with superficial layer is created by the external oblique, the intermediate layer by the internal oblique, and the deep layer by the transverse abdominis.[135]

2.3.2 Core Stability

Core stability is dependent on passive and active elements. Core stability is the ability to preserve and control trunk position and motion over the pelvis and leg during movement and is essential for the production, transfer, and control of force and motion to the extremities.[9] In addition, core stability refers to the ability to conserve or revert back to equilibrium or proper position following perturbation.[9,51] These definitions are in accordance with the definition of functional joint stability typically applied to joints of the extremities.[53]

As with other regions of the body, stability of the core is contingent on passive, active, and neural components.[10,11,53,71,78] Specifically applied to the core, Panjabi[77] labeled these three subsystems as the passive musculoskeletal system, the active musculoskeletal system, and the neural and feedback system. Components of each of these three systems are presented in Table 2. Functional stability of the lumbar spine relies predominantly on the surrounding musculature and neural control with the ligaments and bony architecture providing some assistance at the extremes of motion. In addition, the fascia and intraabdominal pressure also contribute to core stability.

Table 2. Panjabi Core Stability Subsystems

Passive Musculoskeletal System	Active Musculoskeletal System	Neural and Feedback System
Vertebrae	Musculotendinous structures	Neural control centers
Facet articulations		Cutaneous mechanoreceptors
Intervertebral discs		Articular mechanoreceptors
Spinal ligaments		Musculotendinous mechanoreceptors
Joint capsules		
Muscle passive mechanical properties		

2.3.3 Assessing the Dynamic Elements of Core Stability

Assessment of the passive or static components of functional stability is fairly uncomplicated and typically occurs through clinical assessments such as ligament stress tests and joint arthrometry. In contrast, the dynamic components are more difficult to assess. Assessment of the dynamic components entails evaluating the musculotendinous structures that cross a given joint and their associated neural pathways.[53,56] The ability of these structures and pathways to work effectively and efficiently is greatly impacted by the biomechanical and physical characteristics (e.g., range of motion, muscular strength, muscular endurance) of a particular joint.[53] Cowley and Swensen[81] have identified the following as important elements of core stability: muscular strength, muscular endurance, muscular power, and coordination of the lumbopelvic-hip complex.

Assessment of these sensorimotor components can be made by evaluating variables along afferent or efferent pathways, motor output, or a combination of these.[58] It is currently not possible to isolate the central integration and processing component of sensorimotor control. Assessment techniques can be categorized as either peripheral afferent acquisition and transmission measurements or efferent transmission measurements. Examples of each are presented in Table 3. Specific assessment techniques employed in the current study will be discussed in greater detail below.

Table 3. Sensorimotor System Assessment Techniques

Peripheral Afferent Acquisition and Transmission Measurements	Efferent Transmission Measurements
Proprioception	Nerve Conduction Testing
Somatosensory Evoked Potentials	Muscle Activation Patterns
	Muscle Performance Characteristics
	Kinetic Measurements
	Kinematic Measurements
	Postural Control Measures
	Muscle and Joint Stiffness

2.3.3.1 Assessment of Muscle Performance Characteristics

Since the physical characteristics of the musculotendinous unit can influence functional stability, muscle performance characteristics can be evaluated to provide insight into the efferent component of functional stability. Muscle performance characteristics include strength, endurance, and power.[136] These characteristics are related; however, they are not necessarily highly correlated with each other.[137] A deficit in any one or combination of these characteristics may result in impairment, dysfunction, or increased risk of injury. Intervention programs must address all three of these elements in order to minimize the risk of injury as well as optimize performance.

Muscular strength refers to the development of tension within the muscle in response to an external load or force[138] and is the maximal force that a muscle or group of muscles can create.[139-141] The ability of the neuromuscular system to generate adequate force is important for both injury prevention and optimal performance. There are several methods by which muscular strength can be evaluated. A repetition maximum, whether single or multiple repetitions, can be a safe and effective method to identify the dynamic strength of a muscle or group of muscles.[137,140,141] Additional methods include handheld dynamometry and isokinetic dynamometry, which allow for selection of type of muscle contraction (e.g., concentric, eccentric, isometric), speed of the contraction, and joint angle.[140]

Muscular endurance is the ability of a muscle to perform work over time[136] or to generate and sustain force for an extend period of time.[138,139,141] Typically, it is a low level muscular force that is generated. Muscular endurance is necessary to facilitate stabilization and proper alignment during activities of daily living as well as athletic movement. Improving or maximizing muscular endurance, particularly in those with underlying deficits or impairment, may be more vital than improving muscular strength.[137] Further, muscular endurance may be critical in minimizing the risk of injury as greater muscular endurance may help to prolong the time to fatigue. Muscular endurance can be assessed by evaluating the number of repetitions performed (e.g., push-up test, curl-up test, YMCA bench press test) or timed isometric tests (e.g., wall squat, side bridge, trunk flexor endurance test, and modified Biering-Sorensen back extensor endurance test).[141]

Muscular power refers to the work produced by a muscle per unit of time.[136,139] In contrast to muscular endurance, muscular power refers to the ability to generate a high or the highest amount of force within the shortest time period.[138] Muscular power is dependent upon both the ability to provide adequate stabilization and muscular strength, and is essential in athletic movement.[139] Common assessment methods for muscular power include the vertical jump, standing broad jump, two- or three-hop tests, and the Margaria Kalamen Power Test.

2.3.3.2 Assessment of Proprioception

There are three submodalities of conscious proprioception: joint position sense, kinesthesia, and force sense.[53,58,59] Joint position sense describes the ability to accurately identify or replicate body position and can be assessed actively or passively as well as with closed kinetic chain or open kinetic chain motions.[55,60,64,65] Instrumentation utilized in assessing joint position sense includes goniometers, potentiometers, video, and visual analog scales.[142-147] Kinesthesia, or movement sense, describes the ability to detect passive movement of a joint or segment[60,64-66] and is evaluating by measuring threshold to detection of passive motion direction with an isokinetic dynamometer or custom motorized jig.[145,147-149] Force sense describes the ability to reproduce a force or identify a difference in weight[58,59] and can be evaluated with an isokinetic dynamometer or weight system.[147,149]

The goal of assessing the submodalities of conscious proprioception is to evaluate peripheral afferent acquisition and transmission by the mechanoreceptors. Mechanoreceptors are located within articular, cutaneous, and musculotendinous tissue; however, the goal is to isolate the mechanoreceptors located within the target tissue, specifically articular and/or musculotendious tissue.[58] In order to attain this goal, cutaneous mechanoreceptor stimulation can be minimized through the use of compressive sleeves, anesthesia, or ischemia.[58] In addition, other sensory input such as visual and auditory stimuli should be minimized through the use of blindfolds and earplugs/headphones, respectively. As these methods are meant to assess the conscious proprioceptive submodalities, the ability of the subject to maintain focus and concentration is vital.

2.3.3.3 Kinematic Assessment

Biomechanical evaluation also can provide insight into the functioning of the sensorimotor system. Kinematics describes movement without regard to the forces producing motion and includes linear and angular displacement, velocity, and acceleration.[150] Kinematic analyses are conducted using equipment such as high-speed cameras, electromagnetic tracking devices, and accelerometers.[151-155] Kinematic assessment allows for the study of human movement during functional or sports-specific tasks. Examples of uses of kinematic assessment include the identification of risk factors of injury,[36,37] modifications of these risk factors by training or rehabilitation interventions,[99-105] and influence of other factors (e.g., external load, fatigue) on kinematics.[151,152,156]

2.4 CORE STABILITY AND THE LOWER EXTREMITY

It is believed that the core serves as the center of the kinetic chain, serving to generate and transfer force to the distal extremities as well as provide a stable foundation for movement of the extremities.[68,77,82] The core musculature, more specifically the multifidus and the transverse abdominus, have been demonstrated to activate prior to upper and lower extremity movement.[82,157,158] However, it is through pre-activation and continued, coordinated activation of all of the core musculature that results in the creation and maintenance of a stable base.[67,89,92,159-161] Weakness in any of the core musculature may increase the risk of injury[9,83-85] and decrease sports performance.[68,74,87,88,162] Research investigating the relationship between the neuromuscular characteristics underlying core stability and previously identified risk factors for injury is limited.

2.4.1 Kinetic Link between the Trunk and Lower Extremity

It is a common belief that adequate core stability is essential to decrease the risk of new or recurrent injury to the low back[83,84] and the extremities.[9,85] The core is the center of the functional kinetic chain,[9,80] providing the foundation upon which movement of the upper and lower extremities occurs.[68,77,82] The core, also referred to as the "powerhouse" or the engine of limb movement, is responsible for the generation and transfer of force to the extremities.[74] In theory, deficits in core stability result in ineffective energy transfer through the core, increasing stress transferred to tissues of the extremities.[68,75,76,86,87] Injury or deconditioning can result in core stability deficits and can increase the risk of injury as the athlete attempts to maintain the same level of performance.[68,76]

Theoretically, the lumbopelvic-hip complex can influence the positioning of the lower extremity, particularly the knee.[12,45] Deficits in neuromuscular control at the hip can result in the inability to stabilize the pelvis, causing medial-lateral displacement of the trunk. Due to the anatomic link of the femur and the trunk through the hip, this medial-lateral displacement of the trunk can influence frontal plane positioning of the knee.[12,45] For example, hip abduction weakness can result in either contralateral pelvis drop or contralateral hip elevation. If the contralateral pelvis drops, then the center of mass moves away from (medial to) the stance leg, which increases the varus moment at the knee. Knee varus places increased tensile strain on the lateral structures (e.g., lateral collateral ligament, iliotibial band) and increased compressive strain on the medial compartment of the knee. In contrast, when hip abductor weakness is compensated with contralateral pelvis elevation, the trunk is displaced lateral to the stance leg, resulting in shifting of the center of mass towards the stance leg. This creates a valgus moment at the knee, which results in tensile strain on the medial knee structures (e.g., medial collateral ligament, anterior cruciate ligament) and compression of the lateral compartment of the knee. Despite the theoretical contribution of the trunk to the lumbopelvic-hip-lower extremity relationship, research investigating this contribution is sparse, with the majority of knee research stopping at/limited to the hip. As such, the influence of neuromuscular characteristics related to core stability and of trunk kinematics on lower extremity kinematics has not been clearly established in the literature.

2.4.2 Core Stability and Lower Extremity Injury

Few studies have been conducted that examine the relationship between the neuromuscular characteristics that contribute to core stability and lower extremity injury. These neuromuscular characteristics included hip muscular strength, core muscular endurance, and trunk proprioception. Deficits in some, but not all, of these elements were predictive of low back and lower extremity injury. In addition, research has demonstrated that neuromuscular training programs that incorporate core stability training are effective in decreasing the occurrence of and disability associated with low back and overall injury as well as improving postural stability.

Leetun et al.[10] measured preseason core stability in 139 male and female intercollegiate basketball and cross country athletes and then prospectively tracked injuries in the ensuing competitive season. Core stability was assessed using the following tests: isometric hip abduction, isometric hip external rotation, the modified Biering-Sorensen test, side bridge test, and either the straight leg lowering test (basketball) or the McGill flexor endurance test (cross country). During the study period, 48 injuries to the back or lower extremity were recorded, with athletes who sustained a lower extremity injury demonstrating significantly less hip abduction and external rotation strength, but no significant difference in core muscular endurance measures. The authors concluded that core stability is essential for the prevention of lower extremity injury and that the relationship between core strength and lower extremity biomechanics needs to be explored in order to better understand this relationship.

Zazulak et al.[51] investigated the relationship between core proprioception and knee injury in 277 male and female intercollegiate athletes over a three year period. Core proprioception was assessed using two tests of joint position sense. Subjects were seated and the pelvis was passively rotated to the target position. Subjects then attempted to actively or passively rotate to the target position. During the three year follow-up period, a total of 25 knee injuries were recorded. Female athletes who sustained a knee injury or a knee ligament/meniscal injury exhibited significantly greater active repositioning error of the trunk as compared to female athletes who did not sustain a knee injury. No significant difference in proprioception was noted between injured and uninjured male athletes. In addition, the authors reported a 2.9-fold increase of knee injury and a 3.3-fold increase of ligament/meniscal injury in athletes for each degree increase in repositioning error.

Zazulak et al.[52] combined the data from the previous study with demographic data, previous injury history, and another measure of core stability to develop prediction equations for lower extremity injury. Core stability was assessed by measuring trunk displacement angle into extension, flexion, and lateral flexion after a sudden force release. Overall, trunk displacement was significantly greater in athletes who sustained knee, ligament, and ACL injury as compared to uninjured athletes. For female athletes, the final regression model for predicting risk of knee ligament injury included trunk displacement, trunk proprioception, and history of low back pain; however, only history low back pain was a significant predictor for knee ligament injury risk in male athletes.

Although there is limited research investigating the relationship between core stability and lower extremity injury, research has explored the effect of neuromuscular training programs that incorporate core stability exercises on low back pain and postural stability. In respect to low back pain, conflicting results have been reported. Core stability exercises resulted in no significant reduction in the occurrence of low back pain in collegiate athletes over a single year.[163] In contrast, collegiate gymnasts who participated in a preseason core training program reported no new episodes of low back pain.[83] Core training also reduced the recurrence of low back pain[164] and improved disability associated with low back pain[165-167] in the general population. Similar results were seen for overall injuries in firefighters, with time lost due to injuries reduced by 62% and the number of injuries reduced by 42% over a 12-month period in firefighters who participated in a core strengthening and flexibility program.[168]

Deficits in postural stability have been linked to increased risk of lower extremity injury.[169-172] Several studies have investigated the effects of core stability training on postural stability. Filipa et al.[173] reported that an 8 week neuromuscular training program that emphasized the core significantly improved performance on the Star Excursion Balance Test (SEBT) in female high school soccer players. Core training also has been reported to significantly improve postural stability in sprinters[174] and in females with a previous history of low back pain.[175] However, no improvement in the SEBT was seen in recreational and competitive runners following a 6 week core stability training program.[90]

Despite the ability of neuromuscular training programs that include core stability training to reduce the occurrence of and disability associated with low back and overall injury as well as improving postural stability, the associated underlying mechanisms are unknown. Further, due to the limited number of studies as well as the limited number of core stability neuromuscular characteristics utilized, the relationship between core stability and lower extremity injury remains unclear. One method to better understand how core stability may be related to lower extremity injury is to examine the relationship of multiple neuromuscular characteristics underlying core stability as well as trunk kinematics with previously identified and theoretical biomechanical risk factors of lower extremity injury in the laboratory setting.

2.4.3 Core Stability and Lower Extremity Biomechanics

According to Myer et al.,[93] the core provides the foundation upon which the muscles of the lower extremity produce or resist force. Further, since several muscles of the lumbopelvic-hip complex (or the core) cross both the hip and the knee, poor conditioning of these muscles may result in faulty landing mechanics (e.g., increased knee valgus), which may increase the risk of injury to the ACL. Hewett and Myer[12] have indicated that, in terms of evaluating ACL injury risk or prevention, that positioning of the trunk should be considered. Similarly, Zazulak et al.[51] stressed the importance of neuromuscular control of the trunk in order to minimize the risk of knee injury.

Despite this theoretical framework, there have been few studies that have investigated the relationship between neuromuscular characteristics of core stability and positioning of the trunk and/or the knee. There are several studies that have described how changes in trunk position influence knee kinetics and kinematics during dynamic tasks. However, the results may not be generalizable to athletic tasks or an athletic population due to methodological issues (e.g., intentional trunk positioning beyond what is seen in athletics, weighting of the trunk).

There appears to be a limited number of studies that have investigated the core as the lumbopelvic-hip complex rather than only the hip relative to lower extremity biomechanics. Willson et al.[50] explored the relationship of isometric strength of the trunk, hip, and knee with knee frontal plane projection angle (FPPA) during a single leg squat to 45° on the dominant leg. Once the target angle was attained, a photograph was taken and the knee FPPA was determined using digitizing software. Significant weak to moderate positive correlations were found between knee FPPA and trunk lateral flexion strength, hip external rotation strength, and knee flexion strength. Abt et al.[156] examined the effect of fatigue of the core musculature on lower extremity kinematics in cyclists. Following fatigue, cyclists demonstrated significant increases in total frontal plane knee motion, sagittal plane knee motion, and sagittal plane ankle motion. The authors concluded that these fatigue-induced kinematic changes may increase the risk of knee injury; therefore, endurance of the core musculature improves core stability and fosters proper alignment of the lower extremity during cycling.

Several studies have investigated how changing the position of the trunk influences knee kinetics and kinematics. Kulas et al.[49] examined the effect of trunk position adaptation (trunk flexor vs. trunk extensor) to an applied trunk load on knee anterior shear and hamstring muscle forces during a double leg drop landing. Forces were estimated using a biomechanical knee model. Subjects who landed with greater trunk extension relative to the no load landing demonstrated significant increases in peak (17%) and average (35%) knee anterior shear forces. No significant increases were seen in the trunk flexor group. In addition, average hamstring force decreased by 16% in the trunk extensor group but increased 13% in the trunk flexor group when comparing the loaded and unloaded landings. However, there were no differences between the groups in knee flexion angle at initial contact, knee flexion at peak anterior shear force, or maximum knee flexion. Based on these findings it appears that trunk flexion facilitates activation of the hamstrings, mitigating anterior shear forces at the knee during loaded landing, which may decrease strain on the ACL.

Blackburn and Padua[94] also investigated the influence of trunk position on knee kinematics during double leg drop landing. Subjects first performed a double leg drop landing with their natural or preferred trunk position and then repeated the task while actively flexing the trunk. During the trunk flexed landing, subjects landed with significantly greater trunk, hip, and knee flexion at initial contact as well as demonstrated significantly greater peak trunk and hip flexion. However, there were no significant differences in knee valgus, hip adduction, or hip internal rotation. The authors concluded that landing with increased trunk flexion may be a method by which lower extremity kinematics could be altered to reduce load on the anterior cruciate ligament, thereby decreasing the risk of injury. One potential limitation is that subjects may have unconsciously altered hip and knee kinematics during the flexed landing as they were provided instructions on how to perform the task; therefore, the relationship between trunk flexion and lower extremity kinematics demonstrated in this study may not be reflective of the relationship under natural conditions.

The influence of trunk position on neuromuscular characteristics of the lead leg during a forward lunge also has been investigated.[95] Biomechanics and electromyography were collected on ten healthy subjects during normal, trunk forward, and trunk extended lunges. In the trunk flexed condition, a significant decrease in knee flexion angle occurred as compared to a trunk extended lunge along with a significant increase in peak hip flexion angle as compared to a trunk extended and normal lunge. Coupled with these kinematic changes, a significant increase in muscle activity of the hip extensors as assessed with electromyography was seen in the trunk flexed lunge as compared to the trunk extended and normal lunges. However, these changes in muscle activity may not be clinically relevant as the improvement were relative small in relation to the maximum voluntary contraction.

Hewett et al.[41] compared lateral trunk angles and knee abduction angles during ACL-injury events to similar landing and cutting maneuvers that did not result in ACL injury in male and female professional basketball players. Trunk and knee kinematic measurements were obtained from the five sequential frames occurring after initial contact (0 to approximately 200ms post-initial contact) of the injured or control leg. Female athletes who sustained an ACL injury demonstrated significantly greater lateral trunk motion and knee abduction at initial contact when compared to males who sustained ACL injury. Further, ACL-injured females demonstrated a progressive increased in knee abduction and significantly less forward trunk motion than controls. Based on these results, it appears that during ACL injury, specifically in females, that lateral trunk displacement is related to knee abduction associated with ACL loading and injury. In addition, it is plausible that the decrease in trunk flexion may place the hamstrings at a mechanical disadvantage in resisting anterior translation of the tibia, allowing for increased loading on the ACL.

2.5 METHODOLOGICAL CONSIDERATIONS

2.5.1 Stop Jump-Cut Maneuver

There are a variety of dynamic tasks that have been utilized in research investigating noncontact ACL injuries. These tasks include run-to-cut or side-step cutting maneuvers,[8,176] stop jumps,[177,178] and drop landings.[37,49,105,179] Tasks that involve cutting have been performed both as anticipated (direction of cutting known prior to initiating the task) and unanticipated (direction of cutting cued after initiating the task) tasks. Tasks that involve jumping or landing have been performed both as single-legged and double-legged as well as with a cutting or secondary jumping maneuver performed after initial contact from the first landing.

Video analyses of ACL injury events have provided insight into the mechanism of noncontact ACL injury. Typically these injuries occur during plant-and-cut maneuvers[38,40] or single-leg landings.[38] In addition, previous research has demonstrated that laboratory tasks that involve lateral movement after initial contact induce riskier knee kinematics.[180] The SJCM[96] selected in the current study incorporates both a plant-and-cut and single-leg landing. Utilizing a stop jump approach from a standardized distance (40% of subject's height) facilitates the data collection process as fewer trials will have to be discarded and repeated due to missing the force plate or incorrect approach speed as can occur with a running approach. In order to ensure that the subject has not "turned" early into the cutting maneuver, the orientation of the pelvis and of the foot relative to the global coordinate system will be calculated during the squat phase of the SJCM and at initial contact. In addition, the orientation of the pelvis relative to the global coordinate system will be calculated upon completing the cutting maneuver. Minor deviation in pelvis orientation and foot orientation will be permitted in order to allow for natural movement during the SJCM. If it appears that a subject has turned in the direction of the cut prior to initial contact (pelvis and foot ≥10°) or has not completed a 45° cut (±10°), then the pelvis orientation and foot orientation relative to the global coordinate system will be determined. If it is determined that either of these has occurred, then the trial will be discarded and repeated. Analysis of data collected in our laboratory using the same task and methodology[181] indicates that there is minimal change in pelvis orientation relative to the global coordinate system (start position:-0.53°±2.29°; initial contact:-1.63°±7.81°) during the jump forward and that subjects complete a 45° cut (47.20°±2.23°). Previous research has demonstrated that knee kinematics obtained using this task have excellent within session reliability (ICC>0.93).[96]

2.5.2 Measurement of Knee and Trunk Kinematics

Kinematics includes the description of position, velocity, and acceleration, without consideration of the internal or external forces that create movement or change in movement.[182,183] Change in position, or displacement, can be translational (linear) or rotational (angular), with the movement of the human body generally being translational and the movement of the limbs rotational about their respective joints.[182] The magnitude and direction of linear and angular displacements can be calculated using geometry and trigonometry.

Marker-based 3D motion analysis is commonly used in research to calculate trunk, hip, knee, and ankle joint angles.[36,184-187] The position and trajectories of retroreflective markers placed on anatomical landmarks are able to be accurately and reliably recorded with these systems.[188-190] Each marker must be visible by at least two cameras and the 3D data is reconstructed for each marker from the 2D trajectories. Three markers are needed to reconstruct a segment and segment rotations and angles are calculated using Euler angles in a limb rotation algorithm.[187,191]

There are limitations to marker-based motion analysis. Measurement error can be introduced by skin artifact and movement variability; however, such errors are repeatable and systematic.[188] Skin artifact may be compounded during dynamic tasks performed during testing. In the current study, measurement error will be controlled for by placement of the retroreflective markers so as to minimize marker movement caused by soft tissue movement. In addition, incorrect marker placement and variability in marker replacement can induce error; therefore a single experienced researcher will place the markers on all subjects. Since testing will be conducted in a single session, marker replacement error should not be an issue. This instrumentation has been reported to have good-to-excellent reliability for kinematic variables collected during dynamic activities (within session ICC: 0.933- 0.993; between session ICC: 0.595- 0.922).[189]

2.5.3 Selection of Knee and Trunk Kinematics

Marker-based 3D motion analysis allows for the description of movement about three axes (x, y, and z). For the knee, movement can be described in terms of flexion/extension, internal/external rotation, and valgus/varus. For the trunk, the movement can be described in terms of flexion/extension, right/left rotation, and lateral flexion. However, in the current study the kinematic variables of interest are: knee flexion/extension, knee valgus/varus, trunk flexion/extension, and trunk lateral flexion.

Prospective research has demonstrated that individuals who sustain ACL injury demonstrated significantly greater knee valgus at initial contact and greater knee valgus motion as well as less maximum knee flexion than those who remained injury-free.[36] In addition, knee valgus angle at initial contact and maximum were found to be significant predictors of ACL injury. Similarly, ACL-R athletes who sustained a second ACL injury demonstrated significantly greater knee valgus motion than those who remained injury-free.[37] The relationship between ACL injury and knee valgus is further supported by video analysis of ACL injury events.[38,40-42,44] Although knee flexion angle at initial contact or maximum knee flexion has not been demonstrated to be a predictor of ACL injury, video analysis of ACL injury events has shown that the knee is in a relatively extended position at initial contact and/ or time of injury.[38,42]

Trunk kinematics relative to ACL/lower extremity injury or knee kinematics has not been studied extensively. Relative to ACL injury risk or prevention, it has been suggested that neuromuscular control of the trunk should be considered.[12,51] Prospective research has demonstrated that trunk displacement after a sudden release was significantly greater in athletes who sustained knee, ligament, and ACL injury as compared to uninjured athletes.[52] Video analysis of ACL injury events has shown that females who sustained ACL injury demonstrated significantly less trunk flexion than female controls as well as significantly greater lateral trunk displacement than males who sustained an ACL injury.[41] Further, one study has demonstrated that increasing trunk flexion during a double leg drop landing resulted in significant increases in knee flexion at initial contact.[94]

2.5.4 Measurement of Trunk Proprioception

A variety of methods have been developed to evaluate the submodalities of conscious proprioception. Joint position sense can be evaluated by tasks in any combination of active and passive movements where the intent is to accurately reposition, identify, or replicate limb position.[55,60,64,65] In addition, these tasks can be performed as both open and closed kinetics chain tasks.[58] Joint position sense can be assessed using goniometers, video, or motion capture systems. Kinesthesia, or movement sense, refers to the capability to detect passive movement of a joint or segment.[60,64-66] Kinesthesia can be assessed using isokinetic dynamometers or custom jigs that are capable of moving at slow speeds. Finally, force sense signifies the proficiency of a muscle group to reproduce a target torque or to identify a difference in weight (weight discrimination).[58,59] Force sense can be assessed using isokinetic dynamometers.

Trunk proprioception has been assessed in previous research using joint position sense,[149,192-196] threshold to detect passive motion (kinesthesia),[51,149,192,194,197-199] and force reproduction.[149,193] While previous methods have utilized an isokinetic dynamometer or custom-built devices to assess all submodalities of proprioception with good to excellent reliability (ICC: 0.470-0.904)[51,149,192,199] and precision (SEM: 0.194Nm, 0.047-0.73°),[149,192] there are limitations in the use of this instrumentation. The mechanoreceptors of interest in proprioception testing are found in musculotendinous and capsuloligamentous tissues; therefore stimulation of cutaneous mechanoreceptors by the stabilization straps prevents isolation of the mechanoreceptors of interest. In addition, the testing position is in a non-functional (e.g., seated or semi-standing) position. Previous research has demonstrated that trunk position sense is significantly affected by position, with significantly better trunk position sense in standing.[200] It has been postulated that this may be related to preferential activation of mechanoreceptors in weight-bearing structures while in a standing position.[200]

Trunk proprioception also has been assessed using electromagnetic tracking devices.[193,195,200,201] However, poor to good reliability (ICC: 0.06-0.50) with good precision (SEM: 0.15-0.97°) have been reported when using this instrumentation with the pelvis fixed.[202] Although stabilization of the pelvis may better isolate the trunk musculature, stabilization of the pelvis results in a non-functional movement pattern and provides sensory cuing through cutaneous stimulation which may influence proprioceptive ability.

Passive video-based motion capture systems have demonstrated good to excellent reliability (ICC: 0.54-0.99) and precision (SEM: 0.9-2.7°) when used to assess joint position sense.[203] Similar reliability also has been reported for kinematic variables collected during dynamic activities utilizing these systems (within session ICC: 0.933- 0.993; between session ICC: 0.595- 0.922).[189] In the current study, a passive motion capture system will be used to assess active-active joint position sense into flexion and lateral flexion. Subjects will be tested in a standing position and without pelvic stabilization in order to simulate a more functional position and minimize cutaneous stimulation.

2.5.5 Measurement of Isokinetic Trunk Strength

Isokinetic strength assessment is commonly used in the research setting to quantify muscular strength or the ability of a muscle/group of muscles to produce force or torque. Isokinetic movement occurs at the same preset velocity throughout the range of motion, with the dynamometer adjusting its equaling counterforce.[204] Advantages of isokinetic evaluation include isolation of specific muscle groups and accommodating resistance (safer and allows for maximum force to be generated throughout range of motion) as well as providing a quantifiable measure of muscle performance (torque, work, power).[204] Disadvantages of isokinetic evaluation include that assessment can only occur in the cardinal plans of motion and are typically non-weighting, open-kinetic chain.[204]

Isokinetic strength assessment can be performed concentrically or eccentrically as well as at a variety of different velocities. In the current study, isokinetic testing of the trunk musculature will be performed as concentric-concentric reciprocal contractions so as to minimize the risk of delayed-onset muscle soreness that has been reported to occur with eccentric exercise.[205] In addition, a slower test velocity (60°/s) will be used because concentric force generation is greater at slow isokinetic velocities.[204] This testing speed has been utilized in previous research and has demonstrated good to excellent reliability (ICC: 0.74-0.98).[206,207] [208,209] These factors will enable the isokinetic strength data to be collected using reliable methodology as well as all for comparison of results in the current study to be compared to those published previously.

2.6 DEVELOPMENT OF REGRESSION EQUATIONS

It is believed that trunk strength, proprioception, and kinematics as well as sex will be predictive of knee kinematics. However, based on the timing (e.g., initial contact, maximum) and the plane of motion (e.g., frontal vs. sagittal plane) related to the knee joint angle, different independent variables are hypothesized to play a greater role for each of the dependent variables. Dependent and independent variables are listed in Table 4. Each of the dependent variables with the corresponding hypothesized independent variables and rationale are presented in Appendix A.

For trunk strength, the independent variables are trunk extension, right rotation, and left rotation. Trunk extension strength is hypothesized to be a predictor of knee valgus angle at initial contact as well as knee flexion angle at initial contact and maximum knee flexion angle during the SJCM. It is believed that trunk extensor strength is critical in decelerating the trunk segment center of mass at initial contact as well as during the entire landing phase. For knee valgus angle at initial contact, this relationship is an indirect one; if the trunk is not decelerated adequately, then there will be less knee flexion at initial contact which may result in increased knee valgus in an attempt to attenuate landing forces. An inability to decelerate the trunk segment center of mass at initial contact and throughout the landing phase could result in reduced knee flexion angle at initial contact and maximum knee flexion angle. Trunk rotation strength towards the direction of the cutting maneuver (e.g., left trunk rotation strength for a right leg dominant subject-subject will plant on right leg and cut to the left) is hypothesized to be a predictor for total knee valgus excursion during the SJCM. As a subject plants and cuts away from the dominant leg, the trunk musculature must work to rotate the trunk into or towards the direction of the cutting maneuver in order to prevent the trunk from “trailing” the motion, which would result in relative internal rotation and dynamic valgus collapse of the knee.

Table 4. Dependent and Independent Variables

Dependent Variables	Independent Variables
Knee valgus angle at initial contact during the SJCM Total knee valgus excursion during the SJCM Knee flexion angle at initial contact during the SJCM Maximum knee flexion angle during the SJCM	Strength (peak torque in Nm) • Isokinetic trunk flexion strength • Isokinetic right trunk rotation strength • Isokinetic left trunk rotation strength Proprioception (absolute error in degrees) • Trunk flexion active joint position sense • Right lateral flexion active joint position sense • Left lateral flexion active joint position sense Kinematics (degrees) • Lateral trunk displacement angle at initial contact during the SJCM • Maximum lateral trunk displacement angle during the SJCM • Trunk flexion angle at initial contact during the SJCM • Maximum trunk flexion angle during the SJCM Sex • Male, female

For trunk proprioception, the independent variables are trunk flexion, right lateral flexion, and left lateral flexion. For trunk kinematics, the independent variables are trunk flexion and lateral trunk flexion angles, both at initial contact with the force plate and at maximum displacement of the trunk. Trunk flexion proprioception as well as trunk flexion angle at initial contact and maximum displacement are hypothesized to be predictors of knee flexion angle at initial contact and maximum during the SJCM, respectively. The ability to correctly position the trunk in the sagittal plane relative to the base of support will influence knee sagittal plane motion. Previous research has demonstrated that an increasing trunk flexion angle during landing results in a concomitant increase in knee flexion angle.[94] Trunk lateral flexion proprioception towards the dominant leg as well as lateral trunk angle at initial contact and maximum are hypothesized to be predictors of knee valgus angle at initial contact and total knee valgus excursion during the SJCM, respectively. The ability to correctly position the trunk in the frontal plane relative to the base of support will influence knee frontal plane motion.[12,45] When the trunk is displaced laterally relative to the dominant leg during the single leg landing, the vertical ground reaction force will pass lateral to the knee joint center, resulting in increased knee valgus. Deficits in proprioceptive ability in the sagittal and frontal planes may result in overcorrection or undercorrection of trunk position relative to the base of support, thereby resulting in decreased knee flexion angle or deviation from a neutral knee valgus/varus angle, respectively.

3.0 MATERIALS AND METHODS

3.1 EXPERIMENTAL DESIGN

The purpose of this cross-sectional study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex predict knee kinematics during a stop jump-cut maneuver (SJCM). Linear regression was utilized to ascertain the ability of the aforementioned variables to predict knee valgus angle at initial contact and total knee valgus excursion as well as knee flexion angle at initial contact and maximum knee flexion angle during a SJCM. Dependent and independent variables were as follows:

3.1.1 Dependent Variables

- Knee valgus angle at initial contact during the SJCM
- Total knee valgus excursion during the SJCM
- Knee flexion angle at initial contact during the SJCM
- Maximum knee flexion angle during the SJCM

3.1.2 Independent Variables

- Isokinetic trunk extension, and right/left trunk rotation strength (average peak torque in Nm)
- Trunk flexion and right/left lateral flexion proprioception assessed with active joint position sense for trunk repositioning error (degrees)
- Lateral trunk displacement angle at initial contact during the SJCM (degrees)
- Maximum lateral trunk displacement angle during the SJCM (degrees)
- Trunk flexion angle at initial contact during the SJCM (degrees)
- Maximum trunk flexion angle during the SJCM (degrees)
- Sex

3.2 SUBJECTS

Based on the power analysis, 53 subjects were required for the current study (R^2=0.20, four predictors in the final model, power ≥0.80, and two-sided alpha=0.05); however, assuming attrition of 10%, up to 59 subjects may be enrolled in this study. All subjects were healthy adults between the ages of 18-25 years, inclusive, and met the criteria to be classified with a physical activity level of I or II on the Noyes Sports-Activities Rating Scale (Appendix B).[210] Each subject was informed of the methods as well as the risks and benefits associated with the study, after which written informed consent, as approved by the University of Pittsburgh Institutional Review Board, was obtained. Testing was conducted in a single session, lasting approximately 90 minutes, at the Neuromuscular Research Laboratory.

3.2.1 Inclusion Criteria

- Male or female between the ages of 18-25 years, inclusive
- Physically activity level of I or II based on the Noyes Sports-Activities Rating Scale
  - Participates 4-7 days (Level I) or 1-3 days (Level II) per week in:
    - Jumping, hard pivoting, cutting (e.g., basketball, soccer)
    - Running, twisting, turning (e.g., tennis, field hockey)
    - No running, twisting, jumping (e.g., cycling, swimming)
- No previous history of low back pain or low back injury that limited activities of daily living or athletic activities for greater than 1 week

3.2.2 Exclusion Criteria

- Current low back, hip, knee, or ankle injury
- Musculoskeletal injury to the lower extremity within the previous 6 months that limited activities of daily living or athletic activities for greater than 1 week
- Previous history of knee ligament injury
- Previous surgery to the lower extremity or low back
- Any disorder that could affect equilibrium or neuromuscular control
- Allergy to adhesives or adhesive tape

3.3 POWER ANALYSIS

A power analysis was performed using IBM® SPSS® SamplePower (International Business Machine Corp., Armonk, NY). Assuming medium to large effect size (R^2=0.20) and that four predictors would be in the final model, a total of 53 subjects were needed to reach a power of at least 0.80 at a two-sided alpha=0.05. To account for 10% attrition, up to 6 additional subjects may be enrolled, increasing the total sample size to 59.

3.4 SUBJECT RECRUITMENT

Subjects were recruited from the local community, including colleges/universities and health/fitness clubs, through the use of posted flyers. Potential subjects contacted the primary investigator at the Neuromuscular Research Laboratory and underwent a phone screen to determine eligibility. Subjects who were eligible and interested in participating were scheduled for a single test session.

3.5 INSTRUMENTATION

3.5.1 Isokinetic Dynamometer

Trunk extension and right/left rotation strength was assessed with the Biodex III Multi-Joint System Pro (Biodex Medical Inc., Shirley, NY). The Biodex was calibrated according to the manufacturer’s instructions prior to data collection. The Biodex Advantage software v.4.2

(Biodex Medical Inc., Shirley, NY) automatically adjusts the torque. This instrumentation has been reported to have excellent reliability and precision for both trial-to-trial and day-to-day for position (ICC > 0.99, SEM: 0.45-0.60°) as well as for trial-to-trial for torque (ICC > 0.99, SEM: 0.00-0.39Nm).[211]

3.5.2 Video Motion Analysis System

Proprioception of the trunk (active joint position sense for trunk flexion and right/left lateral flexion) as well as biomechanical analyses of the dynamic laboratory task (SJCM) were assessed using a passive video-based motion capture system and Vicon Nexus software (Vicon, Centennial, CO). This system utilizes eight high-speed cameras equipped with infra-red light-emitting-diodes (LEDs), which are reflected off of reflective markers placed on specific anatomic landmarks on a subject. The cameras capture the 2D trajectories of the markers and, using a standard calibration procedure, the 3-D coordinates of each marker can be calculated. Calibration was performed according to the manufacturer using the wand method. Six cameras were mounted on the walls surrounding the capture area and two were positioned on tripods to ensure that each marker can be seen by a minimum of two cameras. Camera data were collected at 200Hz during all tasks. This instrumentation in our laboratory has been determined to be accurate for both position and angular data, with a root mean square error of 0.002m and 0.254°, respectively. Previous authors have reported that optimal combination of camera positioning, calibration, marker size, and lens filter resulted in an overall accuracy of 63±5µm and overall precision of 15µm.[212]

3.5.3 Force Plates

Two force plates (Kistler 9286A, Kistler Instrument Corp., Amherst, NY) were used to identify initial contact and end of contact during the dynamic laboratory task (SJCM). Ground reaction force data were sampled at 1200Hz. Initial contact was identified as the time at which the vertical ground reaction force exceeded 5% of the subject's body weight. End of contact was identified as the time at which the vertical ground reaction force fell below 5% of the subject's body weight following initial contact.

3.6 TESTING PROCEDURES

3.6.1 Subject Preparation

Written, informed consent, as approved by the University of Pittsburgh Institutional Review Board, was obtained prior to data collection. All data collection took place at the Neuromuscular Research Laboratory of the University of Pittsburgh. The following anthropometric data was collected for the kinematic analysis: body mass (kg), body height (mm), leg length (mm), and ankle and knee joint width (mm). Measurements were taken bilaterally as applicable.

3.6.2 Order of Testing

Trunk proprioception testing (trunk flexion and right/left lateral flexion) was performed first in order to minimize the risk of physical and/or mental fatigue influencing the results. Since the subjects already had the motion analysis markers in place from trunk proprioception testing, biomechanical analyses of the SJCM were tested second. Trunk strength testing (extension and right/left rotation) was performed last and the two potential strength testing sequences were equally utilized. A five minute rest period was provided between tests in order to minimize the effects of fatigue.

3.6.3 Trunk Proprioception Testing

Trunk proprioception testing was a modification of the methods developed by Tsai et al. (Figures 1 and 2).[202] In the current study, a passive video-based motion capture system and Vicon Nexus software were used rather than an electromagnetic tracking device. Custom movement guides were used for the practice trials but removed (abdominal portion only) during the test trials in order to minimize cutaneous cues. The order of testing was randomized.

The anthropometric measures listed previously were entered into the Vicon Nexus software. Passive reflective markers were placed on the following anatomical landmarks, bilaterally as applicable, according to the Vicon Plug-in-Gait model (Vicon, Centennial, CO) using double-sided adhesive tape:

- 2^{nd} metatarsal head (dorsal aspect)
- Lateral malleolus (distal tip)
- Posterior calcaneus (at the level of the 2^{nd} metatarsal marker)
- Lateral aspect of lower leg (midpoint between lateral malleolus and lateral femoral condyle markers)
- Lateral femoral epicondyle

- Lateral aspect of the upper leg (midpoint between lateral femoral condyle marker and the greater trochanter of the femur)
- Anterior superior iliac spine
- Posterior superior iliac spine
- Spinous process of 10th thoracic vertebrae
- Spinous process of the 7th cervical vertebrae
- Xiphoid process
- Jugular notch

A static calibration trial was collected following placement of the markers. Subjects were instructed to stand with the feet directly under the hips, toes pointed forward, and the arms abducted to 90° and to remain as still as possible during this static trial. The static calibration trial provided the neutral or 0° joint angle position from which the joint angles were calculated during trunk proprioception and the SJCM task.

3.6.3.1 Trunk Flexion Active Joint Position Sense

The subject was instructed to stand upright with equal weight on both feet near the center of the capture area. A custom guide was placed in front of the subject and the footprint of the guide was marked for accurate replacement during testing. The subject's foot placement also was marked in case the subject moved during testing. The subject forward flexed to 20° of trunk flexion as measured with a goniometer. The guide then was raised so that the guide was in contact with the anterior aspect of the thighs and the anterior abdomen (Figure 1). The subject was instructed to keep his/her feet in the same position for the remainder of the test. The subject then returned to the start position of his/her typical upright posture. The subject then placed a blindfold over the eyes in order to minimize visual cuing. Methods for data collection were as follows:

- Subject forward flexed until the anterior abdomen contacted the guide
- Subject was instructed to concentrate on and remember this position (target position held for five seconds)
- Subject was instructed to depress the trigger, capturing the reference trial
- Subject returned to the starting position of typical upright posture
- The guide was removed (abdominal portion only)
- Subject attempted to replicate the target position, depressing the trigger switch when he/she believed that the target position had been reached, recording the replicated position

This process was repeated a total of five times and the difference between the target position and the replicated position was calculated for each trial. Five trials were performed in order to account for potential data loss and the average absolute difference of the first three good trials was calculated and used in data analyses.

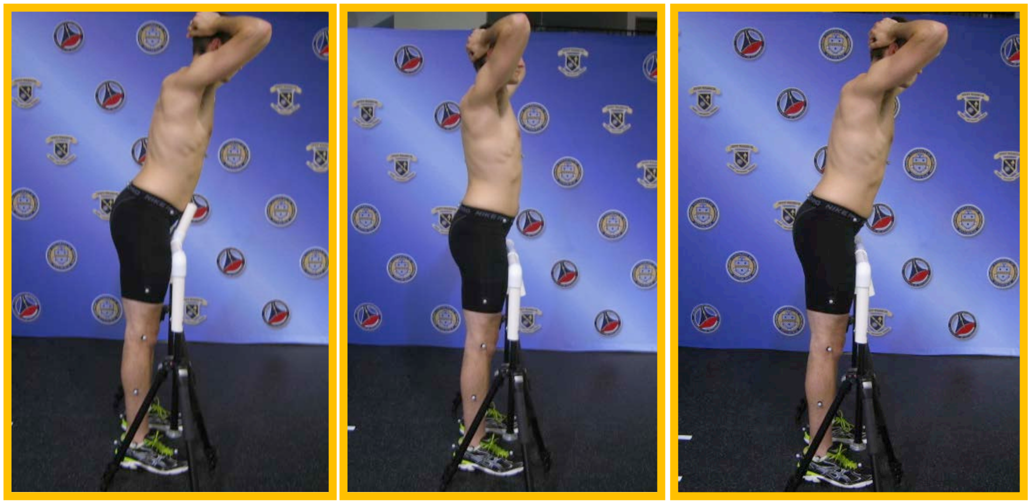


Figure 1. Active Joint Position Sense Assessment for Flexion: Reference target position (left), Neutral position (center), and Replicated target position (right)

3.6.3.2 Trunk Right/Left Lateral Flexion Active Joint Position Sense

The subject was instructed to stand upright with equal weight on both feet near the center of the capture area. A custom guide was placed to the right of the subject and the footprint of the guide was marked for accurate replacement during testing. The subject placed the right hand on top of the head and laterally flexed to the right to 15° of trunk lateral flexion as measured with a goniometer (Figure 2). The guide was positioned so that the guide was in contact with the lateral aspect of the thighs and torso. The subject was instructed to keep his/her feet in the same position for the remainder of the test. The subject then returned to the start position of his/her typical upright posture. The subject then placed a blindfold over the eyes in order to minimize visual cuing. Methods for data collection were as follows:

- Subject laterally flexed until contact is made with the guide
- Subject was instructed to concentrate on and remember this position (target position held for five seconds)

- Subject was instructed to depress the trigger, capturing the reference trial
- Subject returned to the starting position of typical upright posture
- The guide was removed (abdominal portion only)
- Subject attempted to replicate the target position, depressing the trigger switch when he/she believed that the target position has been reached, recording the replicated position

This process was repeated a total of five times and the difference between the target position and the replicated position was calculated for each trial. Five trials were performed in order to account for potential data loss and the average absolute difference of the first three good trials was calculated and used in data analyses. The subject then repeated this test for left lateral flexion using the same procedures.

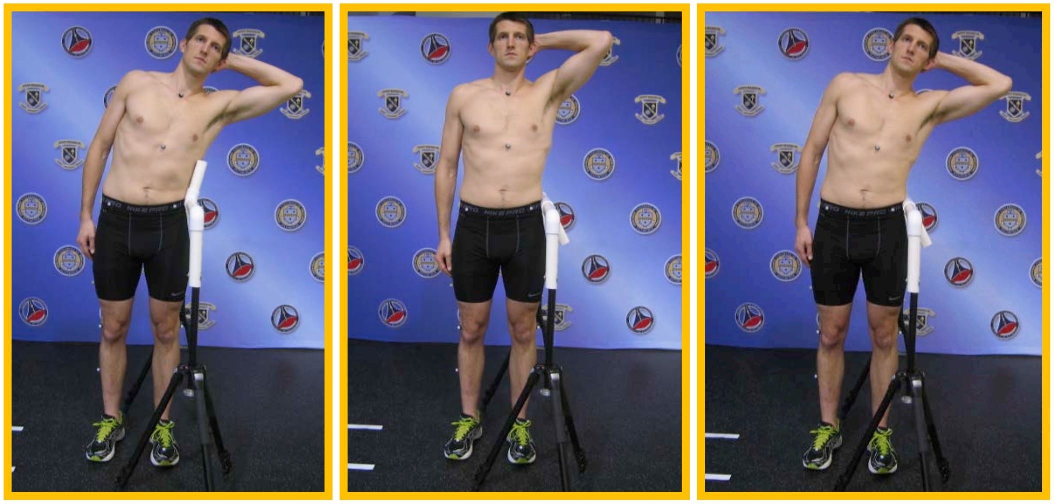


Figure 2. Active Joint Position Sense Assessment for Flexion: Reference target position (left), Neutral position (center), and Replicated target position (right)

3.6.4 Biomechanical Assessment

A variety of tasks such as run-to-cut or side-step cutting maneuvers,[8,176] stop jumps,[177,178] and drop landings[37,49,105,179] have been used previously in ACL injury research. A stop jump-cut maneuver involves jumping forward off of two feet, landing on a single leg, and immediately cutting away from the stance leg. Video analysis of ACL injury indicate that typically these injuries occur during plant-and-cut maneuvers[38,40] or single-leg landings.[38] The SJCM incorporates both of these and, therefore, was used in the current study. Biomechanical assessment of the lower extremity included a SJCM task (Figure 3).[96] The start position of 40% of the subject's height was marked on the floor from the edge of the force plates. Verbal instructions and physical demonstration were provided prior to practice trials; however no instructions were given as to landing technique. The test was performed on the dominant foot, which was operationally defined as the foot with which the subject would use to maximally kick a ball. The subject squatted to approximately 45° of knee flexion ("athletic ready" position), held this position for 4 seconds, and then jumped forward with both legs. The subject landed with the foot of the dominant leg on the force plate, immediately performed a 45° cut away from the dominant leg, and ran past a cone that was 2.5 meters away. Each subject was allowed a minimum of three practice trials to become familiar with the task. After a one minute rest period, the subject performed five test trials. Test trials were discarded and repeated if any of the following occurred: appeared not to initiate the jump with both legs, failed to reach the force plate target area, failed to immediately perform cutting maneuver after landing, or if the non-dominant leg touched the stance leg or the ground.

Figure 3. Stop Jump-Cut Maneuver

3.6.5 Trunk Muscular Strength Testing

Trunk extension and right/left rotation strength were assessed using isokinetic dynamometry (Figures 4and 5). Testing was performed using concentric/concentric reciprocal contractions at 60°/s for all practice and test trials. Although this protocol required that both trunk flexion and trunk extension strength be assessed, only trunk extension strength was of interest in the current study. Consistent verbal cuing was given for all subjects. The average peak torque across the five test trials for all isokinetic strength tests was recorded for data analysis.

3.6.5.1 Trunk Extension

For trunk extension strength testing, the subject was seated in a semi-standing position in the Biodex trunk flexion/extension attachment. The footrest was adjusted so that the knees were in a semi-flexed position and the posterior thighs were supported on the seat of the chair. In addition, the axis of rotation was aligned as per the manufacturer's instructions. The subject was stabilized with two straps across the anterior thighs and straps over the torso. The range of motion limits were set according to each subject's range of motion. Following verbal instructions, the subject performed three submaximal (50% of perceived maximum effort) reciprocal concentric isokinetic extension/flexion repetitions (starting from a flexed position) followed by three maximal effort (100% maximum effort) repetitions for familiarization purposes and to ensure proper stabilization. After a one minute rest period, the subject performed five maximal effort extension/flexion repetitions for the actual test trial. Good-to-excellent reliability has been reported previously using similar methodology and instrumentation (ICC: 0.74-0.98).[206,207]

3.6.5.2 Trunk Right/Left Rotation

For trunk right/left rotation strength testing, the range of motion limits were set at the end ranges of the device. The subject was seated in the Biodex chair and the trunk rotation attachment lowered, placed across the upper torso, and secured with stabilization straps. The hip pads were adjusted so that the subject was seated directly under the axis of rotation of the trunk rotation attachment. Additional stabilization was provided with thigh stabilization pads. Following verbal instructions, the subject performed three submaximal (50% of perceived maximal effort) reciprocal concentric isokinetic right/left trunk rotation repetitions (starting form a position of left rotation) followed by three maximal effort (100% maximal effort) repetitions for familiarization purposes and to ensure proper stabilization. After a one minute rest period, the subject performed five maximal effort right/left trunk rotation repetitions for the actual test trial. This test has demonstrated excellent test-retest reliability in our laboratory using this protocol (ICC=0.890 and SEM=13.5%BW; ICC=0.905 and SEM=12.4%BW for the right and left sides, respectively).[208,209]

Figure 4. Isokinetic Strength Testing Trunk Flexion/Extension

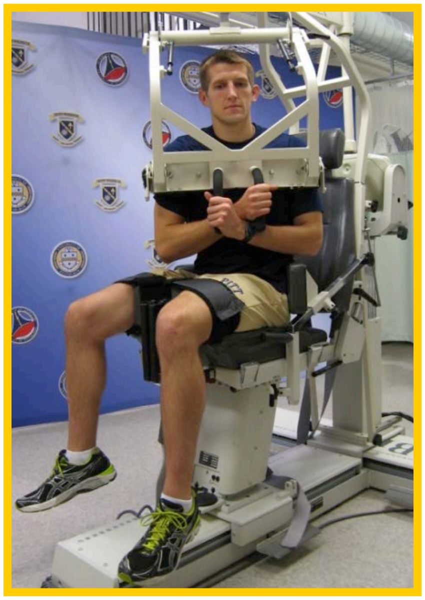


Figure 5. Isokinetic Strength Testing: Trunk Rotation

3.7 DATA ANALYSIS

3.7.1 Data Reduction

3.7.1.1 Trunk Proprioception and Biomechanical Assessment

Trunk and lower extremity kinematics were calculated for the trunk proprioception and SJCM tasks. Kinematic calculations were based on 3-D coordinates of reflective markers, anthropometric measurements, and a subject-specific biomechanical model (Plug-in Gait, Vicon, Centennial, CO). Raw 3-D coordinate data were filtered using a general cross-validation Woltring filter and kinematic calculations (limb rotation and joint angles) were performed in the Vicon Nexus software (Vicon Motion Systems, Inc., Centennial, CO).[213] Orthogonal embedded coordinate systems were calculated for each segment (thorax, pelvis, thigh, shank, and foot), with at least three markers defining a segment. Knee and ankle joint centers were calculated using the respective joint width measurements obtained prior to data collection and the associated embedded coordinate system.[191] The hip joint centers were calculated using Newington-Gage model, which uses the mean distance between the ASIS markers and the distance between the each ASIS and the ipsilateral trochanter (estimated based on leg length measurements).[191] Joint center-based embedded coordinate systems were calculated for each joint. The embedded coordinate systems of the segments then were realigned with those of the joints utilizing the angular offset values from the static trial obtained at the beginning of data collection. Three dimensional angles of rotation for each segment then were calculated based on the realigned embedded coordinate systems by defining the orientation of the distal coordinate system axes relative to the proximal coordinate system axes.[191] Angles for all joints were calculated following a YXZ Euler rotation sequence.[187,190,191] A custom Matlab® (Mathworks, Natick, MA) script was used to identify the kinematic variables of interest described below.

For trunk proprioception, the target position and the replicated position for each motion (trunk flexion, and right and left lateral flexion) were determined as the position when the trigger was depressed by the subject during data collection. The difference between the target position and replicated position was calculated for each of the five trials. Differences across the first three good trials were averaged (average absolute error) and used in data analyses.

Initial contact during the SJCM was identified using ground reaction force data from the force plate. Ground reaction force data were filtered using a zero-lag fourth order low-pass Butterworth filter with a cutoff frequency of 20 Hz. Initial contact was identified as the point when the filtered vertical ground reaction force exceeded 5% of the subject's body weight. This point was used to determine knee flexion, knee valgus, trunk flexion, and trunk lateral flexion at initial contact. In addition, the maximum angle during the stance phase of landing was identified for knee flexion, knee valgus, trunk flexion, and trunk lateral flexion. The stance phase was defined as the time from initial contact until the vertical ground reaction force fell below 5% of the subject's body weight. The orientation of the pelvis relative to the global coordinate system during the squat phase of the stop jump-cut maneuver, at initial contact, and upon completing the cutting maneuver as well as the orientation of the landing foot during the squat phase of the stop jump-cut maneuver and at initial contact were determined using a custom Matlab® (Mathworks, Natick, MA) script for trials in which it appeared that a subject had turned prior to initial contact (±10°) or had not completed a 45° cut (±10°). Total knee valgus excursion was calculated as the difference between maximum knee valgus angle and knee valgus angle at initial contact.

3.7.1.2 Trunk Muscular Strength

Average peak torque (Nm) for trunk extension and right and left rotation was obtained from the Biodex Advantage software v.3.2.

3.7.2 Statistical Analysis

Statistical analyses were performed using Stata (Stata 12, StataCorp LP, College Station, TX). Separate multiple linear regression equations were fit for each of the dependent variables. Subject matter knowledge was incorporated in the model building process. All variables were examined individually. Summary statistics were computed and graphs plotted. Outliers were identified. Data transformations were performed if required. Pairwise scatter plots were created and examined for each variable. Correlation coefficients and collinearity diagnostics were calculated and redundant variables were considered for deletion. The full model was fit and non-significant predictors were deleted. The reduced model was fit. Residuals were examined for linearity; heteroscedasticity; and outliers, high leverage points, and influential points. Any issues identified in this step were fixed. Analysis was conducted to examine if additional variables could be dropped, and if new variables could be included in the model. If variables were dropped or added, then the model was fit and the steps outlined above were repeated (e.g. non-significant predictors deleted, residuals re-examined). Information criteria were used to monitor the fitting process. For the final model, variance inflation factors (VIFs) and residual diagnostics were checked. If needed, analysis was re-conducted to examine if additional variables could be dropped, and if new variables could be included in the model.[214] Statistical significance levels of 0.05 were established *a priori*.

4.0 RESULTS

The purpose of this study was to examine the relationship between knee kinematics and neuromuscular characteristics of trunk functional stability as well as trunk kinematics in a healthy, physically active population. Trunk and knee kinematics were assessed during a stop jump-cut maneuver (SJCM) using passive video-based motion analysis. The dependent variables were knee valgus angle at initial contact (Hypothesis 1a), total knee valgus excursion (Hypothesis 1b), knee flexion angle at initial contact (Hypothesis 2a), and maximum knee flexion angle (Hypothesis 2b). Trunk muscular strength and proprioception were assessed with isokinetic dynamometry and active joint position sense, respectively. Independent variables included peak torque for trunk extension and trunk rotation (non-dominant side or rotation away from the dominant leg); trunk flexion and dominant side lateral flexion (lateral flexion towards the dominant leg) average absolute error during the active joint position sense tasks; trunk flexion and lateral trunk displacement angle at both initial contact and maximum displacement during the SJCM; and sex. Descriptions of the dependent and independent variables are presented in Table 5. Univariate statistics are presented first. Bivariate statistics then are presented in order to examine the relationship between each dependent variable and a single independent variable. Lastly, the results of the multiple liner regression are detailed. Subject matter knowledge was incorporated in the model building process (Appendix A). Statistical significance levels of 0.05 were established *a priori*.

Table 5. Description of Dependent and Independent Variables

		Description
Dependent Variables	Knee valgus angle at initial contact (°)	Knee valgus angle when the vGRF exceeds 5% of a subject's BW
	Total knee valgus excursion (°)	Total knee valgus movement from when the vGRF exceeds 5% of a subject's BW until the vGRF falls below 5% of the subject's BW
	Knee flexion angle at initial contact (°)	Knee flexion angle when the vGRF exceeds 5% of a subject's BW
	Maximum knee flexion angle (°)	Maximum knee flexion angle occurring between when the vGRF exceeds 5% of a subject's BW until the vGRF falls below 5% of the subject's BW
Proprioception	Forward flexion average absolute error(°)	Absolute average repositioning error of the thorax during forward flexion
	Dominant side lateral flexion average absolute error(°)	Absolute average repositioning error of the thorax during lateral flexion towards the dominant side
Trunk Kinematics	Forward flexion at initial contact (°)	Forward flexion angle of the spine when the vGRF exceeds 5% of a subject's BW
	Maximum forward flexion angle (°)	Maximum forward flexion angle of the spine occurring between when the vGRF exceeds 5% of a subject's BW until the vGRF falls below 5% of the subject's BW
	Lateral flexion at initial contact (°)	Lateral flexion angle of the spine when the vGRF exceeds 5% of a subject's BW
	Maximum lateral flexion angle (°)	Maximum lateral flexion angle of the spine occurring between when the vGRF exceeds 5% of a subject's BW until the vGRF falls below 5% of the subject's BW
Trunk Strength	Trunk extension average peak torque(Nm)	Average peak torque of the trunk extensors
	Non-dominant trunk rotation average peak torque (Nm)	Average peak torque of trunk rotators towards the non-dominant side

4.1 SUBJECT CHARACTERISTICS

A total of 53 subjects between the ages of 18 and 25, inclusive, were enrolled in this study. All subjects were currently participating in physical activity, with a physical activity level of I or II on the Noyes Sports-Activities Rating Scale. In addition, subjects had no history of: knee ligament injury and low back or lower extremity surgery; low back pain/injury that limited activities of daily living/sports for more than one week; and current injury to the low back, hip, knee, or ankle. Subjects also were free of lower extremity musculoskeletal injury for the six months prior to testing. All subjects completed all testing procedures and complete data sets are present for each. Subject demographics are presented in Table 6.

Table 6. Subject Demographics

					Interquartile Range			Range		
	Mean	±	SD	Median	(25th	,	75th)	(Min	,	Max)
Age (yrs)	22.0	±	2.1	22.3	19.9	,	23.9	18.5	,	25.9
Height (cm)	172.5	±	8.4	172.1	168.3	,	178.1	155.2	,	193.3
Mass (kg)	71.6	±	10.4	70.4	62.0	,	78.5	54.4	,	94.5
Noyes Sports-Activity Rating	87.7	±	7.6	85.0	85.0	,	90.0	75.0	,	100.0

4.2 UNIVARIATE ANALYSIS

4.2.1 Dependent Variables

The four dependent variables were knee valgus angle at initial contact (Hypothesis 1a), total knee valgus excursion (Hypothesis 1b), knee flexion angle at initial contact (Hypothesis 2a), and maximum knee flexion angle (Hypothesis 2b). Descriptive statistics for the dependent variables are presented in Table 7. Mean normalized knee flexion and mean normalized knee varus/valgus angle across all subjects with standard deviation relative to stance phase are presented in Figures 6 and 7, respectively.

Table 7. Dependent Variables: Knee Kinematics Mean and Standard Deviation

					Interquartile Range			Range		
	Mean	±	SD	Median	(25th	,	75th)	(Min	,	Max)
Knee valgus angle at initial contact (°)	-0.94	±	5.73	-0.62	-4.88	,	3.20	-14.06	,	12.21
Total knee valgus excursion (°)	6.38	±	5.28	5.02	1.99	,	9.71	0.00	,	19.89
Knee flexion angle at initial contact (°)	25.50	±	8.90	25.24	18.80	,	30.54	9.56	,	46.00
Maximum knee flexion angle (°)	61.22	±	7.27	61.82	55.48	,	66.87	45.58	,	75.71

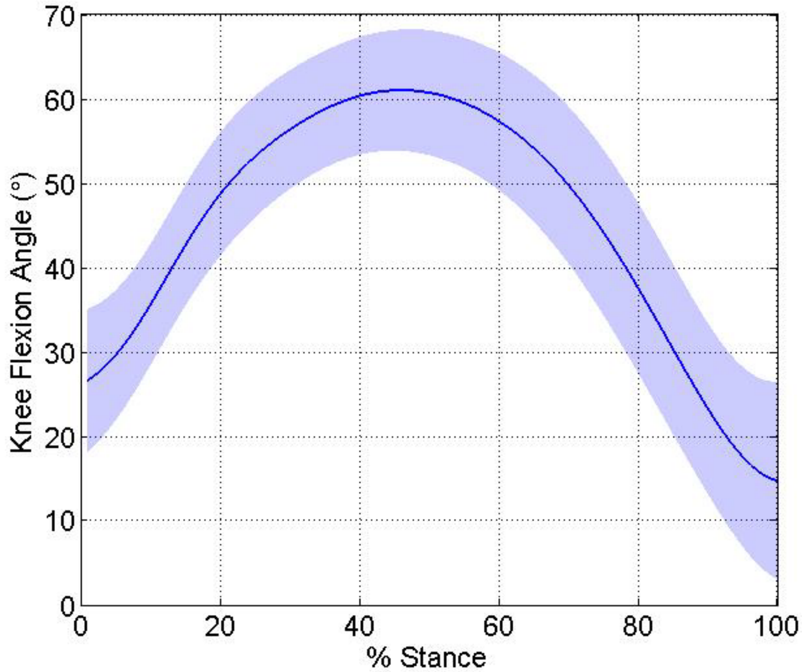


Figure 6. Mean Normalized Knee Flexion Angle (+/- SD) Relative to Stance across All Subjects

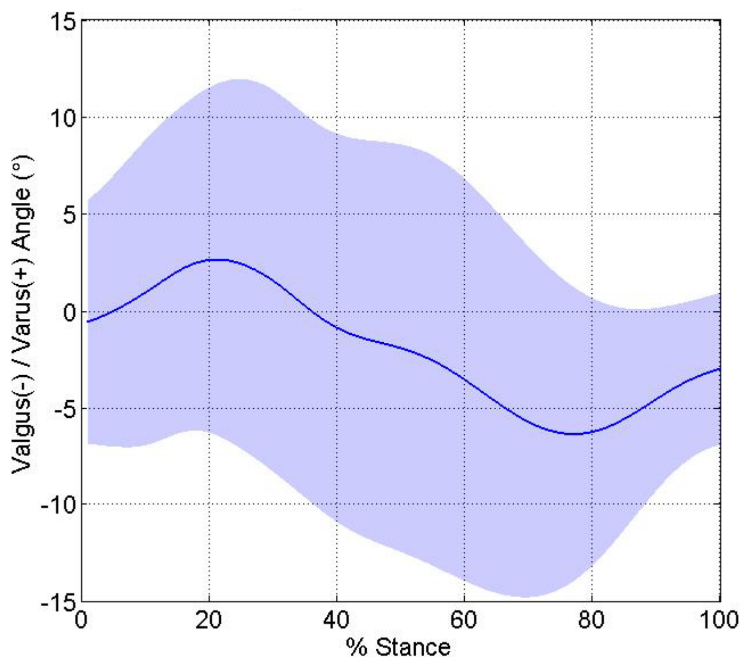


Figure 7. Mean Normalized Knee Valgus/Varus Angle (+/- SD) Relative to Stance across All Subjects

4.2.2 Independent Variables

4.2.2.1 Trunk Proprioception

The average absolute error for forward flexion was used as a predictor variable for both knee flexion angle at initial contact and maximum knee flexion angle (Hypotheses 2a and 2b, respectively). Lateral flexion towards the side of the dominant leg was used as a predictor for knee valgus angle at initial contact (Hypothesis 1a) and total knee valgus excursion (Hypothesis 1b). Table 8 provides the descriptive statistics for these variables.

Table 8. Independent Variables: Trunk Proprioception Average Absolute Error

					Interquartile Range			Range		
	Mean	±	SD	Median	(25th	,	75th)	(Min	,	Max)
Forward flexion (°)	2.28	±	1.27	1.95	1.39	,	3.19	0.34	,	5.64
Dominant side lateral flexion (°)	1.63	±	0.78	1.54	1.09	,	1.99	0.47	,	4.16

4.2.2.2 Trunk Kinematics

Trunk forward flexion angle at initial contact was used as a predictor variable for knee flexion angle at initial contact (Hypothesis 2a) while maximum trunk forward flexion angle during the stance phase was used as a predictor for maximum knee flexion angle (Hypothesis 2b). Trunk lateral flexion angle at initial contact was used as a predictor for knee valgus angle at initial contact (Hypothesis 1a) while maximum trunk lateral flexion during the stance phase was used to predict total knee valgus excursion (Hypothesis 1b). Table 9 provides the descriptive statistics for these variables. Mean normalized trunk forward flexion and mean normalized trunk lateral flexion angle across all subjects with standard deviation relative to stance phase are presented in Figures 8 and 9, respectively.

Table 9. Independent Variables: Trunk Kinematics

					Interquartile Range			Range		
	Mean	±	SD	Median	(25th	,	75th)	(Min	,	Max)
Forward flexion angle at initial contact (°)	20.67	±	9.29	20.43	14.44	,	27.41	1.03	,	38.79
Maximum forward flexion angle (°)	32.32	±	10.75	33.58	24.30	,	39.31	10.08	,	55.00
Lateral flexion angle at initial contact (°)	10.26	±	6.98	10.27	5.17	,	13.25	-4.18	,	28.02
Maximum lateral flexion angle (°)	16.37	±	6.74	16.07	11.87	,	20.30	-0.83	,	32.63

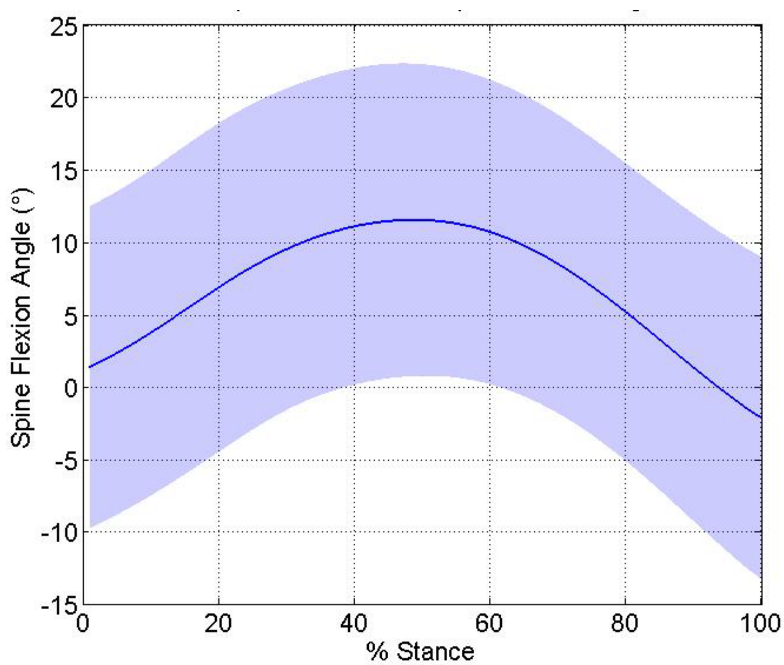


Figure 8. Mean Normalized Trunk Forward Flexion Angle (+/- SD) Relative to Stance across All Subjects

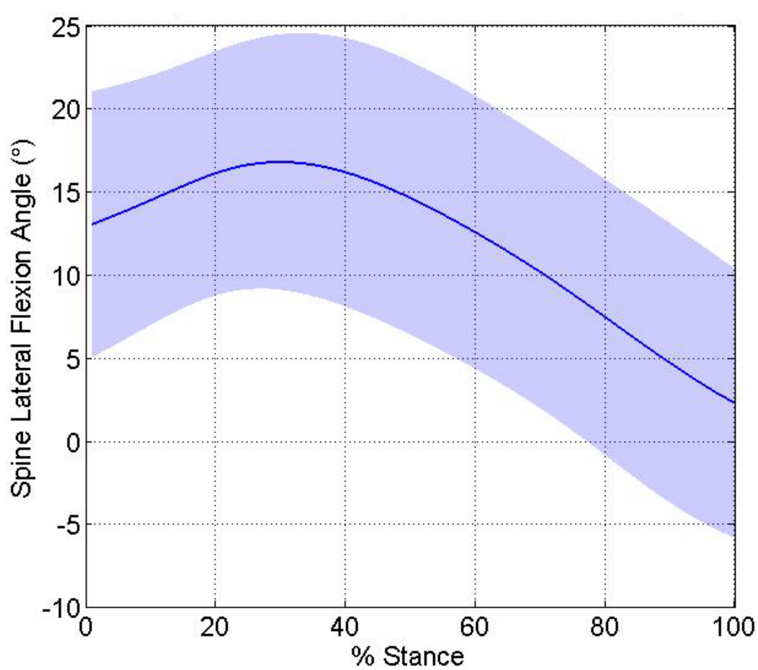


Figure 9. Mean Normalized Trunk Lateral Flexion Angle (+/- SD) Relative to Stance across All Subjects

4.2.2.3 Trunk Muscular Strength

Trunk extension and non-dominant side trunk rotation (rotation away from the dominant leg) average peak torque were used as independent variables. Trunk extension average peak torque was used as a predictor variable for knee valgus angle at initial contact (Hypothesis 1a), knee flexion angle at initial contact (Hypothesis 2a), and maximum knee flexion angle (Hypothesis 2b). Non-dominant trunk rotation average peak torque was used as a predictor for total knee valgus excursion (Hypothesis 1b). Table 10 provides the descriptive statistics for these variables.

Table 10. Independent Variables: Trunk Strength

					Interquartile Range			Range		
	Mean	±	SD	Median	(25th	,	75th)	(Min	,	Max)
Trunk extension (Nm)	270.0	±	95.6	240.9	200.5	,	342.6	95.6	,	638.3
Non-dominant trunk rotation (Nm)	93.6	±	33.6	84.1	65.5	,	119.8	51.0	,	179.2

4.2.2.4 Sex

Sex also was an independent variable in the current study. Twenty five males and 28 females participated in this study.

4.2.3 Normality

Normality was assessed using Shapiro-Wilk tests ($p<0.05$). For the dependent variables, total knee valgus excursion was not normally distributed ($p<0.05$). The following independent variables also were not normally distributed: forward flexion average absolute error, dominant side lateral flexion average absolute error, trunk extension average peak torque, and non-dominant trunk rotation average peak torque ($p<0.05$).

4.3 BIVARIATE ANALYSIS

Each of the four dependent variables was plotted against their respective independent variables. Scatterplots are presented in Appendix C. As knee valgus angle at initial contact (Hypothesis 1a) increases, dominant side lateral flexion average absolute error and trunk extension average peak torque also increase while spine lateral flexion angle at initial contact decreases. As total knee valgus excursion (Hypothesis 1b) increases dominant side lateral flexion average absolute error and non-dominant trunk rotation average peak torque decrease. As knee flexion angle at initial contact (Hypothesis 2a) increases, trunk extension average peak torque also increases.

Pearson correlation coefficients and the corresponding two-sided p-values were calculated for each of the dependent variables and its respective independent variables in order to aid in model interpretation. Significance level was set at $p<0.05$. Correlation matrices for each dependent variable and the respective independent variables are presented in Tables 11 through 14. A complete correlation matrix of all variables is available in Appendix D.

For knee valgus angle at initial contact (Hypothesis 1a), there were no significant correlations with any of the independent variables. For total knee valgus excursion (Hypothesis 1b), there was a significant negative correlation with non-dominant trunk rotation average peak torque (*r*=-0.364, p=0.007), indicating that as non-dominant trunk rotation strength decreased total knee valgus excursion increased. For knee flexion angle at initial contact (Hypothesis 2a), there was a significant positive correlation with trunk extension average peak torque (*r*=0.498, p<0.001), indicating that as trunk extension strength increased knee flexion angle at initial contact increased. For maximum knee flexion angle (Hypothesis 2b), there were no significant correlations with any of the independent variables.

Table 11. Pearson Correlation and 2-sided p-values: Knee Valgus Angle at Initial Contact and Continuous Independent Variables

	Knee valgus angle at initial contact (°)	Dominant side lateral flexion average absolute error (°)	Spine lateral flexion angle at initial contact (°)	Trunk extension average peak torque (Nm)
	r (p-value)	*r* (p-value)	*r* (p-value)	*r* (p-value)
Dominant side lateral flexion average absolute error (°)	0.099 (0.479)			
Spine lateral flexion angle at initial contact (°)	-0.081 (0.562)	-0.084 (0.548)		
Trunk extension average peak torque (Nm)	0.204 (0.144)	0.098 (0.484)	0.055 (0.696)	

Table 12. Pearson Correlation Coefficients and 2-sided p-values: Total Valgus Excursion and Continuous Independent Variables

	Total Knee Valgus Excursion	Dominant side lateral flexion average absolute error (°)	Maximum spine lateral flexion angle (°)	Non-dominant trunk rotation average peak torque (Nm)
	r (p-value)	*r* (p-value)	*r* (p-value)	*r* (p-value)
Dominant side lateral flexion average absolute error (°)	-0.169 (0.226)			
Maximum spine lateral flexion angle (°)	-0.111 (0.429)	-0.132 (0.348)		
Non-dominant trunk rotation average peak torque (Nm)	-0.460 (0.001)*	0.111 (0.429)	0.053 (0.707)	

* Statistically significant correlation ($p<0.05$)

Table 13. Pearson Correlation Coefficients and 2-sided p-values: Knee Flexion Angle at Initial Contact and Continuous Independent Variables

	Knee flexion angle at initial contact (°)	Forward flexion average absolute error (°)	Spine flexion angle at initial contact (°)	Trunk extension average peak torque (Nm)
	r (p-value)	*r* (p-value)	*r* (p-value)	*r* (p-value)
Forward flexion average absolute error (°)	-0.044 (0.752)			
Spine flexion angle at initial contact (°)	0.250 (0.071)	0.126 (0.368)		
Trunk extension average peak torque (Nm)	0.498 (0.003)*	0.040 (0.775)	0.478 (<0.001)*	

* Statistically significant correlation ($p<0.05$)

Table 14. Pearson Correlation Coefficients and 2-sided p-values: Maximum Knee Flexion Angle and Continuous Independent Variables

	Maximum knee flexion angle (°)	Forward flexion average absolute error (°)	Maximum spine flexion angle (°)	Trunk extension average peak torque (Nm)
	r (p-value)	*r* (p-value)	*r* (p-value)	*r* (p-value)
Forward flexion average absolute error (°)	-0.007 (0.996)			
Maximum spine flexion angle (°)	0.265 (0.055)	0.231 (0.096)		
Trunk extension average peak torque (Nm)	0.019 (0.892)	0.040 (0.775)	0.453 (0.001)*	

* Statistically significant correlation (p<0.05)

Since sex is a dichotomous variable, independent samples t-tests were performed in order to assess if there were significant differences in the mean of each of the dependent variables between males and females. There was no significant difference in knee valgus angle at initial contact (Hypothesis 1a) or maximum knee flexion angle (Hypothesis 2b) between males and females (p=0.834 and p=0.203, respectively). However, there were significant differences between males and females for both total knee valgus excursion (Hypothesis 1b; males: 3.7°±3.8°, females: 8.8°±5.3°; p<0.001) and knee flexion angle at initial contact (Hypothesis 2a; males: 29.9°±9.4°, females: 21.6°±6.3°; p=0.001).

Correlations were examined among the independent variables for each of the four regression equations. Significant correlations were found between spine flexion angle at initial contact and maximum spine flexion angle, trunk extension average peak torque, and non-dominant trunk rotation average peak torque (r=0.861, p<0.001; r=0.478, p<0.001; and r=0.435, p=0.001, respectively). Spine lateral flexion angle at initial contact was significantly correlated with maximum spine lateral flexion angle (*r*=0.667, $p<0.001$). Maximum spine flexion angle was significantly correlated with trunk extension average peak torque and non-dominant trunk rotation average peak torque (*r*=0.453, $p=0.001$; and *r*=0.351, $p=0.010$, respectively). Lastly, trunk extension average peak torque was significantly correlated with non-dominant trunk rotation average peak torque (*r*=0.794, $p<0.001$). However, collinearity likely was not an issue as none of the independent variables that were highly correlated were in the same regression equations.

4.3.1 Simple Linear Regression Models

A total for four different hypotheses were tested in the current study. In order to better understand the relationship between the dependent variables and each of the independent variables in each of the hypotheses, simple linear regression was performed. Results are presented in Tables 15-18. Jackknife residuals then were plotted against the predicted values (Appendix E). All scatterplots except for total knee valgus excursion by non-dominant trunk rotation average peak torque are randomly scattered around zero without any pattern, indicating that the assumptions of linearity and homoscedasticity have been met. The scatterplot of total knee valgus excursion by non-dominant trunk rotation average peak torque demonstrates evidence of heteroscedasticity, which was explored further and found not to be an issue. In addition, all scatterplots indicated that there are no obvious outliers.

For knee valgus at initial contact (Hypothesis 1a), none of the independent variables were significant predictors. For total knee valgus excursion (Hypothesis 1b), non-dominant trunk rotation average peak torque and sex were individual significant predictors ($p\leq0.001$), with each variable accounting for approximately 21% and 24% of the variance in total knee valgus excursion. Trunk extension average peak torque and sex were significant predictors of knee flexion angle at initial contact (Hypothesis 2a; $p<0.001$), accounting for approximately 25% and 22% of the variance in the dependent variable, respectively. None of the independent variables were significant predictors of maximum knee flexion angle (Hypothesis 2b).

Table 15. Simple Linear Regression: Knee Valgus Angle at Initial Contact as Dependent Variable

	Knee Valgus Angle at Initial Contact		
	MSE	R^2	p-value
Dominant side lateral flexion average absolute error (°)	33.18	0.010	0.479
Spine lateral flexion angle at initial contact (°)	33.29	0.007	0.562
Trunk extension average peak torque (Nm)	32.12	0.041	0.144
Sex	33.48	0.001	0.834

Table 16. Simple Linear Regression: Total Knee Valgus Excursion as Dependent Variable

	Total Knee Valgus Excursion		
	MSE	R^2	p-value*
Dominant side lateral flexion average absolute error (°)	27.57	0.029	0.226
Maximum spine lateral flexion angle (°)	28.03	0.012	0.429
Non-dominant trunk rotation average peak torque (Nm)	22.38	0.211	0.001*
Sex	21.68	0.236	<0.001*

* Statistically significant ($p<0.05$)

Table 17. Simple Linear Regression: Knee Flexion at Initial Contact as Dependent Variable

	Knee Flexion Angle at Initial Contact		
	MSE	R^2	p-value*
Forward flexion average absolute error (°)	80.55	0.002	0.752
Spine flexion angle at initial contact (°)	75.68	0.062	0.071
Trunk extension average peak torque (Nm)	60.74	0.248	<0.001*
Sex	62.60	0.224	<0.001*

* Statistically significant (p<0.05)

Table 18. Simple Linear Regression: Maximum Knee Flexion Angle as Dependent Variable

	Maximum Knee Flexion Angle		
	MSE	R^2	p-value
Forward flexion average absolute error (°)	53.82	0.000	0.996
Maximum spine flexion angle (°)	50.03	0.070	0.055
Trunk extension average peak torque (Nm)	53.80	0.000	0.892
Sex	52.12	0.032	0.203

4.4 MULTIPLE LINEAR REGRESSION MODELS

Backwards stepwise linear regression was performed for each of the four dependent variables with their respective independent variables.

4.4.1 Knee Valgus Angle at Initial Contact

The final regression model for knee valgus angle at initial contact (Hypothesis 1a) contained the following independent variables: spine lateral flexion at initial contact, trunk extension average peak torque, and sex (Table 19). However, the model was not significant [(F(3,49)=1.38, *p*=0.259] and only explained 7.8% of the variance in knee valgus angle at initial contact. Jackknife residuals were plotted against the predicted values and appeared to be randomly scattered around zero, indicating that the assumptions of linearity and homoscedasticity have been met (Appendix F). In addition, no obvious outliers were detected. The result of the Shapiro Wilk test indicated that the residuals were normally distributed (*p*=0.872). Collinearity was not an issue as none of the VIFs exceeded 10. In addition, no outliers [t(n-p-2, 0.05/2*n)] or influential points (Cook's D) were detected. Four subjects were detected as potential high leverage points. Robust regression was performed (Table 20); however, the parameter estimates did not change, the standard errors increased slightly, and the R^2 decreased slightly, indicating that robust regression did not improve the model fit.

Table 19. Regression Model: Knee Valgus Angle at Initial Contact (OLS)

Multiple Linear Regression					
Source	SS	df	MS	Observations	53
Model	133.56	3	44.52	F(3, 49)	1.38
Residual	1575.31	49	32.15	Prob > F	0.259
Total	1708.87	52	32.86	R^2	0.078
				Adjusted R^2	0.022

Predictor Variables	Coefficients	t	p-value
Sex	2.636	1.22	0.228
Spine lateral flexion angle at initial contact	-0.103	-0.90	0.375
Trunk extension average peak torque	0.021	1.94	0.058
Constant	-6.851	-1.81	0.077

Table 20. Robust Regression Model: Knee Valgus Angle at Initial Contact (WLS)

Multiple Linear Regression			
		Observations	53
		F(3, 49)	1.14
		Prob > F	0.342
		R^2	0.065
Predictor Variables	Coefficients	t	p-value
Spine lateral flexion angle at initial contact	-0.095	-0.78	0.439
Trunk extension average peak torque	0.020	1.78	0.081
Sex	2.804	1.23	0.225
Constant	-6.887	-1.72	0.092

4.4.2 Total Knee Valgus Excursion

The final regression model for total knee valgus excursion (Hypothesis 1b) contained the following independent variables: non-dominant trunk rotation average peak torque and sex (Table 21). This model was significant [(F(2,50)=8.75, *p*=0.001] and explained 25.9% of the variance in total knee valgus excursion. However, neither variable was a significant predictor within the model (*p*>0.05). Jackknife residuals were plotted against the predicted values and appeared to be randomly scattered around zero without any pattern, indicating that the assumptions of linearity and homoscedasticity have been met (Appendix F). In addition, no obvious outliers were detected. The result of the Shapiro Wilk test indicated that the residuals were normally distributed (*p*=0.215). Collinearity was not an issue and no outliers were detected. Two potential influential points were identified and a robust regression was performed (Table 22); however, the R^2 decreased slightly and the parameter estimates and the standard errors did not change much, indicating that the robust regression did not outperform the original regression.

With the other variables held constant, total knee valgus excursion decreased by 0.035 degrees for every one Newton*meter increase in non-dominant trunk rotation. In addition, females tended to have greater total knee valgus excursion than males, by 3.364 degrees. The final regression equation was:

Total knee valgus excursion = 7.906 - 0.035 (Non-dominant trunk rotation average peak torque) + 3.364 (Sex)

Table 21. Regression Model: Total Knee Valgus Excursion (OLS)

Multiple Linear Regression					
Source	SS	df	MS	Observations	53
Model	375.31	2	187.65	F(2, 50)	8.75
Residual	1071.93	50	21.44	Prob > F	0.0006*
Total	1447.24	52	27.83	R^2	0.259
				Adjusted R^2	0.230

Predictor Variables	Coefficients	t	p-value
Sex	3.364	1.80	0.078
Non-dominant trunk rotation average peak torque	-0.035	-1.26	0.215
Constant	7.906	2.27	0.027*

* Statistically significant (p<0.05)

Table 22. Robust Regression Model: Total Knee Valgus Excursion (WLS)

Multiple Linear Regression			
		Observations	53
		F(2, 50)	7.52
		Prob > F	0.001*
		R^2	0.231
Predictor Variables	Coefficients	t	p-value
Non-dominant trunk rotation average peak torque	-0.028	-0.92	0.361
Sex	3.843	1.89	0.064
Constant	6.659	1.76	0.084

* Statistically significant (p<0.05)

4.4.3 Knee Flexion Angle at Initial Contact

The final regression model for knee flexion angle at initial contact (Hypothesis 2a) contained the following independent variables: trunk extension average peak torque and sex (Table 23). This model was significant [(F(2,50)=9.83, p<0.001] and explained 25.4% of the variance in knee flexion angle at initial contact. However, none of the independent variables were significant predictors within the model. Jackknife residuals then were plotted against the predicted values and appeared to be randomly scattered around zero without any pattern, indicating that the assumptions of linearity and homoscedasticity have been met (Appendix F). In addition, no obvious outliers were detected. The result of the Shapiro Wilk test indicated that the residuals were normally distributed (p=0.782). Collinearity was not an issue and no outliers or influential points were detected. One subject was detected as a potential high leverage point and a robust regression was performed (Table 24); however, the changes in the parameter estimates, standard errors, and R^2 were small, indicating that the robust regression did not outperform the original regression.

With the other variables held constant, knee flexion angle at initial increased by 0.028 degrees for every one Newton*meter increase in trunk extension average peak torque. In addition, females tended to have lower knee flexion angle at initial contact than males, by 4.466 degrees. The final regression equation was:

Knee flexion angle at initial contact = 20.167 + 0.028 (Trunk extension average peak torque) – 4.466 (Sex)

Table 23. Regression Model: Knee Flexion Angle at Initial Contact (OLS)

Multiple Linear Regression					
Source	SS	df	MS	Observations	53
Model	1161.78	2	580.89	F(2, 50)	9.83
Residual	2954.61	50	59.09	Prob > F	0.0003*
Total	4116.39	52	79.16	R^2	0.282
				Adjusted R^2	0.254

Predictor Variables	Coefficients	t	p-value
Sex	-4.466	-1.56	0.126
Trunk extension average peak torque	0.028	2.01	0.050
Constant	20.167	3.96	0.000*

* Statistically significant ($p<0.05$)

Table 24. Robust Regression Model: Knee Flexion Angle at Initial Contact (WLS)

Multiple Linear Regression	
Observations	53
F(2, 50)	8.62
Prob > F	0.0006*
R^2	0.256

Predictor Variables	Coefficients	t	p-value
Trunk extension average peak torque	0.030	1.99	0.052
Sex	-4.057	-1.34	0.186
Constant	19.538	3.64	0.001*

* Statistically significant ($p<0.05$)

4.4.4 Maximum Knee Flexion Angle

The final regression model for maximum knee flexion angle (Hypothesis 2b) contained maximum spine flexion angle, trunk extension average peak torque, and sex (Table 25). However, the model was not significant [($F(3,49)=2.27$, $p=0.092$] and only explained 6.8% of the variance in maximum knee flexion angle. Jackknife residuals were plotted against the predicted values and appeared to be randomly scattered around zero without any pattern, indicating that the assumptions of linearity and homoscedasticity have been met (Appendix F). In addition, no obvious outliers were detected. The result of the Shapiro Wilk test indicated that the residuals were normally distributed (*p*=0.662). Collinearity was not an issue. In addition, no outliers or influential points were detected. One subject was detected as a potential high leverage point and a robust regression was performed (Table 26); however, the changes in the parameter estimates, standard errors, and R^2 were small, indicating that the robust regression did not outperform the original regression.

Table 25. Regression Model: Maximum Knee Flexion Angle (OLS)

Multiple Linear Regression					
Source	SS	df	MS	Observations	53
Model	335.21	3	111.74	F(3, 49)	2.27
Residual	2409.52	49	49.17	Prob > F	0.092
Total	2744.73	52	52.78	R^2	0.112
				Adjusted R^2	0.068

Predictor Variables	Coefficients	t	p-value
Maximum spine flexion angle	0.204	2.00	0.051
Trunk extension average peak torque	-0.021	-1.57	0.122
Sex	-3.879	-1.47	0.147
Constant	62.444	12.00	0.000*

* Statistically significant (p<0.05)

Table 26. Regression Model: Maximum Knee Flexion Angle (WLS)

Multiple Linear Regression			
		Observations	53
		F(3, 49)	1.81
		Prob > F	0.157
		R^2	0.010
Predictor Variables	Coefficients	t	p-value
Maximum spine flexion angle	0.185	1.65	0.106
Trunk extension average peak torque	-0.023	-1.53	0.132
Sex	-4.306	-1.48	0.144
Constant	63.789	11.12	0.000*

* Statistically significant ($p<0.05$)

5.0 DISCUSSION

The purpose of the current study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex could predict knee kinematics during a stop jump-cut maneuver (SJCM). Knee kinematics included knee valgus angle at initial contact (Hypothesis 1a), total knee valgus excursion (Hypothesis 1b), knee flexion angle at initial contact (Hypothesis 2a), and maximum knee flexion angle (Hypothesis 2b). Backwards stepwise multiple linear regression analysis was performed in order to determine which independent variables were significant predictors.

For knee valgus angle at initial contact (Hypothesis 1a), it was hypothesized that greater knee valgus angle would be predicted by lower trunk extension strength, higher (worse) dominant trunk lateral flexion proprioception, greater trunk lateral flexion angle at initial contact of the SJCM, and female sex. Multiple linear regression analysis produced a model that contained spine lateral flexion angle at initial contact, trunk extension average peak torque, and sex. However, the model only explained 7.8% of the variance in knee valgus angle at initial contact and was not statistically significant; therefore, the original hypothesis was not supported.

It was hypothesized that greater total knee valgus excursion (Hypothesis 1b) would be predicted by lower non-dominant trunk rotation strength, higher (worse) dominant trunk lateral flexion proprioception, greater maximum trunk lateral flexion angle during the SJCM, and female sex. Multiple linear regression analysis for total knee valgus excursion produced a model that contained non-dominant trunk rotation average peak torque and sex. This model was significant and together these two variables accounted for 25.9% of the variance in total knee valgus excursion; however, neither variable was a significant predictor within the model. The original hypothesis was only partial supported as only two of the independent variables were included in the model. The direction of the relationships of non-dominant trunk rotation strength and of sex with total knee valgus excursion was as hypothesized. With the other variables held constant, total knee valgus excursion decreased by 0.035 degrees for every one Newton*meter increase in non-dominant trunk rotation average peak torque. In addition, females tended to have higher total knee valgus excursion than males, by 3.36 degrees.

It was predicted that greater knee flexion angle at initial contact (Hypothesis 2a) would be predicted by greater trunk extension strength, lower (better) trunk flexion proprioception, greater trunk flexion angle at initial contact during the SJCM, and male sex. Multiple linear regression analysis produced a model that contained trunk extension average peak torque and sex. This model was significant and together these variables accounted for 25.4% of the variance in knee flexion at initial contact; however, neither of the independent variables were significant predictors within the model. The original hypothesis was only partial supported by these finding as only two of the independent variables were included in the model. The direction of the relationship between trunk extension average peak torque and knee flexion angle at initial contact was as hypothesized. In addition, the relationship between sex and knee flexion angle at initial contact was as hypothesized. With the other variables held constant, knee flexion angle at initial contact increased by 0.028 degrees for every one Newton*meter increase in trunk extension average peak torque. In addition, males tended to have higher knee flexion angle at initial contact than females, by 4.47 degrees.

It was predicted that higher maximum knee flexion angle (Hypothesis 2b) would be predicted by greater trunk extension strength, lower (better) trunk flexion proprioception, greater maximum trunk flexion angle during the SJCM, and male sex. Multiple linear regression analysis for maximum knee flexion angle produced a model that contained maximum spine flexion angle, trunk extension average peak torque, and sex. However, the model only explained 6.8% of the variance in maximum knee flexion angle and was not statistically significant; therefore the original hypothesis was not supported.

The dependent and independent variables will be discussed relative to previous research individually and then with respect to the four hypotheses. Limitations will then be presented. This will be followed by the clinical significance, recommendations for future research, and conclusions.

5.1 KNEE KINEMATICS

Knee kinematics of the dominant leg were assessed during stance phase of a SJCM. The following variables were extracted and averaged across three trials: knee valgus angle at initial contact, total knee valgus excursion, knee flexion angle at initial contact, and maximum knee flexion angle.

5.1.1 Knee Valgus Angle at Initial Contact

Knee valgus angle at initial contact was -0.94 ± 5.73° in the current study. These results are similar to those reported in other studies utilizing similar tasks. Beaulieu et al.[215] reported that male and female elite level soccer players landed with 1.3 ± 6.2° and -2.7 ± 7.3°, respectively, during an unanticipated run-and-cut maneuver (45°). Similar results were reported in male and female Division I basketball players during a run-and-cut maneuver (35-55°), with males demonstrating 2.2 ± 2.5° of knee valgus and females demonstrating -2.4 ± 3.7°.[216] Further, nearly identical values to those in the current study were reported in female college soccer players during a run-and-cut maneuver (-0.9 ± 1.4°).[217] During an unanticipated jump-and-cut task, female high school and basketball players landed with -1.4 ± 3.9° of knee valgus.[218] Since knee valgus angle at initial contact is comparable to those reported in previous research using similar tasks, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.1.2 Total Knee Valgus Excursion

In the current study, total knee valgus excursion was 6.38 ± 5.28°. Other studies that have reported knee valgus range of motion have reported similar values. Similar results were reported in female college soccer players performing a planned run-and-cut maneuver (4.5 ± 2.3°).[217] Male university volleyball players demonstrated knee valgus range of motion of 3.2 ± 8.0° and 3.5 ± 9.6° during unopposed and opposed (simulated blocking opponent) jump landing tasks, respectively, while females demonstrated 11.8 ± 10.3° and 8.8 ± 7.8°.[219] When total knee valgus excursion data in the current study are divided by sex, the values are similar to those in the previous study (males: 3.70 ± 3.79°; females: 8.78 ± 5.31°). Based on these comparisons, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.1.3 Knee Flexion Angle at Initial Contact

Knee flexion angle at initial contact was 25.50 ± 8.90° in the current study. This is within the range of knee flexion angle at initial contact reported in other studies using similar tasks. Similar results were reported in college-aged recreational athletes (22.0°),[220] male and female Division I basketball players (30.7± 10.7° and 25.2 ± 1.8°, respectively),[216] and Division I female soccer players (28.1 ± 4.2°)[217] during a planned run-and-cut maneuver. Beaulieu et al.[215] reported that male and female elite level soccer players landed with 15.6 ± 6.1° and 18.0 ± 6.8°, respectively, during an unanticipated run-and-cut maneuver (45°), which is less than the knee flexion angle at initial contact found in the current study. It is possible that these differences may be related to differences in the task (unanticipated run-and-cut vs. planned jump-and-cut) and/or population (elite athlete vs. physically active subjects). In contrast, James et al.[221] reported higher values as compared to the current study in high school and college basketball players (males: 46.0 ± 8.5°; females: 40.2 ± 8.4°). Again, this discrepancy may be due in part to differences in task (run-and-cut vs. jump-and-cut) and/or population (competitive athletes vs. physically active subjects). Since the knee flexion angle at initial contact is comparable to those reported in previous research using subjects of similar age and/or physical activity level, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.1.4 Maximum Knee Flexion Angle

In the current study, maximum knee flexion angle was 61.22 ± 7.27°. These results are similar to those reported in other studies utilizing similar tasks. During an unanticipated run-and-cut task, male and female elite level soccer players demonstrated 57.4 ± 5.0° and 57.9 ± 7.3°of maximum knee flexion, respectively.[215] Similar results were reported in a planned run-and-cut task in male and female Division I basketball players (males: 65.3 ± 5.7°; females: 61.9 ± 5.7°).[216] In contrast, Sell et al.[180] reported higher maximum knee flexion angles in high school basketball players (78.9 ± 8.7°). Although this was a stop-jump task, higher maximum knee flexion angles may have occurred due to differences in the task (unanticipated vs. planned, 90° cut vs. 45° cut) and/or population (high school athletes vs. college-aged physically active subjects). Since the maximum knee flexion angle is comparable to those reported in previous studies, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.2 TRUNK NEUROMUSCULAR AND KINEMATIC CHARACTERISTICS

5.2.1 Trunk Muscular Strength

Isokinetic trunk extension flexion and extension as well as right/left trunk rotation strength was assessed using concentric/concentric reciprocal contractions at 60°/s. Trunk extension and non-dominant trunk rotation (rotation away from the dominant leg) average peak torque were used in the current study. In order to compare this data to previous studies, average peak torque was normalized to body weight (%BW) for the purpose of this discussion.

In the current study, trunk extension average peak torque normalized to body weight was 371.01 ± 106.46 %BW. This is within the range of strength values reported using similar methodology. Trunk extension strength in the current study is nearly identical to that reported in healthy golfers (362 ± 87%) using the same methodology.[202,222] However, trunk extension strength in the current study is lower than that reported in healthy, collegiate wrestlers (510 ±

60%BW).[223] This is to be expected due to differences in training and in physical demand of intercollegiate wrestlers as compared to physically active young adults.

Right and left trunk rotation average peak torque normalized to body weight was 125.82 ± 32.39 %BW and 127.94 ± 32.25 %BW, respectively. This is within the range of strength values reported using identical methodology. In comparison to the values reported by Sell et al.,[224] trunk rotation strength in the current study was less than male triathletes (right: 151.51 ± 25.94 %BW; left: 154.57 ± 30.90 %BW) and male Soldiers (right: 145.12 ± 33.05 %BW; left: 144.82 ± 32.80%) but greater than female triathletes (right: 118.53 ± 24.59 %BW; left: 114.85 ± 25.74 %BW) and female Soldiers (right: 110.49 ± 32.89 %BW; left: 111.62 ± 28.02 %BW). Similar values were obtained in pooled data for male and female Soldiers (control: 128.5 ± 33.5 %BW; experimental: 137.7 ± 26.8%BW).[225] Slightly higher values also were reported by Tsai[202] (right: 141.72 ± 26.77°; left: 146.06 ± 26.40°), but this is to be expected as the subjects were golfers and trunk rotation strength may be more highly developed as a function of the sport. Based on these comparisons, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.2.2 Trunk Proprioception

In the current study, active joint position sense for trunk flexion and trunk lateral flexion to the dominant side was assessed using a passive video-based motion capture system and a custom guide. Subjects actively moved to a target position (Flexion: 20°, Lateral Flexion: 15°), held the position for five seconds, returned to the starting position, and then attempted to actively replicate the position.

The average absolute error for trunk flexion was 2.28 ± 1.27° in the current study. The average absolute error for trunk flexion is similar to the values reported in other studies using active joint position sense, even though there were variations in methodology. The methods used in the current study were a modification of those used by Tsai,[202] who reported similar values in healthy golfers (2.13 ± 0.86°). Nearly identical results to those in the current study were reported in healthy adults (2.3 ± 1.3°).[194] Similarly, Georgy[226] an average absolute error of 2.84 ± 0.94° in healthy adults. Better proprioception may have been demonstrated in the current study because the subjects were younger (22.0 ± 2.1yrs vs. 38.5 ± 5.9yrs) and the task was an active positioning-active repositioning task as compared to a passive positioning-active repositioning task.

The average absolute error for dominant side trunk lateral flexion was 1.63 ± 0.78° in the current study. As with trunk flexion proprioception, the average absolute error for trunk lateral flexion is similar to previously reported values. Tsai[202] reported an average absolute error of 1.57 ± 0.52° and 1.73 ± 0.67° for right and left lateral flexion, respectively. Lee et al.[194] reported an average absolute error of 1.9 ± 0.9° in healthy adults. Since these trunk proprioception values are comparable to those reported in previous studies, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.2.3 Trunk Kinematics

Trunk kinematics were assessed during the SJCM. The following variables were extracted and averaged across three trials: trunk flexion angle at initial contact, maximum trunk flexion angle, trunk lateral flexion angle at initial contact, and maximum trunk lateral flexion angle.

Trunk flexion angle at initial contact and maximum trunk flexion angle were 20.67 ± 9.29° and 32.32 ± 10.75°, respectively, in the current study. Blackburn & Padua[94] reported slightly lower trunk flexion angle at initial contact (14 ± 11°) and slightly larger maximum trunk flexion (49 ± 21°) in physically active males and females. However, the task performed was a vertical drop landing as compared to a SJCM used in the current study. In addition, in the previous study the trunk angle was calculated as the trunk relative to the thigh while the trunk angle was calculated at the trunk relative to the pelvis in the current study. While methodological differences make it difficult to compare the current results to those previously reported, it can be concluded that these results are representative of physically active, healthy college-aged adults.

Trunk lateral flexion angle at initial contact and maximum trunk lateral flexion angle were 10.26 ± 6.98° and 16.37 ± 6.74°, respectively, in the current study. Previous studies have not reported trunk lateral flexion angle at initial contact. The maximum trunk lateral flexion angle in the current study is slightly larger than that reported in other studies. During an unanticipated run-and-cut maneuver, both males (10.3 ± 10.1°) and females (6.4 °± 3.3°) demonstrated slightly lower maximum trunk lateral flexion.[8] The same researchers reported comparable values in healthy, physically active college-aged subjects (8.6 ± 5.3°).[227] However, in both studies angles were calculate based on the torso relative to the vertical in the global coordinate system as compared to the torso relative to the pelvis in the local coordinate system in the current study. When the values reported in the previous studies are evaluated relative to the maximum thorax angle in the frontal plane (torso relative to vertical in the global coordinate system) in the current study, the values are more similar (all subjects: 12.78 ± 8.90°; males: 14.09 ± 10.91°; females: 11.60 ± 6.60°). The small differences between these results and those in the current study may be due to methodological differences (unanticipated vs. planned cutting tasks). Smaller maximum trunk lateral flexion angles also were reported by Houck et al. (9.2 ± 2.8°);[228] however, a walking approach was used for the cutting task. Since the trunk lateral flexion angles in the current study are only slightly larger than those reported previously, it can be concluded that these results are representative of physically active, healthy college-aged adults.

5.3 PREDICTION OF KNEE KINEMATICS

Specific Aim 1 was to determine if trunk muscular strength, trunk proprioception, trunk kinematics during a SJCM, and sex could predict knee valgus angle at initial contact and total knee valgus excursion during a SJCM. Specific Aim 2 was to determine if trunk muscular strength, trunk proprioception, trunk kinematics during a SJCM, and sex could predict knee flexion angle at initial contact and maximum knee flexion angle during a SJCM.

Since more than half of the body's mass is located from the pelvis up,[229] the core musculature and the associated neural structures play a crucial role in controlling and positioning the trunk relative to the lower extremity in order to allow for the production, transfer, and control of force and motion to the extremities.[9] Further, the core provides a stable foundation for movement of the extremities;[68,77,82] therefore, inability of the core musculature to maintain alignment or control of the trunk may lead to malalignment of the lower extremity, particularly the knee, thereby increasing the risk of injury.[12,45,230]

It has been theorized that the core (i.e., lumbopelvic hip complex) can influence the positioning of the lower extremity, particularly the knee.[12,45] Deficits in neuromuscular control can cause medial or lateral displacement of the trunk. Due to the anatomic link of the femur and the trunk through the hip, this medial-lateral displacement of the trunk can influence frontal plane positioning of the knee.[12,45] If the center of mass moves towards (lateral to) the stance leg, knee valgus moment at the knee will increase, which in turn results in tensile strain on the medial knee structures (e.g., medial collateral ligament, anterior cruciate ligament) and compression of the lateral compartment of the knee.

Literature that discusses the theoretical link between the core and the knee has been focused on knee kinematics in the frontal plane. However, there have been two studies that have examined trunk kinematics in the sagittal plane relative to the knee. Blackburn and Padua[94] found that intentionally increasing trunk flexion during a double leg drop landing resulted in a concomitant increase in knee flexion angle. In contrast, Kulas et al.[49] found no significant difference in knee flexion angle in subjects who landed with greater trunk flexion and those who landed with less trunk flexion during a double leg drop landing.

The influence of neuromuscular characteristics related to core stability and of trunk kinematics on lower extremity kinematics has not been clearly established in the literature. Based on the theory presented in the literature as well as the limited number of studies available, independent variables related to neuromuscular control of the trunk (i.e., trunk muscular strength, trunk proprioception) and trunk kinematics were selected based on plane of motion relative to knee plane of motion.

5.3.1 Knee Valgus Angle at Initial Contact

It was hypothesized that greater knee valgus angle at initial contact (Hypothesis 1a) would be predicted by lower trunk extension muscular strength, higher (worse) trunk lateral flexion proprioception towards the dominant leg, greater lateral trunk displacement at initial contact during a SJCM, and female sex. Multiple linear regression analysis produced a model that contained spine lateral flexion angle at initial contact, trunk extension average peak torque, and sex. However, the model only explained 7.8% of the variance in knee valgus angle at initial contact and was not statistically significant; therefore, the original hypothesis was not supported.

5.3.1.1 Trunk Extension Strength

Greater than fifty percent of the body's mass is located from the pelvis up;[229] therefore, the core musculature aids in the control and positioning the trunk relative to the lower extremity, which allows for force and motion production and transfer to the extremities.[9] Further, the core provides a stable foundation for movement of the extremities;[68,77,82] therefore, inability of the core musculature to maintain alignment or control of the trunk may lead to malalignment of the lower extremity, particularly the knee, thereby increasing the risk of injury.[12,45,230]

The task used in the current study, a stop jump-cut maneuver, involved jumping anteriorly and immediately planting-and-cutting away from the stance leg as quickly as possible. In order to perform this task, the trunk extensors must contract eccentrically to decelerate the trunk prior to and during landing. An inability to decelerate the trunk segment center of mass in preparation for initial contact and throughout the landing phase could result in increased knee valgus in an attempt to attenuate forces and decelerate the body.

There is very limited research exploring the relationship between trunk muscular characteristics and lower extremity kinematics or kinetics. Shirey et al.[231] reported that subjects with a high ability to activate the core musculature as assessed with the Sahrmann test demonstrated significantly less knee valgus during a single leg squat, both when the core musculature was intentionally engaged and when the core musculature was not intentionally engaged. Willson et al.[50] found a significant positive correlation between isometric trunk lateral flexion strength and knee frontal plane projection during a single leg squat. Although the correlation was with trunk lateral flexion strength, this study is still important relative to trunk extension strength as many of the trunk extensors become lateral flexors when acting unilaterally.[133-135] Jamison et al.[227] reported that an increase in the co-contraction index of the trunk extensors immediately prior to initial contact during an unanticipated run-to-cut maneuver resulted in significantly greater peak external knee abduction moment. It was postulated that increased co-contraction of the trunk extensors resulted in a stiffer spine and less trunk flexion, which necessitated greater kinetic energy absorption by the lower extremity.

Comparison between the results of the current study and previous research exploring the relationship between trunk muscular characteristics and knee kinematics and kinetics is difficult due to methodological differences. In the current study, muscle strength was assessed using isokinetic dynamometry. In the previous studies, muscle characteristics were assessed using the Sahrmann test,[231] electromyography,[227] and isometric trunk strength.[50] Since these assessments all measure different muscle characteristics, it is understandable that the relationship explored in their studies with knee kinematics or kinetics may differ. More importantly, the current study attempted to investigate the relationship between trunk extension strength and knee valgus angle at initial contact; however, previous studies that have examined trunk strength have only looked at strength relative to maximum knee valgus angles.

Despite these differences, previous research appears to indicate that better ability to activate the core[231] as well as greater strength[50] are related to decreased knee motion in the frontal plane during a single leg squat. Further, increased activation of the trunk extensors immediately prior to landing significantly increased peak external knee abduction moment.[227] In the current study, trunk extension strength was not found to be a significant predictor of knee valgus angle at initial contact. It is possible that trunk muscular strength, as related to frontal plane knee movement, may be more crucial during weight acceptance than prior to or at initial contact during quick-burst tasks (e.g., stop jump-cut maneuver). In addition, greater core muscular strength does not necessarily equate to better ability to activate the core musculature. Based on the findings of Jamison et al.,[227] muscle activation patterns of the trunk extensors rather than muscular strength may be of greater importance during athletic maneuvers, particularly prior to and at initial contact, with respect to knee frontal plane movement.

5.3.1.2 Trunk Proprioception

Previous literature has examined the relationship between trunk proprioception and knee injury. Zazulak et al.[51] reported that female athletes who sustained a knee injury or a knee ligament/meniscal injury demonstrated significantly worse proprioception as assessed during a passive placement-active replacement task in the transverse plane. For all athletes, a 2.9-fold increase of knee injury and a 3.3-fold increase of ligament/meniscal injury were reported for every one degree increase in repositioning error. In a follow-up study, Zazulak et al.[52] found that greater trunk angular displacement following a sudden release, worse trunk proprioception,[51] and previous history of low back pain were significant predictors of knee ligament injury in female athletes.

In the current study, trunk lateral flexion active joint position sense was not found to be a significant predictor of knee valgus angle at initial contact. Trunk proprioception in the transverse plane and its relationship with knee injury has been investigated previously.[51,52] Deficits in trunk proprioception in the transverse plane may be of greater importance in terms of risk of knee injury and, possibly, risky knee kinematics. If an individual is not able to accurately determine the position of the trunk relative to the lower extremity in the transverse plane, then the trunk may “trail” the motion, resulting in relative internal rotation of the thigh and dynamic valgus collapse of the knee. This idea of the relationship between the trunk and the knee in the transverse plane is supported by a recent study that found a significant negative correlation between trunk rotation towards the direction of travel and internal knee varus moment, which is equal to knee external valgus moment.[232] In other words, when there is less trunk rotation towards the direction of the cutting maneuver, knee external valgus moment increases, which may increase the load on the ACL.

Zazulak et al.[52] reported that lateral, extension, and flexion angular displacements of the trunk following a sudden release were significantly greater in athletes who sustained a knee injury, knee ligament injury, or ACL injury as compared to uninjured athletes. Collectively, trunk displacement in these three directions predicted ACL injury with 83% sensitivity and 76% specificity; however, lateral trunk displacement alone predicted knee ligament injury in female but not male athletes. In contrast, trunk proprioception in the frontal plane was not found to be a predictor of knee valgus at initial contact in the current study. This discrepancy between previous research and the findings in the current study may be due to methodological differences. In the current study, proprioception was assessed using an active positioning-active repositioning task. Zazulak et al.[52] assessed trunk angular displacement following a sudden release. It may be that the methods in the current study target the slow adapting mechanoreceptors (e.g., Ruffini endings, Golgi tendon-like organs)[124] whereas the methods employed by Zazulak et al. assess quick adapting mechanoreceptors (e.g., Pacinian corpuscles).[59,60,124] Further, due to the dynamic nature of task (e.g., angular trunk displacement following a sudden release) used by Zazulak et al., it is very likely that muscle characteristics, such as muscle activation patterns, play a role in displacement. While trunk angular displacement following a sudden release may assess overall neuromuscular control and is a predictor of knee injury, it does not help identify which component of neuromuscular control is deficient, making it difficult to develop targeted interventions.

5.3.1.3 Trunk Kinematics

The relationship between trunk kinematics in the frontal plane and knee kinematics or kinetics has been investigated using video analysis of ACL injury events[41] as well as in laboratory studies.[8,232] Using video analysis of ACL injury events, Hewett et al.[41] reported that female athletes who sustained a non-contact ACL injury demonstrated significantly greater lateral trunk motion and knee abduction in the 200ms following initial contact as compared to male athletes. Frank et al.[232] reported that trunk lateral flexion during a side-step cutting task was not significantly correlated with knee extension moment, knee varus moment, or knee external rotation moment. In contrast, Jamison et al.[8] reported a significant positive relationship between frontal plane torso angle away from the cutting direction and knee abduction moment.

In the current study, spine lateral flexion angle at initial contact was not found to be a predictor of knee valgus angle at initial contact. None of the previous studies have reported trunk kinematics and knee kinematics/kinetics at initial contact; therefore, the results of the current study cannot be compared to those of previous studies. It is possible that in order for the position of the trunk in the front plane to influence knee position in the frontal plane weight acceptance needs to occur. This is supported by the greater trunk motion in the frontal plane accompanied by increased knee valgus over the 200ms following initial contact reported during ACL injury events as compared to similar events that did not result in injury.[41]

5.3.1.4 Sex

Differences in knee kinematics at the time of injury have been reported between males and females when using video analysis of ACL injury events; however, these findings are not consistent across studies.[41,42,44] Hewett et al.[41] reported that females athletes who sustained an ACL injury demonstrated significantly greater knee valgus at initial contact than male athletes who sustained an ACL injury. Others, however, have reported that there is no difference in knee valgus angle at initial contact between male and female athletes during ACL injury.[42,44]

Kinematic differences between males and females have been examined in the laboratory setting, but produced conflicting results. It has been reported that females demonstrate significantly greater knee valgus angle at initial contact as compared to their male counterparts when performing tasks such as unanticipated stop jump-cut maneuvers (females: $-3.7 \pm 0.9°$, males: $-1.2 \pm 1.1°$),[96] anticipated run-to-cut tasks ($-2.4 \pm 3.7°$ vs. $2.2 \pm 2.5°$),[216] and medial and lateral single leg drop landings ($-0.5 \pm 2.2°$vs. $3.0 \pm 2.8°$ and $-2.4 \pm 2.0°$ vs. $1.7 \pm 2.3°$, respectively).[233] In contrast, other authors have reported no significant difference in knee valgus angle at initial contact between males and females during drop vertical jumps[234] and side-step cut maneuvers.[235,236]

In the current study, sex was not found to be a significant predictor of knee valgus angle at initial contact. This finding is consistent with some previous studies[42,44,234-236] but not others.[41,96,216,233] One possible explanation for these discrepancies is that previous research has used two-dimensional analysis of video footage of actual ACL injury events.[41,42,44] Knee kinematics recorded in the laboratory during events that do not result in ACL injury may not be identical to those that occur during actual injury events. It also is possible that differences in age and physical activity may explain conflicting results across studies. The subjects in the current study were college-aged (18-25 years old, inclusive) individuals who were physically active a minimum of one day per week in sports or activities that involved jumping, cutting, or running. Some previous studies have utilized middle school or high school athletes,[96,234,236] who have different neuromuscular characteristics than adults. In addition, previous research has utilized subjects who were current intercollegiate and/or professional athletes.[216,233,235] It is possible that neuromuscular control may be less developed in the former and more highly trained in the latter in comparison to the subjects in the current study. In addition, differences in tasks could account for discrepancies across studies. Some studies utilized anticipated[216,235,236] or unanticipated[96] cutting tasks similar to that used in the current study, while others used drop vertical jumps[234] or single leg landings.[233] However, since previous studies have used unique combinations of the aforementioned factors (i.e., injury vs. non-injury events, age/experience, task), it is difficult to ascertain if one or a select combination of factors affects the role that sex may play in knee valgus angle at initial contact.

5.3.2 Total Knee Valgus Excursion

Multiple linear regression analysis for total knee valgus excursion (Hypothesis 1b) produced a model that contained non-dominant trunk rotation (rotation away from the dominant leg) average peak torque and sex. This model was significant and together these two variables accounted for 25.9% of the variance in total knee valgus excursion; however, neither variable was a significant predictor within the model. The original hypothesis was only partial supported by these finding as only two of the independent variables were in the model. The direction of the relationships of non-dominant trunk rotation strength and of sex with total knee valgus excursion was as hypothesized. It was hypothesized that greater total knee valgus excursion would be predicted by lower trunk rotation muscular strength towards the cutting direction, higher (worse) trunk lateral flexion proprioception towards the dominant leg, greater maximum lateral trunk displacement, and female sex. With the other variables held constant, total knee valgus excursion was decreased by 0.035 degrees for every one Newton*meter increase in non-dominant trunk rotation average peak torque. In addition, females tended to have higher total knee valgus excursion than males, by 3.36 degrees.

5.3.2.1 Trunk Rotation Strength

The core musculature must control and position the trunk relative to the lower extremity in order to allow for the production, transfer, and control of force and motion to the extremities.[9] This ensures a stable foundation for movement of the extremities.[68,77,82] If the core musculature is unable to maintain alignment or control of the trunk, then malalignment of the lower extremity may result, thereby increasing the risk of injury.[12,45,230]

The task used in the current study, a stop jump-cut maneuver, involved jumping anteriorly and immediately planting-and-cutting away from the stance leg as quickly as possible. In order to perform this task, the trunk extensors must contract eccentrically to decelerate the trunk prior to and during landing and then the trunk rotators must contract in order to turn the trunk into the direction of the cut. An inability to decelerate the trunk segment center of mass at initial contact and throughout the landing phase could result in reduced knee flexion angle at initial contact and maximum knee flexion angle. Further, if there is less knee flexion at initial contact, then there may be increased knee valgus in an attempt to attenuate forces. If the trunk musculature does not rotate the trunk towards the direction of the cutting maneuver, then the trunk may “trail” the motion, resulting in relative internal rotation of the thigh and dynamic valgus collapse of the knee. In a recent study, it was reported that as trunk rotation towards the direction of a cutting maneuver decreased, internal knee varus moment increased, thereby increasing strain on the ACL.[232] In addition, regression analysis found that trunk rotation and hip adduction moment together were significant predictors of internal knee varus moment, with less trunk rotation towards the cutting direction and greater hip abduction moment predicting greater internal knee varus moment.[232]

There is very limited research exploring the relationship between trunk muscular characteristics and lower extremity kinematics or kinetics. Subjects who are better able to activate the core musculature have been reported to demonstrate significantly less knee valgus during a single leg squat as compared to subjects with low ability to engage the core musculature.[231] Similarly, it has been reported that there is a significant positive correlation between isometric trunk lateral flexion strength and knee frontal plane projection angle.[50] Jamison et al.[227] reported that an increase in the co-contraction index of the trunk extensors immediately prior to initial contact during an unanticipated run-to-cut maneuver resulted in significantly greater peak external knee abduction moment.

Based on previous research, it appears that better ability to activate the core[231] as well as greater lateral trunk flexion isometric strength[50] are related to decreased knee motion in the frontal plane during a single leg squat. Further, increased activation of the trunk extensors immediately prior to landing significantly increased peak external knee abduction moment.[227] In the current study, trunk non-dominant trunk rotation strength, or trunk rotation strength away from the landing leg and towards the direction of the cutting maneuver, was found to be a significant predictor of total knee valgus excursion using multiple linear regression. Based on previous research and the results of the current study, it appears that trunk muscular strength plays an important role in modulating knee motion in the frontal plane during single leg tasks. More specifically, increased trunk rotation strength results in decreased knee frontal plane motion.

5.3.2.2 Trunk Proprioception

The relationship between trunk proprioception and knee kinematics has not been previously investigated. However, the relationship between trunk proprioception and knee injury has been examined. Deficits in trunk proprioception are predictive of knee injury and knee ligament/meniscal injury if female athletes.[51] In addition, significantly greater lateral, extension, and flexion trunk angular displacements following a sudden release were reported in athletes who sustained a knee injury, knee ligament injury, or ACL injury as compared to uninjured athletes.[52] However, lateral displacement alone predicted knee ligament injury in female but not male athletes.

In the current study, trunk lateral flexion active joint position sense was not found to be a significant predictor of total knee valgus excursion. However, previous research has demonstrated proprioceptive deficits of the trunk in the transverse plane in female athletes who sustained knee injuries and ligament/meniscal injuries.[51,52] The findings of previous research, in combination of the findings from the current study, appear to indicate that trunk proprioception in the transverse plane, but not in the frontal plane, may be related to knee injury risk. If an individual is not able to accurately determine the position of the trunk relative to the lower extremity in the transverse plane, then the trunk may "trail" the motion, resulting in relative internal rotation of the thigh and dynamic valgus collapse of the knee. This idea of the relationship between the trunk and the knee in the transverse plane is supported by a recent study that found a significant negative correlation between trunk rotation towards the direction of travel and internal knee varus moment.[232]

Previous research has found that lateral, extension, and flexion angular displacements following a sudden release were significantly greater in athletes who sustained a knee injury, knee ligament injury, or ACL injury as compared to uninjured athletes.[52] In contrast, trunk proprioception in the frontal plane was not found to be a predictor of total knee valgus excursion in the current study. It may be that the task in the current study targets the slow adapting mechanoreceptors (e.g., Ruffini endings, Golgi tendon-like organs)[124] whereas displacement following a sudden release targets quick adapting mechanoreceptors (e.g., Pacinian corpuscles).[59,60,124] While the task used in the previous study assesses overall neuromuscular control and is a predictor of knee injury, it does not help identify which component that contributes to neuromuscular control is deficient, making it difficult to develop targeted interventions.

5.3.2.3 Trunk Kinematics

Video analysis of ACL injury events[41] and laboratory studies[8,232] have examined the relationship between trunk kinematics in the frontal plane and knee kinematics or kinetics. By comparing

ACL injury events to similar events that did not result in injury, it was found that female athletes who sustained a non-contact ACL injury demonstrated significantly greater lateral trunk motion and knee abduction during the 200ms following initial contact than male athletes.[41] Frank et al.[232] reported that trunk lateral flexion during a side-step cutting task was not significantly correlated with knee varus moment or knee external rotation moment. In contrast, Jamison et al.[8] reported a significant positive relationship between frontal plane torso angle away from the cutting direction and knee abduction moment.

Maximum spine lateral flexion angle was not found to be a predictor of total knee valgus excursion in the current study. The maximum spine lateral flexion angle was 16.4 ± 6.7° in the current study. While this value is similar to those in injured females (11.1 ± 2.0°), the trunk lateral flexion angles were much lower in injured males (-5.5 ± 9.5°) and uninjured females (4.2 ± 9.6°).[41] In contrast, the maximum spine lateral flexion angle was comparable between males and females (17.3 ± 7.9° and 15.5 ± 5.5°, respectively) in the current study. It is possible that the discrepancy between this finding in the current study and those reported by Hewett et al.[41] may be due to methodological differences. In the current study, a passive motion capture system was used while video editing software was used in the other study. In addition, trunk lateral flexion angle was calculated as the trunk relative to the pelvis in the current study whereas trunk position was measured relative to vertical in the other. However, the maximum spine lateral flexion angle (relative to the pelvis) and the maximum thorax lateral flexion angle (relative to vertical in the global coordinate system) were similar in the current study (16.4 ± 6.7° and 12.8 ± 8.9°, respectively). These discrepancies may be further explained by differences in the task (planned jump-and-cut maneuver in laboratory setting vs. landing and cutting tasks during basketball game).

Additional analysis of kinematic data in the current study indicates that total knee valgus excursion may be more influenced by trunk kinematics in the sagittal plane than by trunk kinematics in the frontal plane. Significant negative correlations were found between total knee valgus excursion and spine flexion angle at initial contact and maximum spine flexion angle (r=-0.287, p=0.037; r=-0.293, p=0.033, respectively). In contrast, neither spine lateral flexion at initial contact (r=-0.078, p=0.581) nor maximum (r=-0.111, p=0.429) were found to be significantly correlated with total knee valgus excursion.

5.3.2.4 Sex

Inconsistent findings have been reported across studies when examining knee valgus excursion using video analysis of ACL injury events.[41,42,44] Although total knee valgus excursion has not been reported, investigators have examined knee valgus progression from initial contact to 50ms or 5 frames following initial contact. Hewett et al.[41] reported that female athletes who sustained an ACL injury demonstrated significantly greater progressive knee valgus as compared to controls. Significantly greater knee valgus angle at 50ms following initial contact[42] and in the fifth frame[44] following initial contact was reported in females during ACL injury as compared to male athletes who sustained ACL injury. Further, valgus collapse was reported in almost half of the females (9 out of 17) who sustained ACL injury as compared to less than 20% of the males (2 out of 12).[42]

Total knee valgus excursion and peak knee valgus angle have been examined in the laboratory setting, but conflicting results have been reported. Ford et al.[234] found that female high school basketball players demonstrated significantly greater total knee valgus excursion during a drop vertical jump than their male counterparts (females: 7.3 ± 0.5cm, males: 5.3 ± 0.5cm) as well as significantly greater maximum knee valgus angle (females: 27.6 ± 2.8°, males:

16.1 ± 2.1°). Similarly, female collegiate basketball and soccer players have been reported to demonstrate significantly greater total knee valgus excursion during medial (female: 6.6 ± 2.1°, males: 5.1 ± 1.2°) and lateral (females: 6.1 ± 1.8°, males: 4.8 ± 1.1°) single leg drop landings.[233] Significantly greater maximum knee valgus angle has been reported in female collegiate basketball players[216] and healthy adults[237] during run-and-cut tasks. However, no differences between sexes were reported for maximum knee valgus angle in middle school and high school basketball players,[96] state-level athletes,[236] or collegiate soccer players[235] performing stop jump-cut maneuvers.

In the current study, sex was found to be a significant predictor of total knee valgus excursion, with female subjects demonstrating significantly greater total knee valgus excursion than males. The results of the current study are consistent with the findings in some of the previous studies[42,44,216,233,234,237] and inconsistent with others.[41,235,236] One possible explanation for the discrepancies across studies is difference in age of the subjects. The subjects in the current study were college-aged (18-25 years old, inclusive). Two of the previous studies utilized middle school or high school athletes,[96,234] who have different neuromuscular characteristics than adults, while others have used intercollegiate athletes[42,44,216,233,235] or healthy adults.[237] It is possible that neuromuscular control may be more highly trained in the latter and less developed in the former in comparison to the subjects in the current study. The majority of the studies with adult subjects reported significant differences in total knee valgus excursion or maximum knee valgus angle between the sexes, regardless of the task performed. The one exception was Sigward et al.,[235] who reported no difference in any knee kinematics between male and female collegiate soccer players during a run-and-cut maneuver. However, kinematics were only evaluated during the first 20% of stance phase as compared to total stance phase in the current study.

5.3.3 Knee Flexion Angle at Initial Contact

For knee flexion angle at initial contact (Hypothesis 2a), the multiple linear regression analysis produced a model that contained trunk extension average peak torque and sex. This model was significant and together these variables accounted for 25.4% of the variance in knee flexion at initial contact; however, neither of the independent variables were significant predictors within the model. The original hypothesis was only partial supported by these finding as only two of the independent variables remained in the model. The direction of the relationship between trunk extension average peak torque and knee flexion angle at initial contact was as hypothesized. In addition, the relationship between sex and knee flexion angle at initial contact was as hypothesized. It was hypothesized that greater knee flexion angle at initial contact would be predicted by greater trunk extension muscular strength, lower (better) trunk flexion proprioception, greater trunk flexion angle at initial contact, and male sex. With the other variables held constant, knee flexion angle at initial contact increased by 0.028 degrees for every one Newton*meter increase in trunk extension average peak torque. In addition, females tended to have lower knee flexion angle at initial contact than males, by 4.47 degrees.

5.3.3.1 Trunk Extension Strength

The core musculature is responsible for controlling and positioning the trunk relative to the lower extremity.[9] Effective control and positioning of the trunk allows for the production, transfer, and control of force and motion to the extremities;[9] provides a stable foundation for movement of the extremities;[68,77,82] and may result in better alignment of the lower extremity, particularly the knee, thereby decreasing the risk of injury.[12,45,230] During tasks that involve jumping, landing, or change of direction, the trunk extensors must contract eccentrically to decelerate the trunk prior to initial contact and during weight acceptance. An inability to decelerate the trunk segment center of mass in preparation for initial contact and throughout the landing phase could result in decreased knee flexion as the trunk is displaced anteriorly, potentially outside of the base of support.

The relationship between trunk extension strength and knee kinematics in the sagittal plane has not been examined previously. Based on the results of the current study, it appears that trunk extension strength plays an important role in knee kinematics in the sagittal plane at initial contact. More specifically, increased trunk extension strength is related to an increase in knee flexion angle at initial contact. It is possible that increased trunk extension strength allows for the trunk to better positioned over the base of support, thereby promoting better alignment of the lower extremity.

5.3.3.2 Trunk Proprioception

Previous literature has examined the relationship between trunk proprioception in the transverse plane and knee injury. Female athletes who sustained a knee injury or a knee ligament/meniscal injury demonstrated significantly worse proprioception (passive placement-active replacement).[51] For every one degree decrease increase in repositioning error, there was a 2.9-fold increase of knee injury and a 3.3-fold increase of ligament/meniscal injury in male and female athletes. In addition, lateral, extension, and flexion trunk angular displacements following a sudden release were significantly greater in athletes who sustained a knee injury, knee ligament injury, or ACL injury as compared to uninjured athletes.[52]

In the current study, trunk flexion active joint position sense was not found to be a significant predictor of knee flexion angle at initial contact. Previous research has demonstrated that proprioceptive deficits in the transverse plane are related to risk of knee injury.[51,52] Based on the results of previous research and those of the current study, knee injury risk may be more influenced by deficits in trunk proprioception in the transverse plane. However, since previous studies did not assess knee kinematics and the current study did not follow subjects prospectively for knee injury, it is not possible to definitively state if this is accurate.

Lateral, flexion, and extension trunk angular displacement following a sudden release were found to be significantly greater in athletes who sustained a knee, knee ligament, or ACL injury as compared to uninjured athletes.[52] Trunk proprioception in the sagittal plane was not found to be a predictor of knee flexion angle at initial contact in the current study. In the current study, proprioception was assessed using an active positioning-active repositioning task, which may bias towards slow adapting mechanoreceptors (e.g., Ruffini endings, Golgi tendon-like organs).[124] The other study used trunk angular displacement following sudden release, which targets quick adapting mechanoreceptors (e.g., Pacinian corpuscles).[59,60,124] Further, due to the dynamic nature of task used in the other study it is very likely that muscle characteristics, such as muscle activation patterns, play a role in displacement. While a task such as this may assess overall neuromuscular control and is a predictor of knee injury, it does not help identify which component that contributes to neuromuscular control is deficient, making it difficult to develop targeted interventions.

5.3.3.3 Trunk Kinematics

Previous research has investigated change in knee kinematics relative to trunk kinematics in the sagittal plane. Blackburn and Padua[94] reported significant increases in knee flexion angle at initial contact when subjects intentionally landed with greater trunk flexion during a double leg drop landing. In contrast, Kulas et al.[49] reported no significant differences in knee flexion angle at initial contact during landing with and without trunk load, regardless if the trunk was in a relatively more or less flexed position. However, landing in a relatively extended trunk position with a trunk load significantly increased peak and average knee anterior shear forces and significantly decreased average hamstring force. Conversely, landing in a relatively flexed position resulted in no significant changes in knee anterior shear forces, but did result in a significant increase in average hamstring force.

In the current study, trunk flexion angle at initial contact was not a significant predictor of knee flexion angle at initial contact. The lack of agreement across studies could be due to methodological differences. In the study by Blackburn and Padua,[94] subjects were instructed to intentionally increase trunk flexion angle during landing. This may have resulted in conscious or subconscious alteration of knee position during landing. No instructions on landing were provided in the current study or in that by Kulas et al.[49] It also is possible that the tasks performed (double-leg drop landing vs. anticipated jump-and-cut maneuver) may account for some of this discrepancy.

Based on previous research, it may be that increasing trunk flexion angle may be more important in increasing average hamstring force and decreasing knee anterior shear forces, as demonstrated by Kulas et al.,[49] rather than increasing knee flexion angle. By increasing trunk flexion angle, the hamstrings may be placed in a more advantageous position. However, this could not be confirmed or refuted in the current study as muscle activation and knee anterior shear force were not assessed.

Another reason for lack of agreement may be how trunk angle was defined. Blackburn and Padua[94] calculated the trunk relative to the thigh and Kulas et al.[49] calculated the trunk relative to a vertical line in the global coordinate system. In the current study, trunk angle was defined at the position of the trunk relative to the pelvis.

5.3.3.4 Sex

Video analysis of ACL injury events has been used to assess knee kinematics during ACL injury events.[41,42,44] While Krosshaug et al.[42] reported that females who sustained an ACL injury demonstrated significantly greater knee flexion angle at initial contact than their male counterparts (15° vs. 9°, respectively), others have reported no difference between sexes.[41,44] Similarly, conflicting results have been reported in laboratory studies that have examined sex differences in knee flexion angle at initial contact. No significant differences were reported between males and females during a cutting maneuver in middle and high school basketball players,[96] state-level athletes,[236] or collegiate soccer players.[235] However, Malinzak et al.[238] reported that female collegiate recreational athletes demonstrated significantly less knee flexion than their male counterparts across three athletic tasks. In contrast, Fagenbaum and Darling[239] reported that female collegiate basketball players landed with significantly greater knee flexion during three different jump landing tasks at initial contact than male players.

In the current study, sex was found to be a significant predictor of knee flexion angle at initial contact. Female subjects landed with significantly less knee flexion than male subjects. One possible explanation for these discrepancies is that previous research has used two-dimensional analysis of video footage of actual ACL injury events.[41,42,44] Knee kinematics recorded in the laboratory during events that do not result in ACL injury may not be identical to those that occur during actual injury events. It also is possible that differences in physical activity may explain conflicting results across studies. The subjects in the current study were college-aged (18-25 years old, inclusive) individuals who were physically active a minimum of one day per week in sports or activities that involved jumping, cutting, or running. Previous studies that reported no differences in knee flexion angle at initial contact between sexes used middle school or high school athletes,[96] state-level athletes,[236] or intercollegiate/professional athletes[41,44,235] It is possible that neuromuscular control may be more highly trained in the athletes than in the physically active adults used in the current study. Malinzak et al.[238] used collegiate recreational athletes and reported differences between males and females in agreement with the findings of the current study, which further supports that training may influence the relationship between sex and knee flexion angle at initial contact during athletic tasks.

5.3.4 Maximum Knee Flexion Angle

The multiple linear regression analysis for maximum knee flexion angle produced a model that contained maximum spine flexion angle, trunk extension average peak torque, and sex. However, the model only explained 6.8% of the variance in maximum knee flexion angle and was not statistically significant; therefore the original hypothesis was not supported. It was hypothesized that greater maximum knee flexion angle would be predicted by greater trunk extension muscular strength, lower (better) trunk flexion proprioception, greater maximum trunk flexion angle, and male sex.

5.3.4.1 Trunk Extension Strength

The core musculature controls and positions the trunk relative to the lower extremity.[9] In order to provide a stable foundation for movement of the extremities,[68,77,82] the core musculature must effectively control and position the trunk, thereby allowing force and motion to be created and transferred to the extremities.[9] More effective control and positioning of the trunk may result in better alignment of the lower extremity, particularly the knee, thereby decreasing the risk of injury.[12,45,230] During many athletic tasks, the trunk extensors must contract eccentrically to decelerate the trunk prior to initial contact and during weight acceptance. An inability to decelerate the trunk segment center of mass in preparation for initial contact and throughout the landing phase could result decreased knee flexion angle during weight acceptance.

The relationship between trunk extension strength and knee kinematics in the sagittal plane has not been examined previously. Based on the results of the current study, it appears that trunk extension strength is not related to maximum knee flexion angle during weight acceptance when performing a SJCM. In the current study, trunk extension strength was assessed concentrically. Since the trunk extensors must contract eccentrically to decelerate the trunk during weight acceptance, eccentric strength of the trunk extensors may be of greater importance for knee kinematics in the sagittal plane during weight acceptance. In addition, other muscle characteristics of the trunk extensors, such as muscle activation patterns, may be a better predictor of knee flexion angle.

5.3.4.2 Trunk Proprioception

Deficits in proprioception in the transverse plane have been prospectively identified as a risk factor for knee or knee ligament/meniscal injury in female athletes.[51] In addition, regardless of sex, for every one degree increase in trunk repositioning error, there is a 2.9-fold increase of knee injury and a 3.3-fold increase of ligament/meniscal injury. In a follow-up study, it was reported that greater trunk angular displacement following a sudden release, worse trunk proprioception, and previous history of low back pain were significant predictors of knee ligament injury in female athletes.[52]

In the current study, trunk flexion active joint position sense was not found to be a significant predictor of maximum knee flexion angle. While the current study assessed trunk proprioception in the sagittal plane and its relationship to knee kinematics, previous research assessed trunk proprioception in the transverse plane and its relationship with knee injury.[51,52] Based on these collective results, transverse plane trunk proprioception may be more influential than sagittal plane proprioception in terms of risk of knee injury and/or risky knee kinematics. It is, however, not possible to conclusively state that this is true as knee kinematics and injury were not assessed concurrently in the previous studies or in the current study.

In a previous study, it was reported that athletes who sustained a knee, knee ligament, or ACL injury demonstrated significantly greater lateral, extension, and flexion trunk angular displacements following a sudden release as compared to uninjured athletes.[52] In contrast, trunk proprioception in the sagittal plane was not found to be a predictor of maximum knee flexion angle in the current study. Methodological differences between the studies may account for this discrepancy. While in the current study proprioception was assessed using an active positioning-active repositioning task, trunk displacement following sudden release was used in the other. It may be that the former targets the slow adapting mechanoreceptors (e.g., Ruffini endings, Golgi tendon-like organs)[124] whereas the latter targets quick adapting mechanoreceptors (e.g., Pacinian corpuscles).[59,60,124] In addition, trunk displacement following a sudden release most likely assesses overall neuromuscular control (e.g., proprioception, muscle activation, neural pathways) rather than isolating one component (i.e., proprioception) as with the task used in the current study.

5.3.4.3 Trunk Kinematics

Previous research has investigated change in knee kinematics relative to trunk kinematics in the sagittal plane. Blackburn and Padua[94] reported significant increases in maximum knee flexion angle when subjects intentionally landed with greater trunk flexion during a double leg drop landing. In contrast, Kulas et al.[49] reported no significant differences in maximum knee flexion angle during landing with and without trunk load, regardless if the trunk was in a relatively more or less flexed position. When landing in a relatively flexed position, knee anterior shear forces were not significantly changed, but average hamstring force was significantly increased. In contrast, landing with a relatively extended trunk position significantly increased peak and average knee anterior shear forces and significantly decreased average hamstring force.

Maximum trunk flexion angle was not significant predictor of maximum knee flexion angle in the current study. The lack of agreement across studies could be related to differences in the tasks performed and instructions provided to the subjects. In both of the previous studies,[49,94] a double-leg drop landing was performed: in the current study a jump-and-cut maneuver was used. Blackburn and Padua[94] instructed subjects to intentionally increase trunk flexion angle during landing, which may have resulted in subjects consciously or subconsciously altering knee position as well. No instructions on landing were provided in the current study or by Kulas et al.[49] However, increasing trunk flexion angle may be more important in increasing average hamstring force and decreasing knee anterior shear forces rather than increasing knee flexion angle.[49] This cannot be confirmed or refuted in the current study as muscle activation and knee forces were not assessed in the current study.

Differences in defining trunk angle also may account for discrepancies across studies. In one study, trunk angle was calculate as the trunk relative to the thigh.[94] In the other, the trunk angle was calculated relative to a vertical line in the global coordinate system.[49] In the current study, trunk angle was defined at the position of the trunk relative to the pelvis. The maximum trunk flexion angle relative to the pelvis (32.32 ± 10.75°) was less than that when calculated relative to vertical in the global coordinate system (39.65 ± 9.12°); therefore, it may be of benefit to look at trunk position relative to vertical in the global coordinate system.

5.3.4.4 Sex

Video analysis of ACL injury events has been used to assess knee kinematics during ACL injury events.[41,42,44] Krosshaug et al.[42] reported that females who sustained an ACL injury demonstrated significantly greater maximum knee flexion angle than their male counterparts (27° vs. 19°, respectively). However, no sex differences in maximum knee flexion angle were reported by other authors using similar techniques.[41,44] Laboratory studies also have produced conflicting results. No significant differences were reported between males and females during a cutting maneuver in middle and high school basketball players,[96] state-level athletes,[236] or collegiate soccer players.[235] However, female collegiate recreational athletes[238] and collegiate basketball players[216] demonstrated significantly less maximum knee flexion angle than their male counterparts. It is possible that the discrepancy between the results of these two studies and the results of other previous studies is due to the use of several athletic tasks rather than just a single task. In the current study, sex was not found to be a predictor of maximum knee flexion. Overall, this is in agreement with the majority of previous research.

5.4 LIMITATIONS

There are several limitations to the current study. The subjects selected to participate in this study were young, healthy males and females with physical activity levels of I or II on the Noyes Sports-Activities Rating Scale. The frequency of participation varied greatly among the subjects, with some subjects on participating two days per week and others participating seven days per week. It is possible that the neuromuscular and kinematic characteristics may be different in other populations that are involved more frequently and at a higher intensity in activities that involve jumping, landing, planting, or cutting.

The SJCM was a planned task, with the landing foot and the direction of the cut known before performing the task. Since this task started from a stationary position and the direction of the cut was known, it is possible that subjects were able to intentionally position the knee and the trunk during landing and transition into the cut, thereby influencing both knee and trunk kinematics. It is plausible that an unanticipated plant-and-cut task with or without a running approach may better replicate knee and trunk kinematics that occur during athletics.

In the current study, fairly stringent guidelines were utilized to determine what constituted an acceptable SJCM trial. Trials were repeated if both the position of the foot and the pelvis at initial contact deviated by more than ten degrees relative to the start position. This may have resulted in some subjects not performing the task as they would in a less artificial environment. It is possible that the trunk and knee kinematics in these discarded tasks may be more representative of what occurs outside of the laboratory and, potentially, of the "risky" kinematics that may be associated with non-contact ACL injury. In addition, trunk and knee kinematics were averaged across three trials. By averaging across trials, we may be losing the natural variability seen in human movement. For example, when looking at the range for knee flexion angle at initial contact within subjects, the minimum range was 0.85° across trials while the maximum range was 20.59° across trials. Future research should look at the kinematics within individual trials as variability in movement (increased or decreased) may be related to injury risk.

Strength was assessed using isokinetic testing at a slow speed and average peak torque was selected as the isokinetic variable to be utilized in the regression equations. This was to ensure that data could be collected reliably and results could be compared to previous research. It is possible that a different mode of exercise (i.e., eccentric contractions) may be more similar to the type of muscle contractions utilized during dynamic athletic tasks. Further, since human movement of this nature typically occurs at much higher speeds, it may be more beneficial to look at strength at a higher speed when using isokinetic testing. Lastly, it may be that other muscle characteristics such as muscle activation patterns or time to peak torque may be more closely related to one's ability to control the trunk during dynamic movement.

There are a few reasons why, in general, the R^2 values seen in the current study were relatively low. It is possible that not all of the important predictors of knee kinematics were included in the final model. These may include other neuromuscular characteristics of the hip and thigh as well as kinematics of the hip. Another explanation could be that the sample size was too low. However, the current study was adequately powered based on the given effect size and the number of predictors in the final models.

5.5 CLINICAL SIGNIFICANCE

Despite years of research attempting to identify modifiable risk factors for non-contact ACL injury and the development of ACL injury prevention programs, these injuries persist, as demonstrated by a statistically significant increase in the rate of ACL injury at the collegiate level over a 16 year period.[1] Non-contact ACL injuries impose significant burdens, such as the associated medical costs, potential loss of scholarship or salary, decreased quality of life,[29,34] and the development of osteoarthritis.[30,31] Core stability is promoted by healthcare and fitness professionals to be essential for injury prevention; however, there is limited research examining the relationship between the neuromuscular characteristics that underlie trunk functional stability and knee kinematics that have been identified through prospective studies and video analysis to be related to ACL injury.

The results of the current study indicated that knee valgus angle at initial contact and maximum knee flexion angle could not be predicted by any of the independent variables included in their respective regression models. However, the results demonstrated that both total knee valgus excursion and knee flexion angle at initial contact were related to some of the independent variables. More specifically, total knee valgus excursion was predicted by non-dominant trunk rotation (rotation away from the dominant leg) average peak torque and sex. Together, lower non-dominant trunk rotation strength and female sex predicted greater total knee valgus excursion. In addition, knee flexion angle at initial contact was predicted by trunk extension average peak torque and sex. In combination, lower trunk extension strength and female sex predicted lower knee flexion angle at initial contact.

Previous prospective research has demonstrated that healthy athletes who sustained a non-contact ACL injury[36] and ACL-reconstructed athletes who sustained a second non-contact injury[37] demonstrated greater maximum knee abduction angle and greater knee valgus motion, respectively. In addition, video analysis of ACL-injury events have described the knee to be in a relatively extended position at initial contact[42] and that the progresses into significantly greater knee abduction[41,42,44] in individuals who sustain an ACL injury as compared to controls. These findings from previous studies, when combined with the results of the current study, provide evidence of the importance of core strength, particularly trunk rotation and extension strength, in an attempt to promote more favorable knee kinematics during jumping and cutting tasks, which, in turn, may reduce the risk of non-contact ACL injury.

5.6 FUTURE RESEARCH

Future research should investigate how neuromuscular characteristics across several joints/segments (e.g., knee and lumbopelvic-hip complex) collectively contribute to knee joint kinematics. Prospective studies also should be implemented to examine if neuromuscular characteristics, such as deficits in trunk strength, predict non-contact ACL injury and to determine if intervention programs focused on improving trunk rotation and extension strength decrease the risk of non-contact ACL injury. Also, research should explore how variability in trunk and knee kinematics is related to injury risk as well as if greater trunk motion assists in the dissipation of forces during landing. Lastly, it may be important to look at trunk position relative to the global coordinate system rather than the pelvis.

5.7 CONCLUSIONS

The purpose of this study was to determine if trunk muscular strength, proprioception, and kinematics as well as sex could predict knee kinematics during a stop jump-cut maneuver (SJCM). Knee kinematics included knee valgus and knee flexion angles at initial contact, total knee valgus excursion, and maximum knee flexion angle. Multiple linear regression models were developed to investigate the relationship between each dependent variable and their respective independent variables. Results of the regression models in the current study revealed that, together, non-dominant trunk rotation (rotation away from the dominant leg) average peak torque and sex were significant predictors of total knee valgus excursion, accounting for 25.9% of the variability in total knee valgus excursion during a SJCM. In addition, trunk extension average peak torque and sex together were significant predictors of knee flexion angle at initial contact, accounting for 25.4% of the variability in knee flexion at initial contact during a SJCM. These findings indicate that trunk strength plays an important role in knee kinematics during a jumping and cutting task.

APPENDIX A

REGRESSION EQUATIONS: INDEPENDENT VARIABLES AND RATIONALE

1. KNEE VALGUS ANGLE AT INITIAL CONTACT

Independent Variables	Rationale
Strength (peak torque in Nm) • Isokinetic trunk extension strength	• Trunk extensor strength is critical in decelerating the trunk segment center of mass at landing. This is an indirect relationship. If the trunk is not decelerated adequately, then there will be less knee flexion at initial contact which will result in increased knee valgus to attenuate forces.
Proprioception (absolute error in degrees) • Right lateral flexion active joint position sense for right leg dominant ***OR*** • Left lateral flexion active joint position sense for left leg dominant Kinematics (degrees) • Lateral trunk displacement angle at initial contact during the SJCM	• When the trunk is displaced laterally relative to the support leg during single leg stance or landing, the vGRF passes lateral to the knee joint center resulting in increased knee valgus. The greater the lateral displacement of the trunk, the greater the potential for knee valgus.
Sex • Female	• ACL research has demonstrated that females land with greater knee valgus

2. TOTAL KNEE VALGUS EXCURSION

Independent Variables	Rationale
Strength (peak torque in Nm) • Isokinetic right trunk rotation strength for left leg dominant ***OR*** • Isokinetic left trunk rotation strength for right leg dominant	• Subjects will be cutting away from the dominant leg. Inadequate trunk rotation strength into this direction may result in the trunk "trailing" the motion, resulting in relative internal rotation and dynamic valgus collapse.
Proprioception (absolute error in degrees) • Right lateral flexion active joint position sense for right leg dominant ***OR*** • Left lateral flexion active joint position sense for left leg dominant Kinematics (degrees) • Maximum lateral trunk displacement angle during the SJCM	• When the trunk is displaced laterally relative to the support leg during single leg stance or landing, the vGRF passes lateral to the knee joint center resulting in increased knee valgus. The greater the lateral displacement of the trunk, the greater the potential for knee valgus.
Sex • Female	• ACL research has demonstrated that females demonstrate dynamic knee valgus collapse at the time of ACL injury

3. KNEE FLEXION ANGLE AT INITIAL CONTACT

Independent Variables	Rationale
Strength (peak torque in Nm) • Isokinetic trunk extension strength	• Trunk extensor strength is critical in decelerating the trunk segment center of mass at landing. If the trunk is not decelerated adequately, then there will be less knee flexion at initial contact which will result in the inability to attenuate forces.
Proprioception (absolute error in degrees) • Trunk flexion active joint position sense Kinematics (degrees) • Trunk flexion angle at initial contact during the SJCM	• The ability to correctly position and control the trunk in the sagittal plane relative to the base of support will influence knee sagittal plane motion. Increasing trunk flexion during landing has been demonstrated to increase knee flexion.
Sex • Male	• Males land with greater knee flexion at initial contact

4. MAXIMUM KNEE FLEXION ANGLE

Independent Variables	Rationale
Strength (peak torque in Nm) • Isokinetic trunk extension	• Trunk extensor strength is critical in decelerating the trunk segment center of mass at landing. If the trunk is not decelerated adequately, then there will be less knee flexion at initial contact which will result in increased knee valgus to attenuate forces.
Proprioception (absolute error in degrees) • Trunk flexion active joint position sense Kinematics (degrees) • Maximum trunk flexion angle during the SJCM	• The ability to correctly position and control the trunk in the sagittal plane relative to the base of support will influence knee sagittal plane motion. Increasing trunk flexion during landing has been demonstrated to increase knee flexion.
Sex • Male	• Males demonstrate greater maximum knee flexion

APPENDIX B

NOYES SPORTS-ACTIVITIES RATING SCALE

Sports-Activities Rating Scale[210]

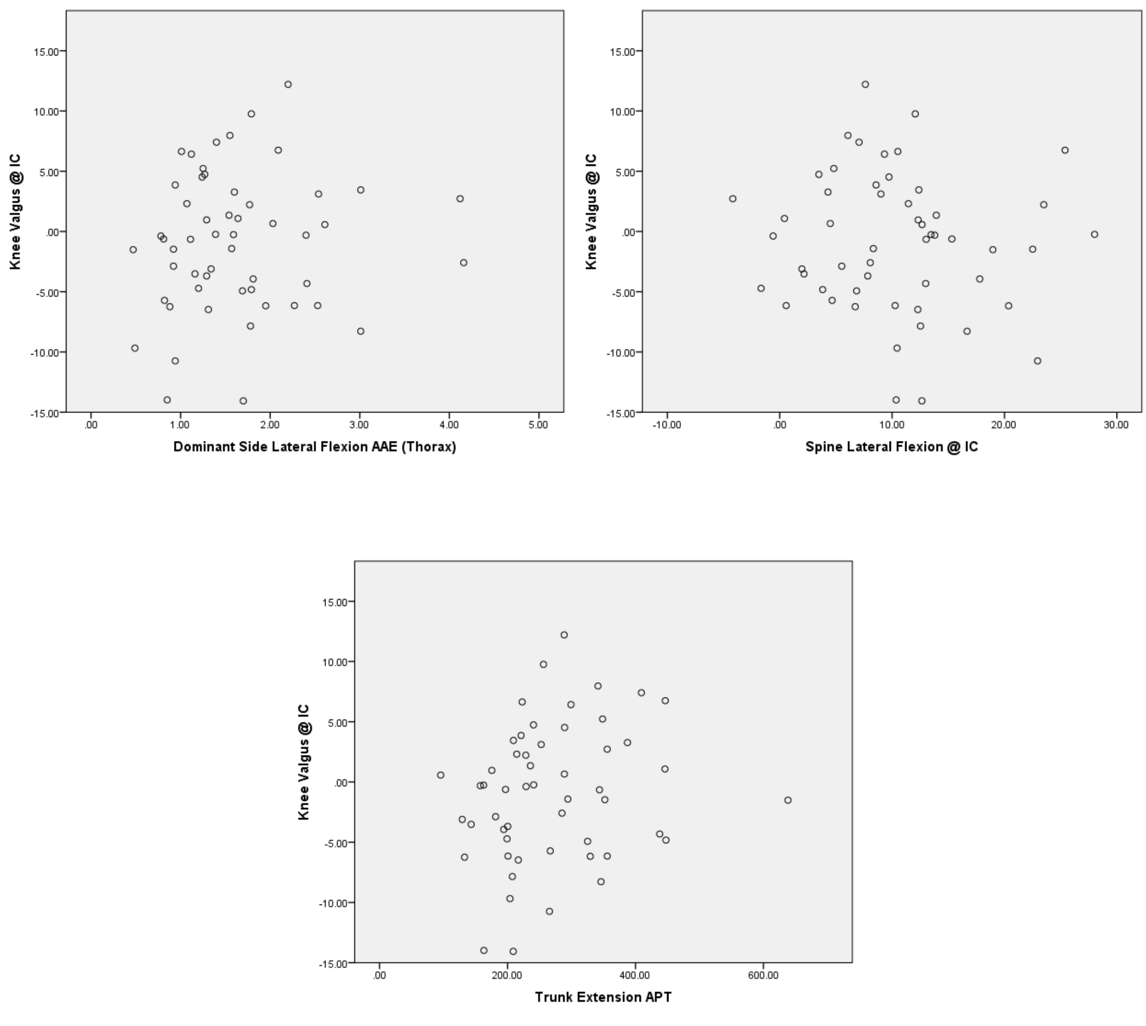
Level	*Activities*	*Points*
Level I	Participates 4-7 days per week	
	• Jumping, hard pivoting, cutting (basketball, volleyball, football, gymnastics, soccer)	100
	• Running, twisting, turning (tennis, racquetball, handball, baseball, ice hockey, field hockey, skiing, wrestling)	95
	• No running, twisting, jumping (cycling, swimming)	90
Level II	Participates 1-3 days per week	
	• Jumping, hard pivoting, cutting (basketball, volleyball, football, gymnastics, soccer)	85
	• Running, twisting, turning (tennis, racquetball, handball, baseball, ice hockey, field hockey, skiing, wrestling)	80
	• No running, twisting, jumping (cycling, swimming)	75

NOYES SPORTS-ACTIVITIES RATING SCALE (continued)

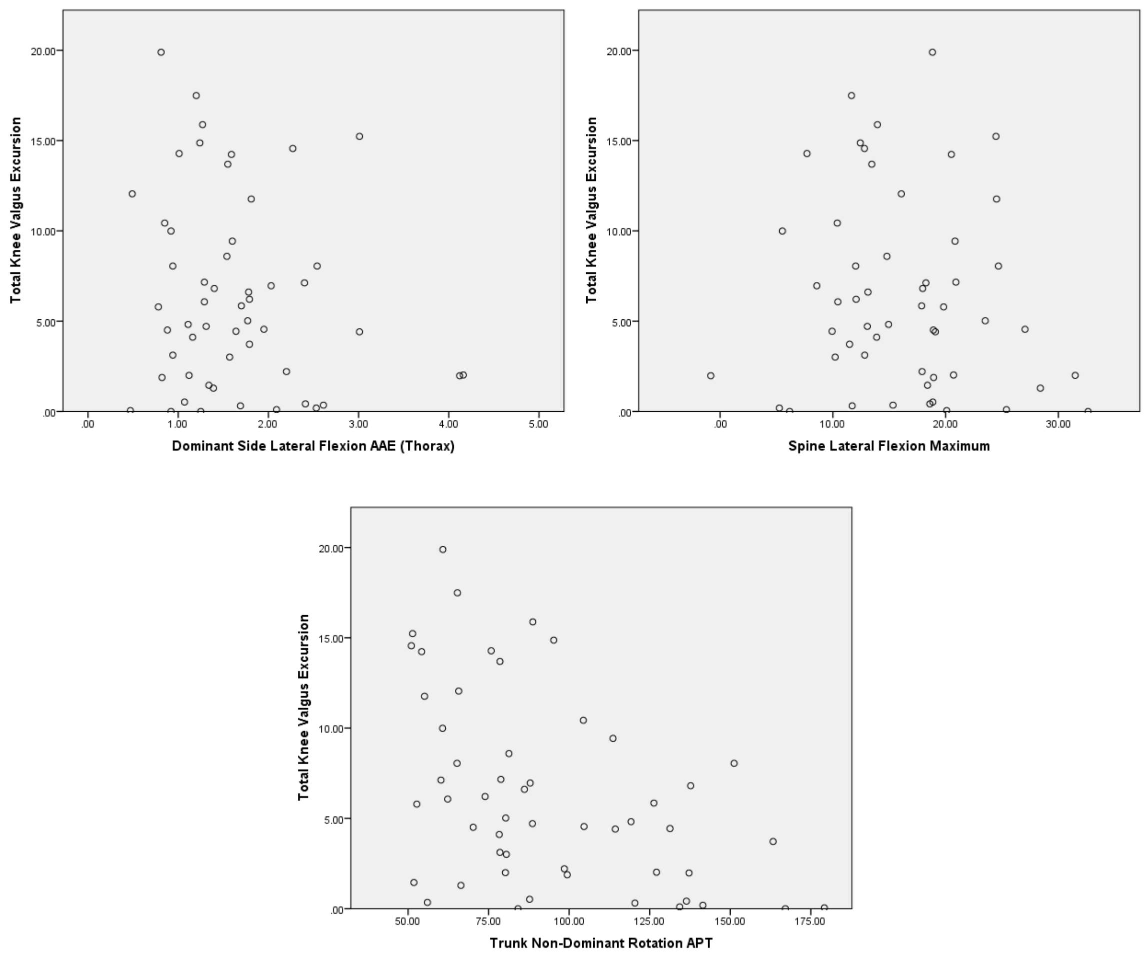
Level	*Activities*	*Points*
Level III	Participates 1-3 times per month	
	• Jumping, hard pivoting, cutting (basketball, volleyball, football, gymnastics, soccer)	65
	• Running, twisting, turning (tennis, racquetball, handball, baseball, ice hockey, field hockey, skiing, wrestling)	60
	• No running, twisting, jumping (cycling, swimming)	55
Level IV	No sports	
	• Performs activities of daily living without problems	40
	• Has moderate problems with activities of daily living	20
	• Has severe problems with activities of daily living- on crutches, fully disabled	0

APPENDIX C

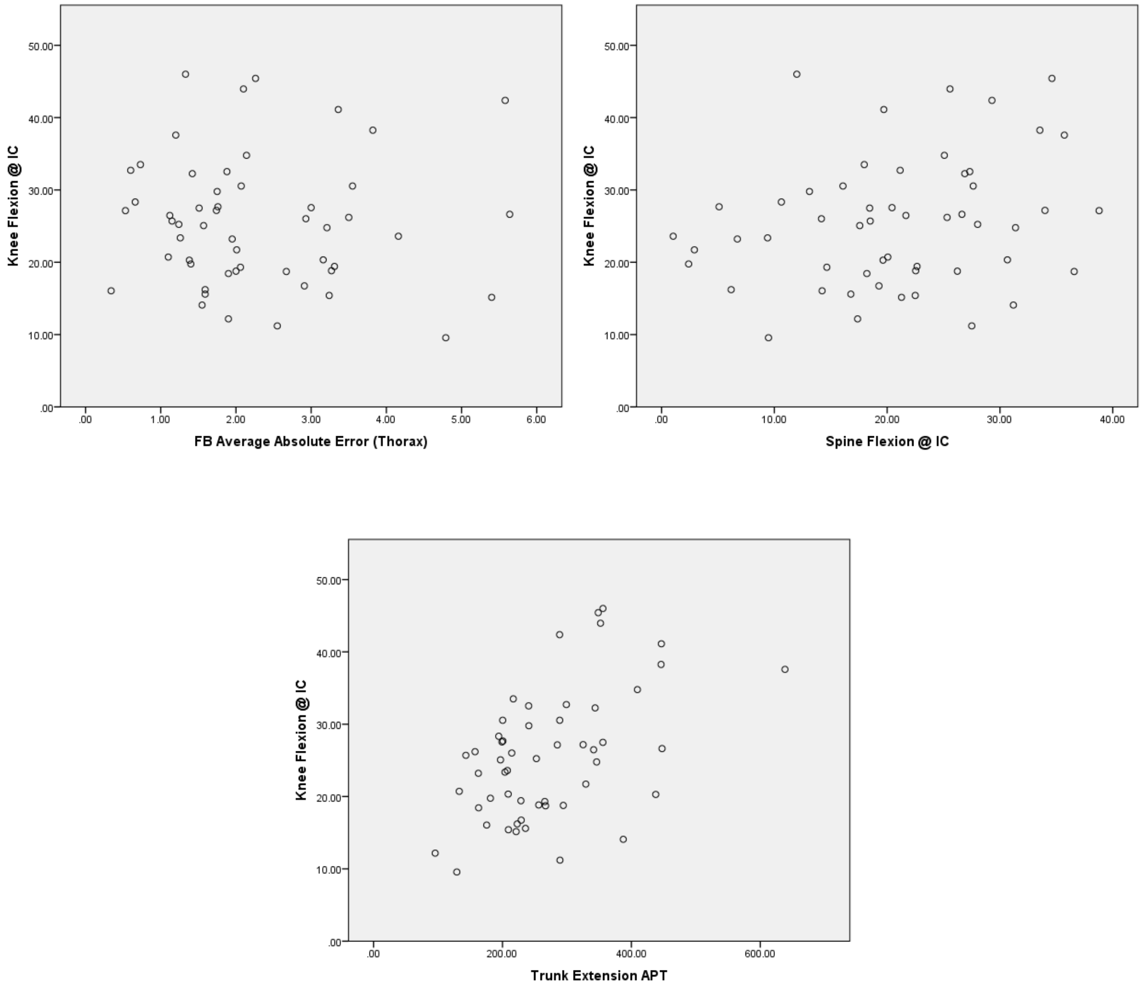
SCATTERPLOTS OF THE DEPENDENT AND INDEPENDENT VARIABLES



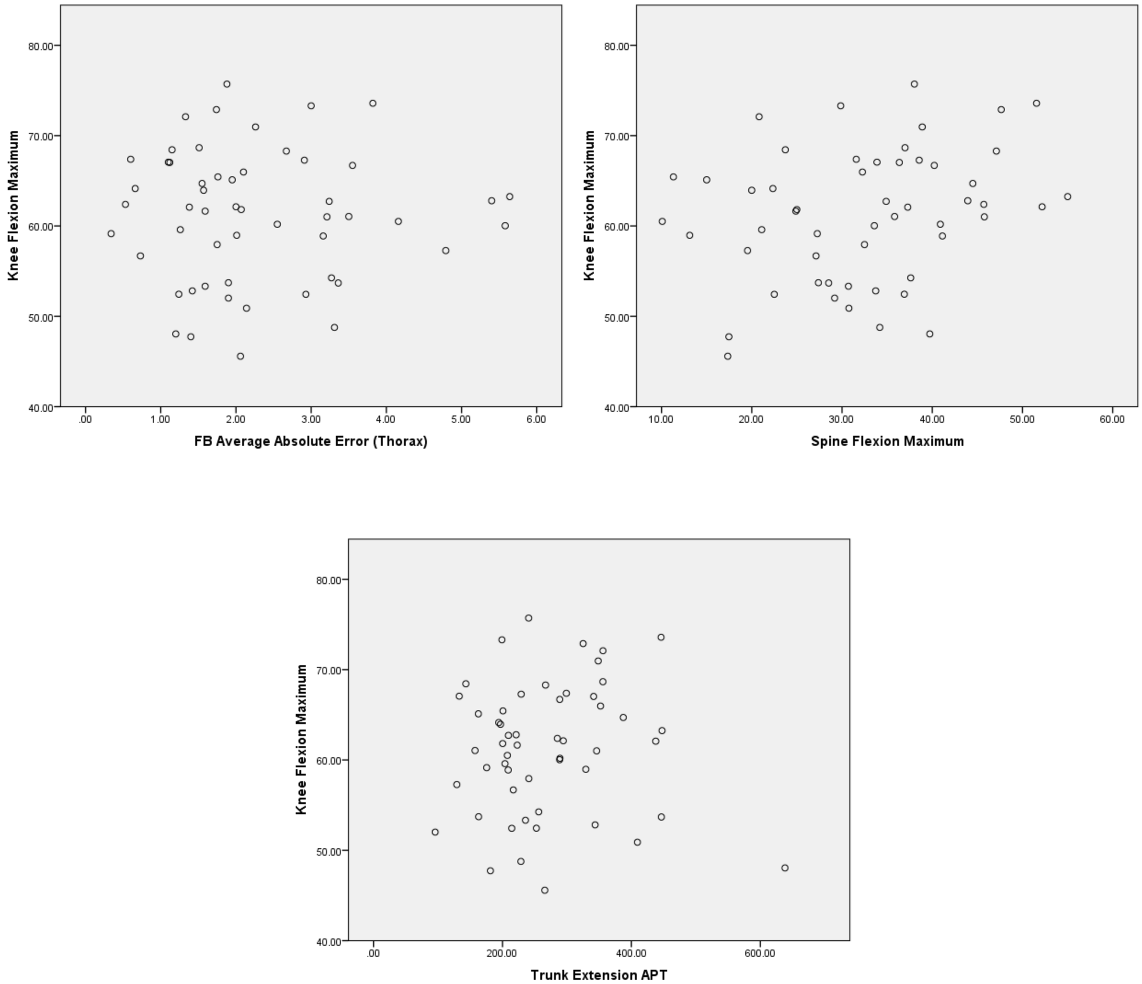
Knee Valgus Angle at Initial Contact and Its Respective Independent Variables



Total Knee Valgus Excursion and Its Respective Independent Variables



Knee Flexion Angle at Initial Contact and Its Respective Independent Variables



Maximum Knee Flexion Angle and Its Respective Independent Variables

APPENDIX D

CORRELATION TABLE: DEPENDENT AND INDEPENDENT VARIABLES

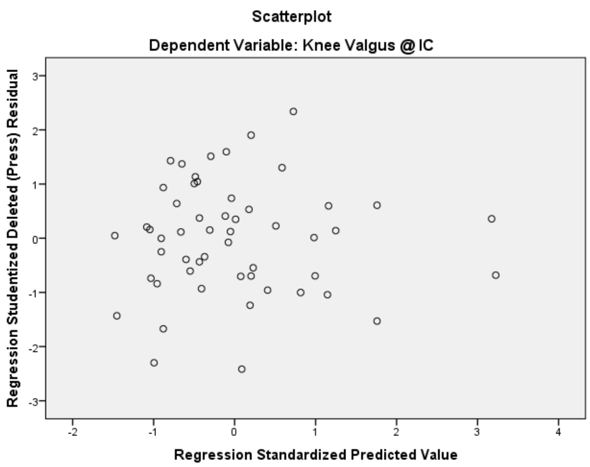
	Knee Valgus @ IC	Total Knee Valgus Excursion	Knee Flexion @ IC	Maximum Knee Flexion	Forward Flexion AAE	Dominant Lateral Flexion AAE	Spine Flexion @ IC	Spine Lateral Flexion @ IC	Maximum Spine Flexion	Maximum Spine Lateral Flexion	Trunk Extension APT	Trunk Non-Dominant Rotation APT
Total Knee Valgus Excursion	0.013 0.929											
Knee Flexion @ IC	0.159 0.256	-.307* 0.025										
Maximum Knee Flexion	0.026 0.853	0.112 0.425	0.260 0.060									
Forward Flexion AAE	0.133 0.341	-0.139 0.320	-0.044 0.752	-0.001 0.996								
Dominant Lateral Flexion AAE	0.099 0.479	-0.169 0.226	0.058 0.678	0.106 0.448	0.043 0.758							
Spine Flexion @ IC	0.201 0.149	-.287* 0.037	0.250 0.071	0.202 0.147	0.126 0.368	0.149 0.285						

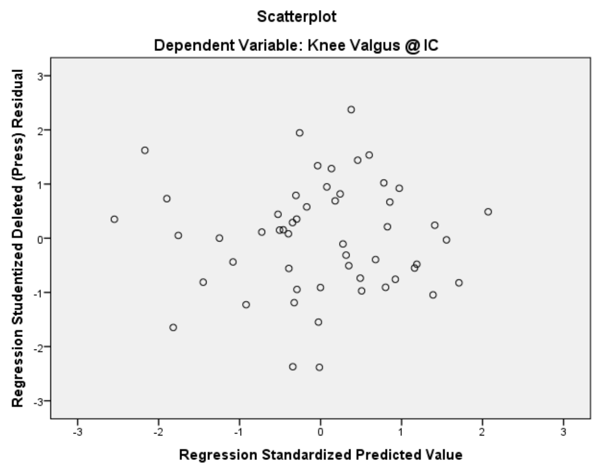
	Knee Valgus @ IC	Total Knee Valgus Excursion	Knee Flexion @ IC	Maximum Knee Flexion	Forward Flexion AAE	Dominant Lateral Flexion AAE	Spine Flexion @ IC	Spine Lateral Flexion @ IC	Maximum Spine Flexion	Maximum Spine Lateral Flexion	Trunk Extension APT	Trunk Non-Dominant Rotation APT
Spine Lateral Flexion @ IC	-0.081	-0.078	0.023	-0.567*	-0.097	-0.084	-0.171					
	0.562	0.581	0.870	0.000	0.490	0.548	0.220					
Maximum Spine Flexion	0.213	-.293*	0.026	0.265	0.231	0.176	0.861*	-0.257				
	0.126	0.033	0.855	0.055	0.096	0.207	0.000	0.063				
Maximum Spine Lateral Flexion	0.001	-0.111	-0.010	-0.210	-0.071	-0.132	-0.002	0.667*	-0.126			
	0.996	0.429	0.941	0.130	0.611	0.348	0.991	0.000	0.368			
Trunk Extension APT	0.204	-0.364*	0.498*	0.019	0.040	0.098	0.478*	0.055	0.453*	0.025		
	0.144	0.007	0.000	0.892	0.775	0.484	0.000	0.696	0.001	0.861		
Trunk Non-Dominant Rotation APT	-0.135	-0.460*	0.425*	-0.071	0.044	0.111	0.435*	0.104	0.351*	0.053	0.794*	
	0.336	0.001	0.002	0.611	0.755	0.429	0.001	0.458	0.010	0.707	0.000	
Sex	-0.030	0.486*	-0.474*	-0.178	-0.106	-0.161	-0.492*	0.102	-0.369*	-0.135	-0.676*	-0.732*
	0.834	0.000	0.000	0.203	0.449	0.250	0.000	0.465	0.006	0.334	0.000	0.000

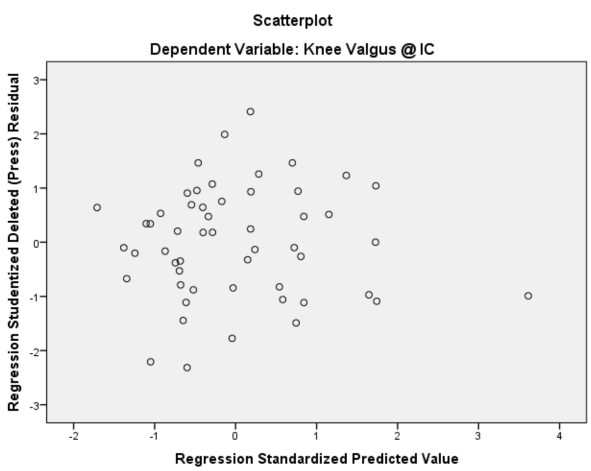
* Statistically significant correlation ($p<0.05$)

APPENDIX E

SIMPLE LINEAR REGRESSION: JACKKNIFE RESIDUALS VERSUS PREDICTED VALUES

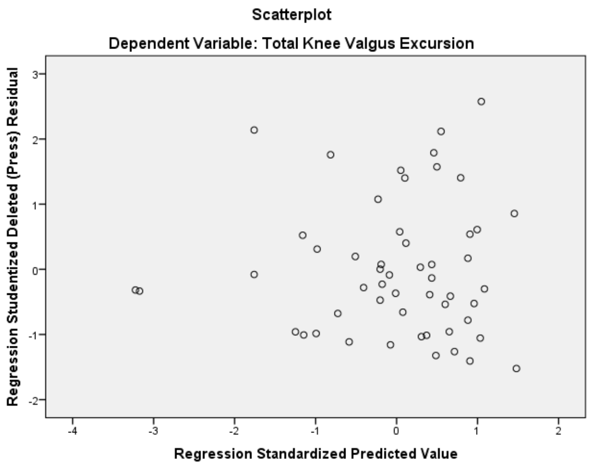


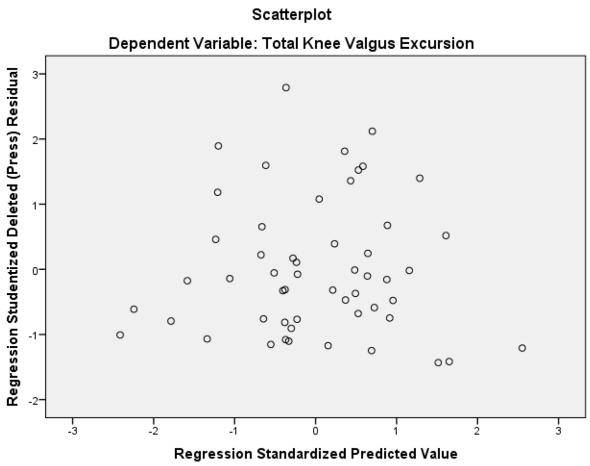


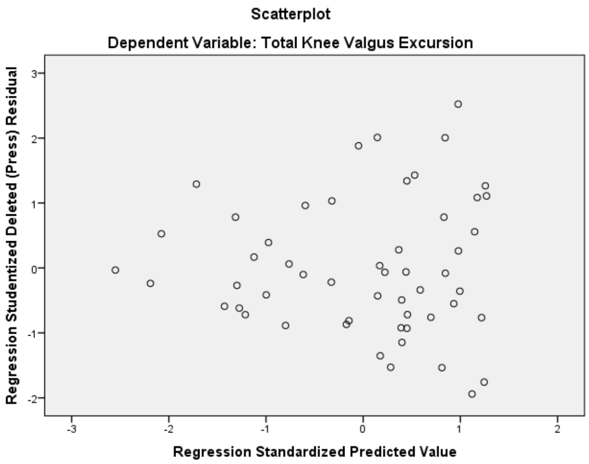


Knee Valgus Angle at Initial Contact and Independent Variables

a) Dominant side lateral flexion average absolute error
b) Spine lateral flexion angle at initial contact
c) Trunk extension average peak torque

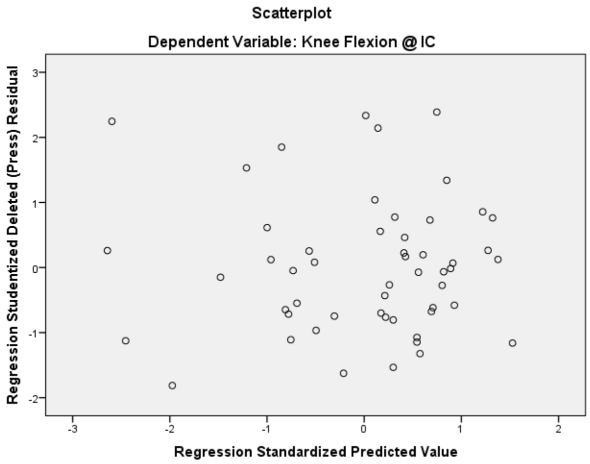


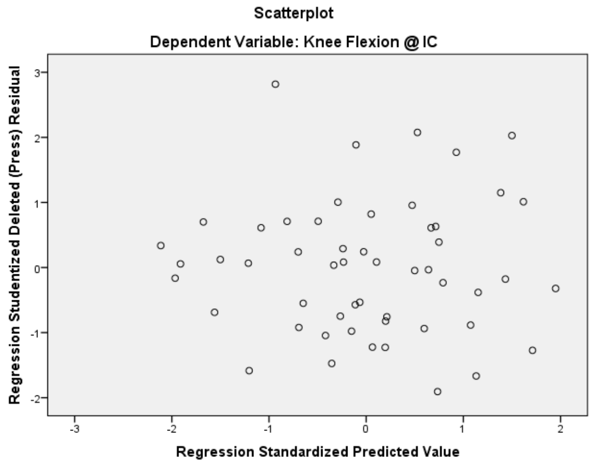


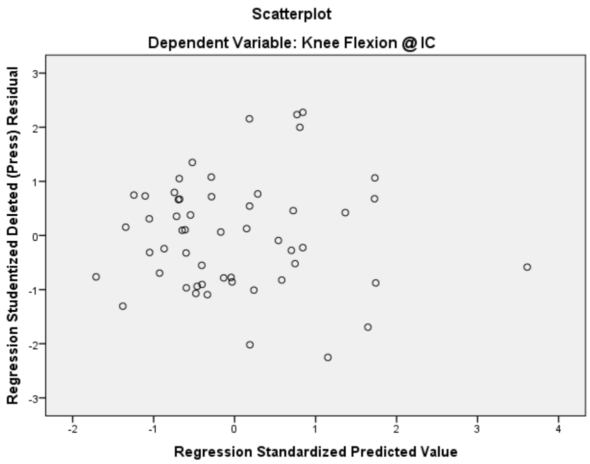


Total Knee Valgus Excursion and Independent Variables

a) Dominant side lateral flexion average absolute error
b) Maximum spine lateral flexion angle
c) Non-dominant trunk rotation average peak torque

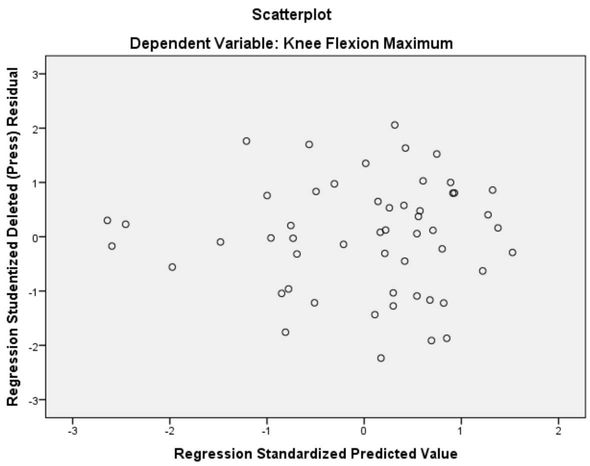


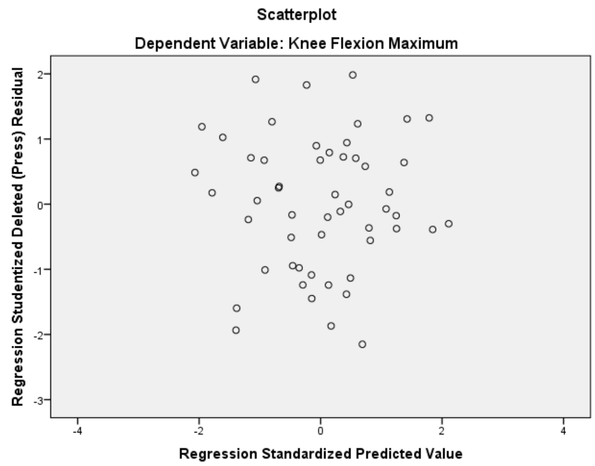


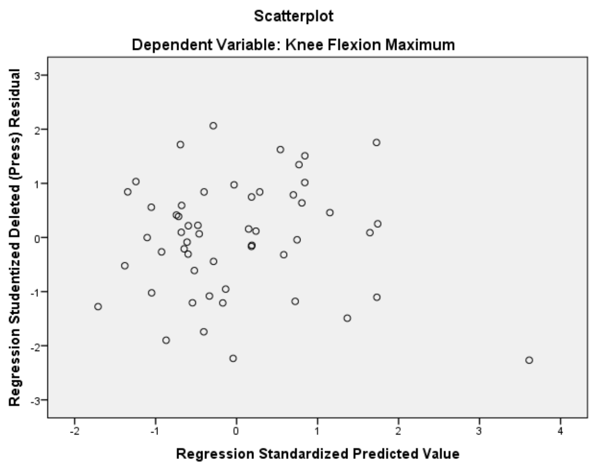


Knee Flexion Angle at Initial Contact and Independent Variables

a) Forward flexion average absolute error
b) Spine flexion angle at initial contact
c) Trunk extension average peak torque





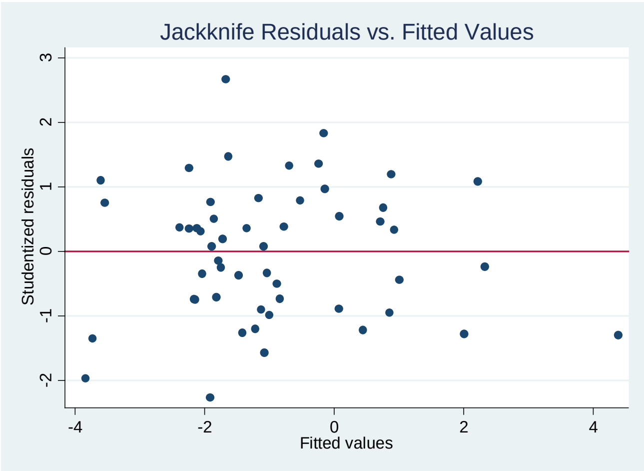


Maximum Knee Flexion Angle at Initial Contact and Independent Variables

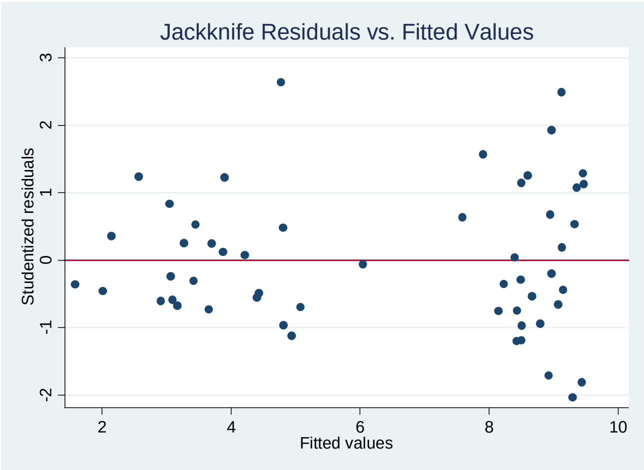
a) Forward flexion average absolute error
b) Maximum spine flexion angle
c) Trunk extension average peak torque

APPENDIX F

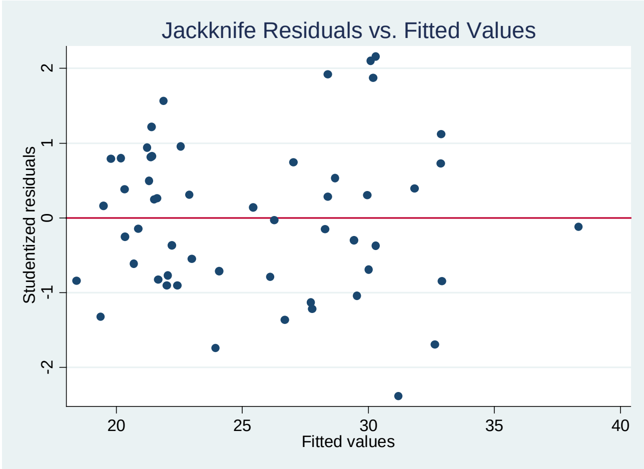
MULTIPLE LINEAR REGRESSION: JACKKNIFE RESIDUALS VERSUS PREDICTED VALUES



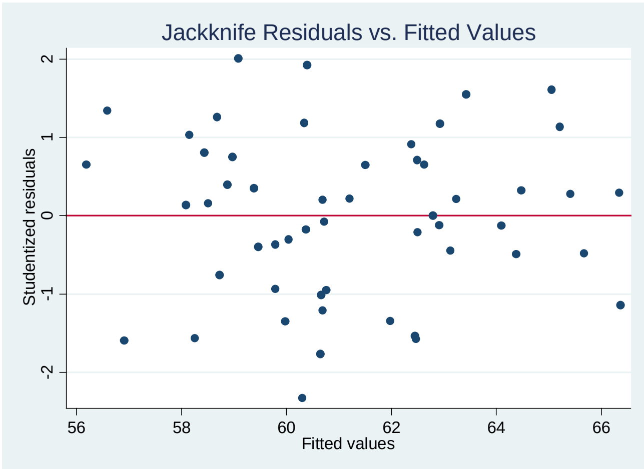
Knee Valgus Angle at Initial Contact



Total Knee Valgus Excursion



Knee Flexion Angle at Initial Contact



Maximum Knee Flexion Angle at Initial Contact

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