THE EFFECT OF JUMP DISTANCE ON BIOMECHANICAL RISK FACTORS FOR ACL INJURY DURING LANDING AND THEIR RELATIONSHIP WITH SENSORIMOTOR CHARACTERISTICS AT THE KNEE

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There has been an abundance of research investigating risk factors for anterior cruciate ligament (ACL) injury and demonstrating the importance of biomechanical characteristics, particularly in females. However, there have been many different landing tasks used with varying demands. Previous research has demonstrated that different landing tasks significantly alter demand and biomechanical characteristics. However it is unknown how changes in landing demand using the same task may alter landing biomechanics related to ACL injury. Therefore, the purpose of this study was to examine the effects of jump distance during a double-leg stop-jump on biomechanical risk factors of ACL injury and muscle activation and examine the contribution of sensorimotor characteristics on these biomechanical characteristics.

Fifty-three recreationally active healthy females were recruited to participate in this study. Each participant underwent a single test session that included demographic and anthropometric assessment, dominant knee threshold to detect passive motion, landing biomechanics and muscle activation measurement, and dominant knee time to peak torque and peak torque testing. Biomechanical and muscle activation parameters relative to ACL injury were compared between jump distances using repeated measures ANOVA. Multiple linear regression was used to assess the relationship between the biomechanical characteristics and sensorimotor characteristics (threshold to detect passive motion, time to peak torque, and peak torque).

The results of this study demonstrated that increases in jump distance significantly increased landing demand and significantly impacted risk factors for ACL injury and muscle activation strategies. These findings illustrated that studies utilizing tasks with different demands cannot directly compare results or make inference to injury risk based previous findings. This study suggested that a jump distance of 40% to 60% body height is used during a double-leg stop-jump task to assess landing biomechanics related to ACL injury. Additionally, sensorimotor characteristics had significant relationships with knee flexion angle at initial contact, peak knee flexion, and peak knee abduction moment. Further research is needed to identify sensorimotor characteristics that contribute to frontal plane knee motion during landing.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

PREFACE

Thank you to all who have helped me through this dissertation and my educational career, especially my wife and family.

1.0 INTRODUCTION

Recreational sports and activities are popular and are growing in interest throughout the United States and the world. The United States Department of Health and Human Services and the American College of Sports Medicine have published position statements and guidelines for the public that advocate the participation in physical activity for a healthy lifestyle.[1-4] However, an unfortunate consequence of sport and recreational activity is unintentional musculoskeletal injury. Almost one quarter of all American high school students report at least one injury due to sports participation. Overall, Americans sustain an estimated seven million musculoskeletal injuries attributed to sports and recreational activities annually.[5] These unintentional musculoskeletal injuries can be a significant obstacle for the continuation of an active lifestyle, potentially affecting the physical, mental, and social aspects of a person's life.[6-11] Some sports related musculoskeletal injuries are also associated with the possible development of long-term disability.[7-9] Due to the implications of sport and recreationally related unintentional musculoskeletal injuries, the prevention of such injuries is a serious concern among medical professionals as the number of people who participate in sports and subsequent injuries increase.[12]

Knee injuries are a particular concern within sports medicine due to the higher incidence and morbidity associated with them.[13-16] Injury to the anterior cruciate ligament (ACL) is a specific concern in sports because returning to a similar level of function after an injury to this ligament most often requires surgical intervention and a lengthy rehabilitation process.[6] In United

States high school sports the knee has the highest prevalence (53.9%) of injuries that require surgical treatment[17] and it is not uncommon for other knee pathologies to develop subsequent to the initial trauma, regardless of surgical or conservative treatment.[7] Patients who undergo ACL reconstructive surgery have a very high incidence of future knee joint osteoarthritis.[7,18-21] Some evidence even suggests that over 50% of patients who sustain injury to the ACL will develop knee osteoarthritis regardless of if reconstructive surgery is performed.[7,9] The morbidity associated with these injuries has created a burden on health care[7,8] and has established the need for extensive research to be done with the aim to minimize the risk of sustaining such an injury.

In an effort to determine appropriate methods for preventing ACL injuries research has been trying to target specific characteristics that may place athletes at greater risk.[22-25] Although a range of individual characteristics have been shown to be risk factors for this injury, one of the most widely focused on set of characteristics is landing biomechanics. This high interest in landing biomechanics related to ACL injury is most likely due to the ability to modify these characteristics through training.[26-28]

The evaluation of an athlete's movement quality, including the ability to land and change direction safely, is not only relevant to what has been observed as an injury mechanism in video footage,[29,30] but also can provide key information regarding the potential risk of sustaining an ACL injury.[23,31,32] Throughout literature there has been different tasks and variations of the same task used to evaluate landing biomechanics. However, these variations and differences impose a different demand on subjects, which may be a cause of some of the discrepancies seen in biomechanical risk factors for ACL injuries. Additionally, training programs targeting landing biomechanics look to challenge specific landing characteristics based on previous research. Previous research has demonstrated differences in kinematics and kinetics between different jump directions[33] and tasks but it is unknown how altering the demand of a landing task by jump distance will affect risk factors for ACL injury or if a linear relationship exists between landing demand and these risk factors. In order to target specific landing characteristics related to ACL injury it is necessary to determine appropriate biomechanical task demands to create the desired challenge. Therefore, in order to better standardize and appropriately use landing tasks to evaluate biomechanical characteristics related to ACL injury we must establish the effect of landing demand on such risk factors for injury. By determining how knee loading and the presentation of risk factors change as landing demand increases researchers and clinicians will be provided with evidence for more appropriate landing standardization for specific characteristics of interest.

1.1 ANTERIOR CRUCIATE LIGAMENT INJURY

1.1.1 Epidemiology of Anterior Cruciate Ligament Injuries

Anterior cruciate ligament injury is still among the most common serious lower extremity injuries suffered in sports today.[13,34] It has been estimated that approximately eighty thousand ACL injuries occur each year in the United States.[35] Data from the National Collegiate Athletic Association's (NCAA) injury surveillance system has reported an average of over three hundred ACL injuries each year in their sample of only 15% of all NCAA athletes.[13] Although American football has some of the highest rates of ACL injuries reported, other sports such as soccer, basketball, gymnastics, volleyball, and handball are more commonly investigated due to a higher amount of non-contact ACL injuries, which are thought to be more preventable.[13,14,34,36,37] Similarly, epidemiological studies investigating the effect of gender have demonstrated increased risk of non-contact ACL injuries in females compared to males in comparative sports.[13,14,34,38] Although there has been a proliferation of risk factor and interventional research regarding the prevention of non-contact ACL injuries in female athletes, there has been no documented decline in ACL injury rates in any large scale injury surveillance system, such as the NCAA. Injuries to the ACL are still a major problem and studies are still needed to investigate potential means of increasing the effectiveness of intervention methods.

1.1.2 Mechanisms for Non-Contact Anterior Cruciate Ligament Injury

Video analysis of non-contact ACL injuries suggests that a common mechanism for these injuries is an abrupt deceleration during a landing or change of direction maneuver with a sudden valgus knee collapse (knee abduction).[29,30,39] This valgus collapse mechanism seems to occur during the very early part of stance phase.[39,40] Although valgus knee displacement is not the most direct loading mechanism of the ACL, the combination of anterior tibial displacement and a abduction moment does produce the highest amount of strain in the ACL.[41] Biomechanical studies have also added that low knee flexion angle and increased posterior ground reaction force produce a high knee flexion moment which increases strain at the ACL via increases in quadriceps force.[42-45] Although these mechanical factors may increase strain at the ACL the effects of the sensorimotor system on landing position needs to be examined as the mechanism by which these factors can be changed.

1.2 SENSORIMOTOR SYSTEM

Riemann and Lephart[46] describe the sensorimotor system as the "sensory, motor, and central integration and processing components involved in maintaining joint homeostasis during bodily movements."[46] This maintenance of joint homeostasis can be referred to as functional joint stability.[46,47] Joint stability has been defined as "the state of a joint remaining in or promptly returning to proper alignment through an equalization of forces" and moments.[46] In order for this to occur, the human body must understand, react to, and prepare for it's external and internal environments. The success of this process is dependent on various forms of sensory input from the visual, vestibular, and somatosensory systems. The most important information from an orthopaedic and injury prevention perspective originates from the somatosensory component of the sensorimotor system. Mechanoreceptors within the cutaneous, muscular, joint, and ligamentous structures are responsible for the sensory information responsible for joint position sense and kinesthesia.[46,47] This afferent information arising from the peripheral areas is referred to as proprioception, a subcomponent of the sensorimotor system.[46]

1.2.1 Proprioception

The roles of proprioception in motor control and the maintenance of functional joint stability have previously been described in two distinct categories.[48] The first category of proprioception is "with respect to the external environment" in the form of perturbations, such as walking on an uneven surface.[48,49] These perturbations or unexpected environmental changes require a modification to the current motor process to maintain postural control and functional joint stability. The second category of proprioception is “the planning and modification of internally generated motor commands” with the goal of coordinating complex mechanical interactions of specific joint motions.[48,49] Proprioception provides critical information in regards to specific joint position and motion required for coordinated joint motion. This afferent feedback is responsible for providing continuous information pertaining to the external and internal environments as needed to apply appropriate neuromuscular control strategies to regain and maintain joint stability.

1.2.2 Neuromuscular Control

Neuromuscular control is a general term that refers to the nervous system’s contribution to task performance.[46,47] Within sports medicine literature this term has been specifically used to define “the unconscious activation of dynamic restraints occurring in preparation for and in response to joint motion and loading for the purpose of maintaining and restoring joint stability.”[46] Recreational and sport activities impose a range of joint perturbations that range from very small to very large. Through proprioception the body can identify joint perturbations, but through neuromuscular control and motor recruitment an equalization of forces and moments can be achieved, thus restoring and maintaining functional joint stability. Adequate neuromuscular control must include proper timing, magnitude, and pattern in order to most efficiently counter both externally and internally produced joint forces and moments. The success of neuromuscular control at maintaining and restoring functional joint stability is dependent on proper feed-forward and feedback strategies.

1.2.3 Sensorimotor System and Non-Contact Anterior Cruciate Ligament Injury

The sensorimotor system (SMS) is responsible for the gathering of afferent information (proprioception), the processing and integration of this information with regard to previous experience, and the application of corrective actions (neuromuscular control) with the goal of maintaining and restoring functional knee joint stability.[49] When an athlete is performing dynamic and higher risk movements the successful execution of these described pathways is critical for injury prevention. Through adequate integration of afferent sensory input and preemptive and reactive motor output successful joint stability can be achieved.[49] However, if there is a problem, error, or inefficiency at any of these steps there is potential for the loss of functional knee joint stability, and thus the potential for injury. Based on non-contact ACL injury mechanisms seen in video analysis, it appears that these athletes are landing with improper landing strategies and may be using inefficient neuromuscular responses to counteract the landing demand.[29,30,40] Previous research has identified relationships between measurable characteristics of the sensorimotor system and landing characteristics important for knee loading.[50] Additionally, it is possible to enhance these systems and potentially lower the risk of sustaining non-contact knee injuries with specific training.[51-53] Therefore, it is imperative to consider the SMS and its implications on functional knee joint stability when investigating risk factors or developing intervention strategies for non-contact ACL injury.

1.3 EVALUATING RISK OF ANTERIOR CRUCIATE LIGAMENT INJURY

The identification of risk factors is important for the development of injury prevention interventions.[54] The most direct loading mechanism of the ACL is the application of an anteriorly directed shear force causing the proximal tibia to displace anteriorly relative to the femur.[41] Although anterior shear is the most direct loading mechanism Markolf et al.[41] also demonstrated that the addition of other forces, including tibial rotation and abduction moment, further increase forces translated to the ACL.[41] Over the years, potential risk factors including biomechanical, neuromuscular, and anatomical factors, have been associated to non-contact ACL injuries.[23,24,29,30,55-57] From an injury prevention perspective, biomechanical and neuromuscular factors are the most useful as these measures have the potential for modification through interventions such as physical training, which may help decrease forces at the ACL.[58-61]

1.3.1 Modifiable Characteristics Predictive of Anterior Cruciate Ligament Injury

There is an evidential hierarchy to injury risk factors. Characteristics that are able to significantly predict the occurrence of ACL injuries in prospective cohort studies serve as the highest evidence of a potentially causal relationship between internal characteristics and injury.[54] Although there have been studies that have identified multiple characteristics that are predictive of future ACL injury, very few have demonstrated this relationship with modifiable risk factors.[23,62] Landing biomechanics, specifically knee abduction angle and moment, have been able to predict future occurrence of ACL injury.[23]

1.3.2 Modifiable Risk Factors for Anterior Cruciate Ligament Injuries

Predictive risk factors are very important for the creation of preventative actions but it is also very important to consider other characteristics that have demonstrated a relationship with ACL injury. Descriptive studies have offered evidence of other relationships with ACL injury. Laboratory studies have demonstrated that certain landing mechanics increase tibial anterior shear forces at the time of peak ground reaction force.[31] This may be due to an increase in that axial load of the tibia and increased quadriceps pull on the proximal anterior tibia.[63-65] Gender comparative studies have found that females tend to land with greater peak knee abduction angle, smaller peak knee flexion angles and at initial contact, and time to peak knee flexion, suggesting that these characteristics may be risk factors for ACL injury.[31,66-69] These characteristics have face validity as risk factor for knee injury as they result in less than optimal knee loading and may be dissipating more force by using static structures rather than musculature surrounding the knee.[31,70-72] Research has also demonstrated that subjects with ACL deficient and reconstructed knees exhibit different landing mechanics such as lower knee flexion moment less abduction motion that likely results in lower anterior tibial shear force.[73-75] Biomechanical evaluation is a valuable resource for assessing risk of ACL injury, however many of these studies have utilized different tasks or have executed the same task in different manors resulting in different demands placed on the knee.

1.4 CURRENT ANTERIOR CRUCIATE LIGAMENT INJURY PREVENTION

To date there have been a number of studies that have investigated the use of a preventative training program aimed at the reduction of lower extremity injury incidence and a few of these have been aimed specifically toward preventing ACL injuries.[61,76-81] Within this literature there have been mixed results regarding the effectiveness of these training programs on decreasing ACL injury rates among athletes.

Anterior cruciate ligament prevention programs have utilized various training techniques including plyometric, balance, strength, and agility training.[23,78,80,82-84] Intervention training programs were completed either in pre-season or during season with a wide range of training session duration ranging from ten to forty-five minutes.[61,83] The majority of ACL injury intervention studies have been conducted using female adolescent to young adult athletes participating in sports such as soccer, volleyball, basketball, or handball[23,61,77,79,83,85,86] and follow-up durations have a wide range of three months to three years.[59,76] With such a wide array of interventions it is not surprising also to find a similar wide range of positive and negative results.

More recently the literature investigating ACL injury prevention has produced multiple meta-analyses to formally compare and contrast existing research, identifying characteristics associated with successful interventions.[58,60,87,88] Similar to many interventions there seems to be an exposure relationship demonstrated by studies with increased training and thus seem to be more effective at preventing injuries.[87] Regarding to specific training components within the interventions, meta-analyses has shown that plyometric and strength training are very beneficial whereas balance training may not be.[60,88] Additionally, technique coaching for biomechanical feedback has demonstrated positive effects on landing biomechanics related to ACL injury.[27,89] The success of plyometric training and landing coaching suggests that training of the sensorimotor system and biomechanical characteristics are important components required for ACL intervention protocols. However, there is little research to support and guide clinicians in choosing jump landing parameters that might target specific risk factors for ACL injury.

1.5 EVALUATION OF LANDING BIOMECHANICS FOR ACL INJURY

Anterior cruciate ligament injury prevention literature has demonstrated the usefulness and wide range of findings related to landing biomechanical characteristics. The three most commonly used tasks that are selected to evaluate landing biomechanics during a controlled athletic maneuver include drop-landings,[23,26,62,67,90] jump-landings,[27,42,57,66,91,92] cutting maneuvers,[63,93,94] or some combination of these. Each of these tasks is selected to elicit a specific demand to the subject that could be eliciting vertical ground reaction forces, posterior ground reaction forces, or change of direction forces. However, even between studies that have used very similar tasks aimed at eliciting similar demands there are still discrepancies in methodology that may change the actual demand placed on the knee joint during landing. Previous research has demonstrated that by changing a task parameter, such as jump distance, the demand experienced at the knee also changes as demonstrated by increased posterior ground reaction forces and tibial accelerations.[91] Additional research regarding the standardization of biomechanical tasks, such as jump distance, will provide evidence for task standardization leading to better comparisons to be made between research studies and aide in the selection of task parameters to impose more specific demands in future research.

1.6 DEFINITION OF THE PROBLEM

The measurement and evaluation of landing biomechanics is widely used in sports medicine for the purposes of describing mechanisms for injury, establishing risk factors, determining risk for injury, and evaluating interventions and outcomes of injury prevention programs and rehabilitation. The use of biomechanical evaluation of athletic maneuvers can be a critical piece in multiple stages of injury prevention. However, researchers employ a significant number and variety of landing tasks that make between study comparisons difficult and may limit the capability of biomechanical analyses to provide informative data to guide injury prevention and rehabilitation. A well accepted critical piece of performing biomechanical analysis is the standardization of the tasks to ensure similar demand between subjects. Current research has suggested that tasks completed with different demands (i.e. jump distance) do change biomechanical characteristics during landing.[91] How task demand affects specific landing biomechanics related to ACL injury remains unknown. This makes the selection of specific task parameters, such as jump distance, difficult to justify based on evidence.

1.7 PURPOSE

There is a need to investigate the relationship between jump distance and biomechanical ACL risk factors to determine specific demands that are more relevant and biomechanically sensitive for specific landing characteristics. To determine the most relevant and biomechanically sensitive task for ACL injury risk factors this dissertation will employ an investigative study with two purposes. The first purpose is to assess the effect of jump distance on biomechanical risk factors for ACL injury. The second purpose will be to examine how the relationship between sensorimotor system characteristics and landing biomechanics change throughout increasing jump distance.

1.8 SPECIFIC AIMS AND HYPOTHESES

Specific Aim 1: Determine if biomechanical risk factors and characteristics related to ACL injury change as jump distance increases from twenty to eighty percent of the subject's body height.

Hypothesis 1a: As jump distance increases the demand during landing will also increase as expressed by a significant increase in vertical and posterior ground reaction forces.

Hypothesis 1b: As jump distance and landing demand increase there will also be a significant increase in the expression of ACL risk factors (increase knee abduction angle peak and at initial contact, knee abduction moment, peak vertical and posterior ground reaction forces, and proximal anterior tibial shear force).

Hypothesis 1c: As jump distance and landing demand increases there will be significant changes in kinematic and kinetic measures related to knee joint loading (increased knee flexion peak and at initial contact, and proximal anterior tibial shear force).

Hypothesis 1d: As jump distance and landing demand increase muscle activation of the quadriceps and hamstrings pre-landing and post-landing activity will also increase.

Specific Aim 2: Determine if components of the sensorimotor system (proprioception, time to peak torque, and peak torque) can significantly predict the expression of biomechanical characteristics related to ACL injury and if this relationship changes with jump distance.

Hypothesis 2a: Threshold to detect passive motion, time to peak torque, and peak torque will each independently contribute to the variance seen in knee flexion and abduction angles at initial contact, peak knee flexion and abduction angles, peak knee flexion moment, peak abduction moment, and peak proximal anterior tibial shear force.

Hypothesis 2b: Threshold to detect passive motion, time to peak torque, and peak torque will together significantly contribute to the variance seen in knee flexion and abduction angles at initial contact, peak knee flexion and abduction angles, peak knee flexion moment, peak abduction moment, and peak proximal anterior tibial shear force.

1.9 STUDY SIGNIFICANCE

The overall aim of this study is to help determine how using different jump distances to standardize a forward jump and countermovement task, such as the stop-jump, will effect landing biomechanics related to ACL injury. Increasing jump distance during a stop-jump task has previously been shown to increase peak posterior ground reaction force, demonstrating an increase in landing demand.[91] However, current methodology for examining landing biomechanics lacks justification for how landing demand is standardized. Results from this dissertation will determine more relevant and biomechanically sensitive jump distance for specific risk factors for ACL injury. Future research will be able to use these results as evidence based justification for specific jump distance standardization. Clinicians such as athletic trainers or physical therapists will be able to determine appropriate landing demands during rehabilitation to isolate and train specific landing characteristics.

The second aim of this study is to determine how the relationship between the sensorimotor system and landing characteristics change as jump distance increases. The sensorimotor system and its characteristics are critical components of performance and injury prevention as they are responsible for the detection of joint perturbations and the execution of appropriate motor response.[53,95,96] Results from this aim will determine which components of the sensorimotor system predict specific landing characteristics related to ACL injury and if these relationships hold true as landing demand increases. Clinicians will be able to better determine appropriate training, rehabilitation, and intervention strategies that target specific risk factors related to ACL injury.

2.0 LITERATURE REVIEW

This review of pertinent literature will focus on non-contact ACL epidemiological, basic science, methodological, and intervention research as it applies to biomechanical landing characteristics and the prevention of non-contact ACL injuries. This section will first provide an overview of the incidence and impact of ACL injuries. Next, the mechanisms by which the ACL is injured are introduced and the contributions of the sensorimotor system to the cause and prevention of non-contact ACL injuries will be discussed. The biomechanical risk factors and interventions for non-contact ACL injury will be then reviewed with a focus on landing biomechanics. Lastly, important methodological considerations pertinent to this dissertation will be discussed.

2.1 EPIDEMIOLOGY OF ACL INJURIES

Anterior cruciate ligament injury is still one of the most common serious lower extremity injuries suffered in sports today.[34,97] It has been estimated that approximately 80,000 ACL injuries occur each year in the United States.[35] In 1982, the National Collegiate Athletic Association (NCAA) began tracking injury information in intercollegiate sports through the Injury Surveillance System (ISS). Over a period of sixteen academic years (1988-2004), 4,800 cases of confirmed ACL injuries, an average of 313 each year, were reported using this system.[97] This study sample is estimated to represent 15% of the total NCAA player population, equating to a possible 2,000

ACL injuries each year in the fifteen collegiate sports tracked by this system.[97] Additional proof of the magnitude of ACL injury in athletes is demonstrated by ACL injuries accounting for 31% of claims for an insurance agency specializing in youth recreation.[98] Such high injury rates have created a major financial burden estimated at two billion dollars in medical costs annually.[35]

Higher injury rates have raised interest in studying certain competitive sports over others. Sports such as soccer, basketball, gymnastics and handball have shown an increase in ACL injury rates compared to other sports, both in the NCAA and internationally.[34,36,97] Anterior cruciate ligament injury rates as high as 2.29 injuries per 1,000 match hours have been reported in Norwegian team handball for a single season.[99] A unique component to the ACL injuries seen in these sports is that the majority of them occur in a non-contact situation, providing possible avenues for prevention through training and modification of established risk factors.[14,36]

Just as certain sports demonstrate a high incidence of non-contact ACL injuries, sports like soccer, basketball, and handball have also shown gender differences in injury rates.[99-102] Four of the five sports with the highest incidences of ACL injuries are female sports.[97] Much of the literature has found that females are at a higher risk of non-contact ACL injuries than males.[14,34,38,97] This difference in gender injury rates has likely been a catalyst for an increase in research regarding injury mechanisms, risk factors, and potential injury prevention programs focused on females.

Despite a proliferation of ACL injury mechanistic and intervention research, injury rates have not been shown to diminish. According the NCAA ISS, ACL injury rates have held constant over sixteen years of injury surveillance.[97] In fact, from 1988 to 2004 there was an average 1.3% annual increase in ACL injuries. This seems to suggest some discrepancy between current preventative research and the actual application of injury prevention programs. Although this epidemiological study by Hootman et al.[97] was published almost a decade ago there is no recent update to these NCAA ACL injury statistics.

2.2 CONSEQUENCES OF ACL INJURY

Anterior cruciate ligament injuries are associated with great time-loss and high morbidity both in short-term and long-term consequences.[103-106] The traumatic injury to the ACL results in typical clinical signs and symptoms of traumatic joint disruption include pain and joint effusion, kinesiophobia, loss of mechanical knee joint stability, as defined by increased anterior joint laxity, loss of neuromuscular control and proprioception, and arguably the most significant acute consequences of this injury, loss of function.[107-114] In some cases, reconstruction of the ACL has been shown to immediately restore mechanical knee joint stability[115,116] but disruptions in the sensory motor system still linger as the patient progresses through rehabilitation.[117,118] Reider et al.[117] demonstrated that diminished proprioception in the ACL reconstructed knee lasted up to six months post-operatively.[117] Similarly, Nagai et al.[119] investigated the restoration of rotational proprioception after anatomical double-bundle reconstruction in patients twelve to fifteen months after surgery. Although most proprioceptive characteristics were restored, there were still deficits in ACL reconstructed limb.[119]

Injury and reconstruction of the ACL inherently involves a large amount of time loss due to surgical intervention and rehabilitation before returning to play. There is currently no published data on length of recovery and rehabilitation in adolescent or collegiate sports but data from the National Football League show that the rehabilitation process after ACL injury or surgical intervention lasts an average of 10.8 months.[106] However, that same study reported that only 63%

of athletes undergoing ACL reconstruction and rehabilitation ever return to the field for game play.[106] A similar study of professional soccer athletes showed that only 71% were still able to participate in competitive soccer after a follow-up of four years after surgery. If professional athletes return to sport participation at a rate of 29 – 37%, it is likely that adolescent athletes have a much less return to sport rate.

There is also a psychological impact on athletes experiencing this injury that may contribute to low return to play numbers.[120-122] Smith and Milliner[120] describe in a literature review how it is possible for athletes who are removed from play to experience depression and suicidal tendencies.[120] Although these studies do not directly look at ACL injury there has been a positive correlation established between severity of injury and emotional state of the injured athletes.[122] The longer rehabilitation process associated with this injury would suggest that ACL injuries might be a higher risk injury for athlete depression.

Injury to the ACL, despite surgical repair, has been shown to have long-term consequences even after successful return to sport. Multiple studies have demonstrated that regardless of surgical repair there is a very high incidence of early onset osteoarthritis (OA) following ACL injury.[7,18,20,21,123] Lohmander et al.[20] conducted a twelve-year follow up study examining the long-term effects of ACL injury on OA onset in female soccer players. 82% of women had radiographic changes in knee joint space index on weight-baring x-rays and 75% of women reported symptoms affecting their activities of daily living.[20] Additionally, multivariate analysis showed that having reconstructive surgery was not a significant predictor of the development of knee OA.[20] Kessler et al. also performed a cohort study examining the effects of ACL rupture on the development of knee OA and found conflicting results regarding specific effects of reconstruction. They demonstrated that subjects who underwent ACL reconstruction had better knee joint stability but developed higher incidence on knee OA (42% vs. 25%) at an eleven year follow up. However, Kessler et al restricted their definition to knee OA with grade two or higher, leading to lower incidence overall.

Understanding the short and long-term consequences of ACL injury is important for establishing a need for specific research related to this injury. ACL injury and subsequent reconstruction imposes multiple physical and mental, and short and long-term effects that will negatively impact the lives of individuals who suffer this injury. Advancements in prevention and rehabilitation of ACL injuries will decrease the impact on the health care system and increase the long-term health of athletes.

2.3 MECHANISMS OF NON-CONTACT ACL INJURIES

In the epidemiological processes of disease prevention the identification of specific mechanistic factors is a fundamental step that is the basis for effective prevention programs.[124] This same step is also important for the prevention of non-contact ACL injury because researchers and clinician must first understand how the injury takes place before focused strategies of prevention can be developed. Within ACL injury research studies have investigated both the mechanical cause of ACL strain and injury as well as the functional, or athletic, mechanisms for injury.

2.3.1 Mechanisms of ACL Strain

The ACL is one of the primary stabilizing ligaments of the knee.[125] It originates on the medial aspect of the lateral femoral condyle and inserts on the anterior aspect of the tibial plateau.[125] The orientation of the ACL places it in an ideal position to resist anterior tibial translation force. A few studies have investigated specific isolated motions of the knee that cause strain to the ACL.[126,127] Anterior tibial displacement relative to the femur, or an anterior tibial translational force, produces the highest and most direct loading of the ACL.[126,127] Although anterior tibial translation is the most direct loading mechanism, injuries to the ACL occur under various conditions and the addition of combined motion may place further strain on the ACL. Additionally, classic in-vitro ACL strain studies only measure strain at one specific knee position. In an effort to overcome these limitations of previous ACL strain research Markolf et al.[41] conducted another cadaveric study describing the biomechanical loading of the ACL under combined stresses throughout knee flexion. Again, isolated anterior tibial force was the most direct loading mechanism of the ACL, however, the greatest recorded force in the ACL was due to the combined loading of anterior tibial force and internal tibial rotation when the knee was near full extension.[41] In knee flexion, the greatest stress in the ACL occurred with the combined loading of anterior tibial force and abduction moment.[41] Research investigating the specific loading patterns of the ACL are important for determining what specifically applied forces produce strain in the ACL and thus increase risk of injury to the ligament. However, during sport participation forces are not applied to the knee in this manner.

2.3.2 Knee Kinematics of Non-Contact ACL Injury

The use of competition injury video footage has allowed researchers to evaluate movement patterns and landing mechanisms that may cause ACL injury. In 2004 the first study was published that used competition video for actual ACL injury to examine potential injury mechanisms.[128] Olsen et al.[128] analyzed twenty different ACL injuries that occurred in Norwegian handball players in the 1998 – 1999 season. A standardized reporting from was used for the coach examiners to evaluate each of the injury videos. Questions/categories ranged from type of activity, contact with another player, speed/intensity, push-off leg, and landing-leg. Physician examiners used another reporting from for the evaluation that asked questions regarding foot position, knee position at foot strike, relative time of injury during landing, movement direction, and weight distribution. The results of this study determined that ACL injuries in team handball most often occur during a plant-and-cut or a one-leg landing from a jump shot.[128] The specific injury mechanism during the cut or landing appeared to be a forceful knee abduction collapse from a position in which the knee is in near full extension with the addition of some sort of knee rotation.[128]

In a similar manner, Krosshaug et al.[40] examined ACL injury video to continue examining potential injury mechanisms but also to compare male and female injury mechanics. Twenty-seven of the thirty-nine cases occurred during a one-legged landing, two-legged, landing, or cutting maneuver (twelve male and fifteen female).[40] When analyzing the landing positions of these cases they found that females landed with higher knee and hip flexion than did male players at initial contact (15° vs. 9° and 27° vs. 19°, respectively, $p < 0.05$).[40] Knee abduction angle was not different at initial contact (4° vs. 3°, $p = 0.071$) but women did show more movement into abduction collapse with larger knee abduction angles at thirty-three milliseconds after initial contact (8° vs. 4°, $p = 0.018$).[40] This study also estimated the time point of rupture ranging from twenty-five to forty-six milliseconds after initial contact.[40]

Two additional studies compared ACL injury video during competition with video of a matched control performing similar movements but without injury. Boden et al.[29] used a total of twenty-nine ACL injury videos (eighteen women and eleven men) that included twelve videos in the sagittal view, six in the coronal-anterior view, and eleven in the coronal, posterior view. Twenty-seven control videos with athletes performing similar tasks and in similar camera angles were analyzed to compare with injury video data. They found no significant difference in knee flexion angle at initial contact or through the next five frames. Hip flexion data did show a significant difference between injured subjects and controls with the former using greater hip flexion during initial contact as well as the five frames after (52.4° vs. 33.4°, $p < 0.05$).[29] However, there was no gender difference in hip or knee flexion during landing.[29] In the coronal view there were no significant differences in knee abduction angle at initial contact between injured and controls or between genders. However, similar to Krosshaug et al.,[40] injured subjects moved into increased abduction angles during the frames after initial contact whereas controls remained in a similar abduction position as initial contact.[29] This same research group published a similar study that aimed to compared ACL injury video data between injured females to male injuries and female controls.[30] Similar results were found for knee flexion and abduction angles during landing with the exception of a significant difference in knee abduction angles at initial contact between injured females and injured males but not female controls.[30]

2.3.3 Summary

The identification of specific mechanisms that cause injury to the ACL is a critical piece of the injury prevention process. Researchers must first know and understand the mechanisms by which an injury occurs before considering potential modifiable risk factors that may be predictive of future injury. Video evidence from ACL injuries has determined common landing mechanisms that occur during injury which may be biomechanical risk factors for ACL injury.[29,30,40] The most common landing characteristic in ACL video analysis seemed to be the movement into greater knee abduction angle during landing, or valgus knee collapse. It is important to note that the characteristics discovered from injury video analyses are physical mechanisms of ACL injury. Researchers and clinicians must also consider the contributions of the sensorimotor system that may play a role in the use of such biomechanical characteristics.

2.4 SENSORIMOTOR SYSTEM AND NON-CONTACT ACL INJURY

2.4.1 Sensorimotor System Defined

The maintenance of joint homeostasis is crucial in the execution of successful joint and body motion. A hierarchy of systems is responsible for such successful motion through finely regulated feedforward and feedback control systems.[46] The feedforward control system describes the anticipatory actions that occur in preparation for an expected event.[129] The feedback system is characterized by the constant processing of afferent information and efferent control responses.[46] These two systems are housed within the sensorimotor system, a subcomponent of the more comprehensive motor control system.[46,47] Lephart and Fu[47] describe the sensorimotor system as the sensory, motor, and central integration and processing components involved in maintaining joint stability during body movements.[47]

2.4.2 Joint Stability

Joint stability is defined as the state of a joint remaining in or promptly returning to proper alignment through equalization of forces and moments.[46] It is common to refer to joint stability in regards to its subtypes of mechanical joint stability and functional joint stability. Mechanical, also known as clinical, joint stability refers to the integrity of passive joint components (ligaments, joint capsule, cartilage, and/or bony geometry) that act to limit excessive bony movement and excursion.[46] Clinically, this can be easily measured by assessing passive joint laxity using an arthrometer such as the KT-1000.[130] When the ACL is ruptured the knee's ability to resist anterior tibial translation is greatly diminished, and therefore results in an increased tibial translation when measured with the KT-1000.[108] This loss of mechanical joint stability due to a torn ACL can cause signs of instability including feelings of giving way, thus limiting the functional joint stability.[105,111,125] Use of the Lysholm knee score has demonstrated that individuals who have sustained a ligamentous rupture and have not had reconstructive surgery have diminished knee function scores and report higher symptoms compared to patients who have received reconstructive treatment.[103,105]

Functional joint stability is the complementary relationship between mechanical joint stability and the surrounding dynamic components (musculature) through appropriate feedforward and feedback controls.[46] After rehabilitation it is possible for individuals to restore functional joint stability despite not having the ACL reconstructed and mechanical joint stability completely restored. In these cases mechanical joint stability is still impaired but functional joint stability is still achieved through adapted feedforward and feedback neuromuscular strategies.[113,131-134] These individuals have been referred to as copers.[134] However, measurement of the dynamic components of functional joint stability is not as straight forward as ligament laxity testing. We are only able to quantify certain dynamic components, proprioception and neuromuscular control.[46,135]

2.4.3 Proprioception

Proprioception is the acquisition and processing of sensory information concerning the external and internal environmental condition of the body and can be separated into two categories.[48,49] The first category involves the use of information from the somatosensory system to adjust and modify motor patterns in response to the external environment, an uneven surface for example.[48,49] The second category involves the use of information from the somatosensory system in the planning and modification of internally generated movements through the determination of segmental motion and position.[48,49] The motor control system is under constant review and modification based on proprioceptive information.[49]

Proprioceptive information originates from peripheral mechanoreceptors located in both static and dynamic structures surrounding the joint.[136] When an injury such as ACL rupture occurs, the sensory information that was once provided by the intact ligament is no longer provided for the integration of motor control processes and therefore proprioceptive ability of the joint is diminished.[112,137-139] Studies have consistently shown that individuals that are ACL deficient have proprioceptive deficits as measured using threshold to detect passive motion and joint position sense.[138,140-143] A recent meta-analysis by Relph et al.[139] determined that ACL deficient knees have significant proprioceptive deficits in joint position sense compared to contralateral knees (std. mean difference = 0.52, $p < 0.001$), control subjects (std. mean difference = 0.35, $p = 0.001$), and reconstructed knees (std. mean difference = 0.52, $p < 0.001$).[139] Similar findings were found when comparing ACL deficient knees with control knees when proprioception was measured using threshold to detect passive motion (std. mean difference = 0.38, $p = 0.03$).[139] Conversely, Fonseca et al.[144] investigated proprioception using threshold to detect passive motion and position sense to determine that individuals who have good performance ratings based on the Cincinnati Knee Rating System do not have proprioceptive deficits compared to healthy controls.[144] This may suggest that individuals who are able to restore function despite being ACL deficient may also be able to restore proprioceptive ability. This study also highlights the potential trainability of proprioception.

Proprioception is important for the modification and correction of motor control during movement. This system is thought to be a critical piece of injury prevention because proprioception is an integral part of the sensorimotor system and is the first line of defense in corrective actions for motor control.[47,145] Research has established differences in proprioception between trained and untrained individuals.[146,147] To determine if extensive training has an effect on knee joint proprioception Lephart et al.[147] examined the knee joint proprioception of collegiate gymnasts and untrained healthy non-gymnasts using threshold to detect passive motion testing.[147] This study determined that trained gymnasts had increased proprioception compared to non-gymnasts.[112] These results highlight the potential trainability of proprioception, which may increase protective mechanisms for sport related injury. This leads to the question of whether diminished proprioception could also be a risk factor for injury.

Proprioceptive deficits likely contribute to the occurrence of sports injury and re-injury through a diminished reflex response, thus less ability for appropriate corrective responses in joint position.[53] In an effort to determine the relationship between proprioception, strength, and landing kinematics Nagai et al.[148] used regression analysis to quantify the relationship between an individual's threshold to detect passive motion, knee flexion and extension peak torque, and landing position during a single-leg stop-jump task.[148] This study determined that even when accounting for strength, proprioception was a significant contributor to knee flexion angle at initial contact ($r^2 = 0.274$, $p = 0.001$).[148] The researchers acknowledge that a limitation of this study was that only men were used and the results may not be generalizable to a female population. To the author's knowledge this is the only study investigating the relationship between proprioception and landing kinematics. Proprioception provides critical sensory information used to optimize motor control and is necessary for the maintenance of functional joint stability.[49,142] However, it is equally important to consider the efferent side of motor control when evaluation the sensorimotor system.

2.4.4 Neuromuscular Control

Neuromuscular control is a general term that refers to the nervous system's control of muscle activation and is the second half of the sensorimotor system. It is highly dependent on proprioceptive information.[46] The activation of musculature can be conscious or reflexive in nature and, although it is difficult to discern the difference during sport maneuvers such as landing, there is likely a combination of conscious and reflexive activation.[46] The measurement of muscle activation, via electromyography (EMG), during landing is a common practice in ACL injury research that allows investigators to describe the motor recruitment patterns and strategies used during different tasks and how they may vary between different population or demands.[31,63,94,131,133,149-153]

2.4.4.1 Neuromuscular control in ACL deficient patients

Kalund et al.[133] were among the first to use EMG analysis to describe neuromuscular differences between healthy and ACL-deficient (ACL-D) patients.[133] They found that during level walking there was no significant difference in muscle activation onset time between healthy and ACL-D subjects (differences = 0.019 – 0.0115 sec, $p \geq 0.05$).[133] However, when the demand of the task was changed to up hill walking (25° treadmill angle) ACL-D patients utilized significantly shorter hamstring onset times than the healthy controls did (differences = 0.066 - 0.111, $p<0.005$).[133] There was no significant difference in quadriceps onset time during any speed or treadmill incline.[133] These results were in line with previous research that describes the hamstring musculature as an agonist to the ACL in the restriction of proximal anterior tibial displacement.[154-156]

Later, Swanik et al.[150] examined the difference in reactive muscle activation between ACL-D, ACL-reconstructed (ACL-R), and healthy control subjects during four different functional tasks. The ACL-D group demonstrated increased hamstring activation during running compared to the ACL-R and control group.[150] Additionally, the ACL-D group showed significantly decreased quadriceps activation during landing when compared to the control group.[150] These group differences were not significant during the hopping and downhill walking tasks, demonstrating the specificity of neuromuscular strategies to each task and demand.[150] No significant neuromuscular difference was observed between the ACL-R and control groups. In a separate publication,[131] the same research group examined the anticipatory activation between the same groups. There were no significant differences in hamstring activation between groups during the 150ms time period prior to initial contact.[131] However, the ACL-D group did show side-to-side differences in quadriceps and gastrocnemius activation compared to the ACL-R and control groups.[131]

Together these studies show that ACL-D patients use different neuromuscular strategies to achieve joint stability due to the lack of mechanical joint stability that would be supplied by an intact ACL. These strategies also shed light on neuromuscular strategies that may decrease loading of the ACL in those with an intact ACL. Research has also investigated the difference in neuromuscular control strategies between genders due to the gender disparity in the incidence rate of ACL injuries.[34]

2.4.4.2 Neuromuscular control differences between males and females

Significantly different neuromuscular control strategies related to the ACL have been identified between genders. Palmieri-Smith et al[157] found that during a forward jump hop task females are more likely to have lower quadriceps-to-hamstring (Q:H) co-contraction indices as compared to males.[157] This suggests that females use a more quadriceps dominant landing strategy, which may increase loading of the ACL and lead to higher incidence of ACL injuries among females. This study also found that the medial Q:H index accounted for 0.792% of the observed variance in peak knee abduction moment in women, which is a predictive risk factor of non-contact ACL injury.[23,157] Fujii et al.[158] performed a similar study investigating the influence of hamstring muscle activity on tibial internal rotation during landing.[158] They found that increased lateral hamstring activation resulted in less internal tibial rotation, possibly decreasing loading and injury risk to the ACL.[158] This relationship was only significant in females.[158]

Sigward and Powers[159] investigated the gender differences in muscle activation in soccer athletes during a cutting task.[75] Although they were not able to find a difference in cutting kinematics, they were able to demonstrate that females utilized higher quadriceps activation than males (191% vs. 151% maximal voluntary isometric contraction (MVIC)).[159] The authors suggested that this use of increased quadriceps activation is part of an “at risk” pattern for ACL injury.[159] A similar study by Landry et al.[160] also used soccer players to investigate the gender difference in muscle activation during an unanticipated straight-run or cutting maneuver.[160] Again, differences in hamstring activation were found to show decreased hamstring activation in females during the cutting and straight run task.[160] This study also found that females used increased lateral gastrocnemius activation during both tasks.[160]

Sell et al.[33] also used an unanticipated change of direction maneuver to investigate gender difference in muscle activation during landing. They again found that females used significantly less hamstring activation and co-contraction values compared to males.[33] These differences among genders suggest that males and females may use different neuromuscular strategies during landing in order to maintain joint stability, which may be a potential factor in the greater injury incidence of ACL injuries seen in females.

2.4.4.3 Neuromuscular control and landing characteristics

Aside from group comparison research, as the previous sections have discussed, another important aspect of neuromuscular control is how it affects landing. Few studies have attempted to define how different neuromuscular control characteristics relate to different landing characteristics. In 2000 Colby et al.[63] sought to describe the kinematic and muscle activity characteristics that are used during four different sport-like maneuvers (sidestep cutting, cross-cut, stopping, and landing).[63] They found that subjects commonly used increased quadriceps activation compared to hamstring and landed with small knee flexion angles at initial contact.[63] The authors suggested that these kinematic and muscle activation characteristics may contribute to higher risk of sustaining and ACL injury and that changing these patterns may lead to a reduction in injury incidence.[63] Later, Sell et al.[33] determined that the direction and whether the task is reactive or not does not significantly affect hamstring or quadriceps activation.[33]

In 2009 Shultz et al.[64] performed a research study investigating the contributions of thigh muscle activation on landing biomechanics but found that quadriceps and hamstring activation were not a significant predictor of landing mechanics.[64] They did find that increased quadriceps activation was significantly related to peak proximal tibial shear force during landing when controlling for strength and joint excursions.[64] This finding is interesting when considering gender comparison studies that have identified increased quadriceps activation in females as a potential risk factor for non-ACL injury.

2.4.5 Summary

Successful joint stability can be achieved through adequate integration of afferent sensory input and preemptive and reactive motor output.[49] However, deficiencies due to injury or inefficiency will lead to loss of functional knee joint stability and thus increase the potential for injury. Based on non-contact ACL injury mechanisms seen in video analysis it appears that these athletes are landing with improper landing strategies and may be using insufficient neuromuscular responses to counteract the landing demand.[29,30,40] Previous research has identified relationships between measurable characteristics of the sensorimotor system and landing mechanics important for knee loading.[33,50,133,148,159-161] Additionally, it is possible to enhance these systems and potentially lower the risk of sustaining non-contact knee injuries with specific training.[51-53] Therefore, it is imperative to consider the SMS and its implications on functional knee joint stability when investigating risk factors or developing intervention strategies for non-contact ACL injury.

2.5 RISK FACTORS FOR NON-CONTACT ACL INJURY

The identification of risk factors is arguably the most important piece of injury prevention and is crucial in the development of effective intervention strategies. Various physical, biomechanical, neuromuscular, and environmental factors need to be analyzed for possible predisposing factors and causal factors of ACL injury. Biomechanical and neuromuscular factors are among the most useful measurable characteristics because they can be modified through training and rehabilitation.[28,52,78,162,163] However, few research studies have been published that assess the ability of biomechanical characteristics to predict the occurrence of future non-contact ACL injury.

2.5.1 Predictors of Non-Contact ACL Injury

Prospective cohort studies are most effective and reliable for determining causal injury risk factors.[164] Prospective studies are generally more reliable than retrospective because there is greater potential for control of data collection, as it does not rely on subject recall or previous record keeping.[165] These studies are used to compare the outcomes of one or more groups exposed to a risk factor to the outcomes of a control group.[166] The major advantages of this design in comparison to other observational designs are the ability to determine incidence of an injury and establish a temporal sequence between risk factor and outcome.[166] The establishment of temporal sequence is necessary to determine a causal relationship because in order for a factor to cause an injury, it must occur prior to the injury event.[54,167] The disadvantages seen with prospective cohort studies are the need for relatively large sample sizes and that the design can be time consuming.[54] Only a few prospective cohort studies have been used to assess the effect of injury risk factors on ACL injury and fewer have investigated modifiable characteristics.[23,168-170]

Hewett et al.[23] conducted a prospective study examining potential biomechanical risk factors for non-contact ACL injuries in competitive female athletes.[23] Two hundred and five female soccer, basketball, and volleyball players underwent biomechanical evaluation of a drop jump task that involved dropping off of a 31 cm box onto two force plates and immediately performing a maximal vertical jump.[23] During a thirteen-month surveillance period, nine ACL injuries were reported and confirmed (seven during soccer, and two during basketball).[23] Subjects who sustained ACL injury demonstrated significant increases in knee abduction angle and moment and increased ground reaction forces compared to the uninjured group.[23] The stance time was 16% shorter in the injured compared to the uninjured group, suggesting that the injured group experienced motion, forces, and moments more quickly.[23] It was also determined that an increase in abduction moment was a good predictor of ACL injury.[23]

Myer et al.[170] investigated the relationship of hamstring and quadriceps strength to ACL injury in female athletes using a matched case control study where strength was measured prior to ACL injury as part of a larger prospective study.[170] From 2002 to 2007, one hundred thirty-two competitive female and male soccer and basketball athletes were prospectively screened with quadriceps and hamstring isokinetic strength testing at 300°/s.[170] There were twenty-two subsequently confirmed female ACL injuries (sixteen during soccer and six during basketball).[170] All uninjured players were used as controls (eighty-eight female controls and thirty-two male controls).[170] Comparisons between the injured and control groups revealed a significant deficit in the hamstring strength in the injured group compared to male controls.[170] Quadriceps strength values in the injured group were not significantly different from that in the male or female control groups.[170] Although this study was not a prospective cohort study, these results still suggest that decreased isokinetic hamstring strength may place an athlete at an increased risk of sustaining an ACL injury.[170] The risk for injury may further increase if this characteristic is compounded with other neuromuscular characteristics such as decreased hamstring activation during landing.

Lastly, in 2010 Paterno et al.[168] published a prospective study investigation potential risk factors for the occurrence of a second ACL injury.[168] They found that a deficit in postural stability was able to significantly predict the occurrence of a second ACL injury with a sensitivity of 0.92 and specificity of 0.88.[168] Additionally, increased abduction moment impulse, total two-dimensional frontal plane knee excursion, and asymmetries in sagittal plane knee moments at initial contact where significant predictors of second ACL injury when controlling for one another.[168] Although this is focusing on characteristics predictive of re-injury it can still be relevant to first time injuries because this study does suggest that there may still be sensorimotor deficits on the injured side that have not resolves as demonstrated with decreased postural stability. However, there were no proprioceptive or neuromuscular measures included in this analysis.

Uhorchak et al.[169] conducted a prospective study to investigate risk factors for non-contact ACL injury. Eight hundred and fifty-nine cadets (739 males, 120 females) at the United States Military Academy underwent a physical (anthropometrics, joint laxity, and flexibility), radiographic (condylar width, notch width, tibial width, and eminence width), and strength (quadriceps and hamstrings) assessment prior to a four year observation period.[169] Of the twenty-four non-contact ACL tears (sixteen males, eight females) seen during this study period, multiple factors were demonstrated to be related to a significant increase in risk of ACL injury.[169] Potentially modifiable risk factors found in both sexes included higher than normal body mass index (BMI) and generalized joint laxity.[169] Although generalized joint laxity is not directly modifiable through training, it has been shown that training can increase proprioception,[171] which could possibly compensate for joint laxity via improved dynamic joint stability.[49] The presence of one or both risk factors greatly increased a persons relative risk for ACL injury.[169]

Overall, evidence from prospective research studies has demonstrated the importance of landing characteristics as risk factors for future non-contact ACL injury.[23,168] Most notably, frontal plane knee motion, as measured by peak or total knee abduction angle, and peak frontal knee loading, as measured by peak knee abduction moment, are common characteristic show to be risk factors among prospective analyses.[23,168] There is also evidence that decreased hamstring strength will also add to a person's risk of sustaining a non-contact ACL injury.

2.5.2 Other Potential Risk for Non-Contact ACL Injury

Predictive risk factors are very important for the creation of preventative actions but it is also very important to consider other characteristics that have demonstrated a relationship with ACL injury. Descriptive studies have offered evidence of relationships between landing characteristics and ACL injury. Laboratory studies have demonstrated that certain landing mechanics are related to increased anterior tibial shear force.[152] Sell et al.[152] performed a study investigating the predictors of proximal anterior tibial shear force (PATSF) during a stop-jump task and was able to define a significant relationship between PATSF and landing characteristics, such as peak posterior GRF, knee flexion moment, knee flexion angle, and vastus-lateralis activation at the time of peak posterior GRF (adjusted $R^2 = 0.8503$).[152] Gender also was a significant predictor when controlling for landing characteristics.[152] The greatest pairwise correlation was between peak PATSF and knee flexion moment at peak posterior GRF ($r = -0.8986$).[152] This negative coefficient is relative to the increasing knee flexion moment being an increasingly negative value due to the defined anatomical coordinate system joint rotations. This may be due to an increase in axial load of the tibia and increased quadriceps pull on the proximal anterior.[63-65] Previous research has determined that increased knee and hip motion reduce knee loading during landing, but increased quadriceps activation during landing is predictive of increased PATSF.[42,172]

Gender comparative studies have found that females tend to land with greater peak knee abduction angle, smaller peak knee flexion angles, knee flexion at initial contact, and at initial contact, and time to peak knee flexion, suggesting that these characteristics may be risk factors for ACL injury.[31,66-69] In a study by Lephart et al.[173] the authors describe a deficit between male and female landing strategies during landing.[173] Female subjects used significantly less knee and hip flexion displacement and less time to maximum angular displacement during a drop-landing task.[173] Mclean et al.[174] found that these same gender differences with the addition of peak knee abduction angle exist during a sidestep cutting maneuver.[174] In another gender comparison study Sell et al.[33] determined that during a reactive stop-jump task females again use less favorable kinematics during landing, including decreased peak knee flexion and increased peak knee abduction angles.[33] This study also was able to demonstrate that females exhibit increase PATSF and knee flexion moment with increased medial hamstring activation and co-contraction of the quadriceps and hamstrings.[33] The increased co-contraction value is contradictory to findings of previous studies[64,159] but the authors of this study and previous research suggest that this may be a compensatory mechanisms in response to the increased knee joint loading described.[33,175]

The landing characteristics discussed thus far in this section have face validity as risk factors for knee injury because they result in less than optimal knee loading and may be dissipating more force by using static structure rather than musculature surrounding the knee.[31,70-72] Norcross et al.[71] investigated landing characteristics in relation to quantified energy absorption during a drop-jump landing.[71] Subjects that utilized landing strategies that produced high initial energy absorption (first 100ms) also displayed significantly higher posterior GRF, PATSF, and knee extension moment.[71]

In addition to kinematic measures, a study by Sigward et al.[159] determined that females land with significantly less knee flexor moments, greater knee adductor moments, and greater quadriceps activation during a sidestep cut.[159] Also using a sidestep cut task, Mclean et al.[174] found that greater knee abduction moment was associated with increased hip internal rotation and increased knee abduction angle at initial contact and that this relationship was significantly greater in females.[174] Biomechanical evaluation is a valuable resource for assessing risk of ACL injury, however many of these studies have utilized different tasks or have executed the same task in different manners resulting in different demands placed on the knee.

2.5.3 Summary

As demonstrated, there have been many studies investigating landing characteristics that may be related to the risk for non-contact ACL injury. Many of these have shown positive relationships between landing characteristics and potential for injury. However, these studies have also utilized different biomechanical evaluation methods. Would these relationships and findings have been seen if using a different task or different parameter? These differences are most notably seen in the discrepancies between the tasks that are being evaluated. Throughout the literature there seem to be too many types of landings, including static landings and counter-movement landings. Static landings are those where the subject is asked to jump to or drop onto a landing area and remain still once they have landed. Counter-movement landings include some sort jump or change of direction upon landing. This type of landing includes tasks such as stop-jumps, drop-jumps, and cutting maneuvers. It is well accepted that counter-movement tasks are more sport-like but with each difference in the task or demand of that task my challenge the subject in a different way that limits the generalizability of findings. Recent findings from the University of Pittsburgh's Warrior Human Performance Research laboratory have demonstrated that different landing tasks elicit different landing characteristics. Even with-in the same type of task, discrepancies may still exist as the task demand changes with other parameters such as jump distance. These discrepancies between tasks are not just seen in risk factors studies. Prevention research has also utilized different biomechanical tasks for the evaluation of potential intervention strategies aimed to reduce the risk of ACL injury.[26-28,32]

2.6 INTERVENTION STRATEGIES AND LANDING BIOMECHANICS

The evaluation of landing biomechanics also has been an important measure in the development of training strategies to establish the effectiveness of potential intervention strategies on landing biomechanics. In 1996 Hewett et al.[26] performed a small-scale intervention study (n=11) to examine the effects of a jump-training program on lower extremity landing biomechanics and strength.[26] This study utilized a vertical jump-landing task to simulate a volleyball block and evaluated landing biomechanics relative to sport.[26] They found that subjects were able to increase their jump height by 10% and decrease their landing impact by an average of 0.8 times body weight ($p<0.01$) and knee abduction/adduction moment by 1.3 to 2.1 %bodyweight times height ($P<0.05$).[26] Lephart et al. performed a similar study with a higher sample size of twenty-seven to investigate neuromuscular and biomechanical changes due to a plyometric training program versus a basic resistance program.[51] This study also evaluated landing biomechanics using a landing from a maximal vertical jump task.[51] After an eight-week training period the researchers found that there was a significant overall decrease in peak knee flexion moment and peak hip flexion moment but no significant group interaction.[51] Plyometric training significantly lowered the reactivity time for the medial hamstrings during landing.[51] Although these two studies utilized similar methodology with a very similar population (high school female athletes with a mean age of approximately fifteen years) some of the landing characteristics were quite different including vertical GRF (4.2 vs. 2.4 times body weight) and knee abduction moment (0.027 vs. 0.035 Nm / bodyweight x height).[26,51] This discrepancy may be due to the simulated volleyball block that was incorporated into the Hewett et al.[26] protocol which may have altered the relative demand of the task, thus leading to a different biomechanical response.[26,51]

In two intervention studies by Myer et al.[176,177] researchers used the identical methodology as a previous risk factor analysis.[23] They used a drop-vertical jump (DVJ) tasks that began with the subject dropping off a thirty-one cm platform and immediately performing a maximal vertical jump upon landing.[176,177] Myer et al.[177] reported similar knee abduction moment values for the pre-test measures as Hewett et al. (0.44 vs. 0.54 Nm/kg*m, respectively).[23,177] A different research study used this same task to evaluate another training program but used a twenty-five cm box.[178] Although there was only a six cm difference in the drop heigh and the procedure was otherwise identical Pfile et al.[178] reported a peak abduction moment of 0.12 Nm/kg*m, 4.5 times lower than what was previously reported by Myer et al.[177]

Despite the positive effects on landing characteristics, each of these research studies have demonstrated there is a concern for clinicians or other researchers try to develop efficient and effective means of evaluating athletes for these risk factors. Although each study provides important and relevant findings, there are distinct differences in the type of task and the specific demand of these tasks which may affect the landing characteristics of each subject between different studies and especially between research groups. Different landing tasks will elicit different biomechanical responses. However, this comparison also suggests that even though the same task execution is used, different demand of the same task may also change the biomechanical response.

2.7 METHODOLOGICAL CONSIDERATIONS

2.7.1 Threshold to Detect Passive Motion

There are two common methods for assessing proprioception, they are: 1) threshold to detect passive motion and direction and 2) active joint position sense. Both of these techniques have been used in relation to knee injuries, however, TTDPM has been demonstrated to have higher reliability values compared to active joint position sense.[148,179,180] The test for TTDPM also has been described to target slow-adapting mechanoreceptors due to the slow speeds[53,135] utilized. The test is thought to rely less on ligamentous or articular structures being placed in a state of tension when a starting position of 45° of knee flexion is used.[112] This method has also been shown to be related to knee kinematics during a stop-jump task similar to the aims of this dissertation.[148] For these reasons TTDPM will be used to assess proprioception in this dissertation.

A previous study by Lephart et al.[147] determined that TTDPM testing with the knee positioned at 45° is able to distinguish between those with exceptional proprioception (gymnasts) and healthy controls.[147] The reason this specific study is relevant is because this dissertation will use healthy individuals and thus will need a measure that is sensitive enough to measure proprioception in a healthy population.

This methodology enables targeting of the slow adapting mechanoreceptors such as the Ruffini endings and Golgi like organs.[46,53] Because of this, it is important to ensure that the speed the dynamometer will be moving is slow enough to specifically target these receptors. Previous studies using this methodology have used speeds ranging from 0.25° to 0.5° per second.[112,147,148] The reliability of TTDPM at this speed and a starting position of 45° of knee flexion has yielded good to excellent reliability (ICC [3,k] = 0.879 – 0.917, SEM = 0.194° – 0.216°).[148] This dissertation will use a testing speed of 0.25 degrees per second and a starting position of 45° of knee flexion.

Lastly, there are additional systems that contribute to the sensorimotor system, including the visual system, that will also provide information regarding limb motion that is not originating from the targeted joint mechanoreceptors. Additionally, the motion of the proprioception jig will also induce changes in pressure and contact on the skin (cutaneous stimulus) as the system passively moves that joint. The Biodex system also produces a clicking sound as the dynamometer begins its motion that would provide cueing to the initiation of the test. Therefore, during TTDPM testing it is critical to eliminate the input of the visual, cutaneous, or auditory cues that may confound the testing results.[145] To control for this each subject will be blind folded with earplugs and headphones producing white noise to eliminate any visual or auditory cuing to the motion of the dynamometer or knee joint. A pneumatic sleeve will be placed over the lower leg and will be attached to the proprioception jig to minimize the cutaneous feedback from the proprioception device. Each subject will be required to indicate which direction the knee joint was moving upon detection of motion to minimize guessing and to further assess the proprioceptive ability.

2.7.2 Knee Isokinetic Strength

Isokinetic dynamometry is a common method used to assess muscular performance and time to peak torque (TTPT) as components of the sensorimotor motor system.[153,175] Isokinetic strength testing will hold the speed at which the joint is moving at a constant velocity while the subject performs reciprocal maximal effort motions toward knee flexion and extension. Pilot testing at the Neuromuscular Research Laboratory has established that isokinetic hamstring strength testing has very good to excellent intersession reliability for TTPT torque (ICC (1,2) = 0.99; SEM = 7.5msec) during five reciprocal extension and flexion motions at 240°/s. within a set ROM limit of 0° to 60°.[181]

Deficits in hamstring strength have been shown to be a predictor of future ACL injury in female athletes.[170] Previous studies also have demonstrated differences in hamstring activation between ACL-D and healthy subjects as well as between males and females during landing and change of direction tasks.[33,160] Therefore, this dissertation will investigate the contribution of hamstring strength and force generating capabilities as measured by peak torque production. This dissertation will also assess the contribution of hamstring TTPT as a measure of neuromuscular control.

Previous pilot testing was able to determine that the mean knee velocity in the sagittal and frontal plane occur at much faster speeds than the commonly assessed speed of 60°/s. Knee motion occurs at an approximate speed of 240°/s. Additionally, a prospective risk factor analysis by Myer et al.[170] determined that isokinetic strength measures at 300°/s. were predictive of future ACL injury. Therefore, this dissertation will use an isokinetic testing speed of 240°/s.

2.7.3 Hamstring and Quadriceps Surface Electromyography

This dissertation will measure the muscle activation of the quadriceps and hamstrings prior to and during landing to quantify the neuromuscular control patterns at the knee prior to and during landing. Surface EMG (sEMG) is a commonly used method for assessing neuromuscular control or the efferent side of the sensorimotor system.[46,135] The measures of preactivity and reactivity in the quadriceps and hamstrings represent the feedforward and feedback control systems of the neuromuscular control system.[135] Previous studies from our laboratory have demonstrated that sEMG measures of the quadriceps and hamstrings are reliable (ICC (2,1) = 0.664 – 0.989, SEM = 9.448 – 29.324 %MVIC).[182]

Previous research has demonstrated that the actions of the quadriceps muscle group applies an anterior translational force on the proximal tibia that causes strain in the ACL.[154-156] By measuring the strain in the ACL, Renstrom et al.[154] determined that significant increases in strain occur due to quadriceps force from zero to forty-five degrees of knee flexion. The same study also determined that simulated hamstring force was protective to the ACL by applying a posterior translational force, but this was only significant in knee flexion angles greater than thirty degrees.[154] Because of these described implication to ACL strain, this dissertation will investigate the muscle activation of both the quadriceps and hamstring groups. Additionally, previous research that has identified muscle activation differences between healthy and ACL-D and between genders has specifically investigated the vastus-medialis, vastus-lateralis, and the medial (semimembranosus and semitendinosus) and lateral (biceps femoris) hamstring groups.[33,118,149,160,183] These muscles are superficial which will allow muscle activity to be measured using sEMG techniques. The center of the semimembranosus and semitendinosus muscles of the medial hamstring group are very close to one another and therefore run the risk of cross-talk when measured using sEMG. For this reason it is suggested to measure them as one group and refer to the data collected as the medial hamstring group. Therefore, this dissertation will also examine the muscle activation of these four specific locations using sEMG.

Signal processing technique is a critical piece of EMG processing. Previous research has suggested that sampling frequencies of at least 1000 Hz is adequate for sEMG data collection to avoid the risk of aliasing due to a low sampling frequency.[184] However, more recent guidelines from the International Society of Electromyography and Kinesiology (ISEK) state that a minimum sampling frequency of 800 Hz is adequate using a band pass filter of 10-400 Hz.[185] The ISEK also states higher sampling frequencies are recommended for increased resolution and accuracy.[185] Additionally, a recent study at the Neuromuscular Research Laboratory utilized a sEMG sampling frequency of 1500 Hz and found medial hamstring preactivity was a significant predictor of total knee abduction displacement.[181] A 1500 Hz sampling frequency will be used in this dissertation.

Another component of sEMG signal processing is normalization of the signal. It is common practice and recommended by the ISEK to normalize the filtered sEMG signal to the sEMG signal collected during a maximal voluntary isometric contraction (MVIC).[33,118,153,159,185] This technique converts the units of measure from volts to %MVIC. Not only does this procedure of normalization make the interpretation of the results more intuitive but it also allows for the comparison of results between subjects. This dissertation will normalize all sEMG signals to the average muscle activation during a five-second MVIC trial.

It is also important to consider the temporal parameters and specific calculations by which preactivity and reactivity will be defined, as this will effect the measures interpretation and relation to the sensorimotor system and neuromuscular control. Preactivity is the muscle activation present just prior to initial contact and represents the feedforward activation responsible for increased muscle stiffness.[46,135] A 150-millisecond window will be used to measure muscle preactivity as this timeframe is specific to the feedforward activity in preparation for landing.[46,135] A 150-millisecond window will also be used to measure reactivity. This timeframe is specific to the feedback (reflexive) muscle activation due to initial joint loading during landing.[46,135] During these specific timeframes the integrated sEMG signal (iEMG) will be calculated to describe the muscle activation throughout the window. The iEMG is reported in a value of %MVIC*Seconds and is a commonly used to describe muscle activation levels with respect to time.[51,150,152,153,186,187]

2.7.4 Two-Leg Stop-Jump Biomechanics

Three-dimensional (3D) kinematic measurements using video based marker trajectory is a commonly used method for assessing lower extremity landing kinematics and kinetics throughout sports medicine literature.[33,42,73,148,152,153,177,188] The Plug-in-Gait[189] biomechanical model used in this dissertation is based on the Helen Hayes ridged segment model developed by Kadaba et al.[190,191] Sixteen 0.014m reflective markers are placed bilaterally throughout anatomical landmarks on the lower extremity (anterior superior iliac spine, posterior superior iliac spine lateral thigh, lateral femoral condyle, lateral shank, lateral malleolus, heel, and head of the second metatarsal). Windolf et al.[192] has determined that the Vicon optical tracking system has an accuracy of 63 ± 5μm and overall precision of 15μm. Camera based motion capture with a similar system and biomechanical model has also been shown to be a reliable measure of lower extremity joint angles (ICC(3,1) = 0.595 to 0.922 and coefficient of multiple correlation (CMC) = 0.650 to 0.982) and joint moments angles (ICC(3,1) = 0.592 to 0.870 and CMC = 0.711 to 0.957).[193]

Many different tasks have been used to evaluate landing biomechanics related to ACL injury but counter movement tasks are considered to be more sport-like and thus more relevant to athletic populations. Data from the UPitt Warrior Human Performance Research Center has demonstrated that two-leg landing tasks elicit greater knee abduction angles and moments compared to single-leg tasks. In a prospective ACL injury risk factor study, Hewett et al.[23] used a two-leg landing task followed by a maximal vertical jump.[23] This dissertation will measure 3D kinematics and kinetics during a two-leg stop-jump task as the task has been previously used by a number of studies to evaluate landing biomechanics related to ACL injury.[31,42,152,194]

In an attempt to standardize the effort and jump height a previous dissertation by Clark[181] used a VERTEC Jump Trainer (Sports Imports, Columbus, OH) to provide a target for the subject to reach for during the counter movement jump after the initial landing.[181] It is important to ensure that each subject is performing the second jump at a comparable effort and this may influence the initial landing. However, this does not control for the height of the initial jump, which likely will influence the initial landing to a greater extent. Therefore, this dissertation will use a 30.5cm hurdle and specific verbal cueing to attempt to control the jump height of the initial jump, as well as a VERTEC to ensure consistent countermovement jump demand.

3.0 METHODOLOGY

This dissertation is a cross-sectional comparative study further evaluating the validity of current biomechanical risk factors for non-contact ACL injury and assessing the relationship between components of the sensorimotor system and biomechanical risk factors for ACL injury.

3.1 DEPENDENT AND INDEPENDENT VARIABLES

3.1.1 Specific Aim 1: Effect of Jump Distance

The dependent variables used for this specific aim will be biomechanical landing characteristics that have been associated with ACL injury risk (listed below).

- Peak vertical ground reaction force (% body weight)
- Peak posterior ground reaction force (% body weight)
- Peak knee flexion angle (degrees)
- Knee flexion angle at initial contact (degrees)
- Peak knee abduction angle (degrees)
- Knee abduction angle at initial contact (degrees)
- Peak knee abduction moment (Newton*meters / bodyweight (kg) * height (m))

- Peak proximal anterior tibial shear force (% bodyweight)

Additional variables that were used to investigate the effect of increasing jump distance are:

- Pre-landing muscle activation of the vastus lateralis, vastus medialis, lateral hamstring, and medial hamstrings
- Post-landing muscle activation of the vastus lateralis, vastus medialis, lateral hamstring, and medial hamstrings

The independent variable for this specific aim will be jump distance. The jump distances that will be used will be twenty, forty, sixty, and eighty percent of each subject's height from the edge of the force platform.

3.1.2 Specific Aim 2: Effect of Jump Distance on the Relationship Between Sensorimotor System and Biomechanical Risk Factors for ACL Injury

The sensorimotor system is a critical aspect of injury prevention and a target for interventions aimed at preventing ACL injuries. Therefore, each biomechanical risk factor will have a separate regression equation to determine the association between measured aspects of the sensorimotor system and each biomechanical risk factor. The biomechanical variables that will be included in this specific aim are all those listed above in Specific Aim 1. The independent variables that will be used in Specific Aim 2 as predictors of landing biomechanics will be components of the sensorimotor system. These variables include:

- Threshold to detect passive motion and direction toward knee extension (degrees)
- Threshold to detect passive motion and direction toward knee flexion (degrees)
- Knee extension time to peak torque (milliseconds)

- Knee flexion time to peak torque (milliseconds)
- Knee extension peak torque (Nm/kg)
- Knee flexion peak torque (Nm/kg)

3.2 SUBJECT RECRUITMENT

Ethical approval of this dissertation will be acquired from the University of Pittsburgh's Institutional Review Board (IRB). Potential subjects will be recruited from the University of Pittsburgh population and surrounding institutions and recreational sport organizations with the use of recruitment fliers. To ensure homogeneity of participants all potential subjects will be screened over the phone for inclusion and exclusion criteria as listed in the next section.

3.3 SUBJECT CHARACTERISTICS

Females have been at the forefront of non-contact ACL injury research due to the increased incidence of this injury in the female athlete population.[34] Additionally, previous research has demonstrated a gender effect on both landing biomechanics and neuromuscular control strategies.[66,67,94,195,196] Therefore, only females will be recruited for this dissertation. Healthy females that are physically active at least three days per week for a minimum of thirty minutes a session will be recruited to participate in this study. Potential participants will be recruited using flyers posted throughout fitness and activity centers of the University of Pittsburgh.

3.3.1 Inclusion Criteria

Potential subjects will be included in this study if they meet the following criteria:

- Female
- Aged 18 to 35
- Physically active at least three days per week for at least 30 minutes each session

3.3.2 Exclusion Criteria

Potential subjects will be excluded from participation in this study if any of the following criteria are true:

- Known history of ACL injury or reconstruction
- Known history of surgical procedure involving the lower extremity
- Previous lower extremity musculoskeletal injury in the last six weeks that required medical attention (i.e. doctor's appointment, physical therapy, emergency room, or urgent care) and has been at least three weeks since the last sign or symptom of initial injury
- Knowingly pregnant
- Known history of any other musculoskeletal or neurological condition that may affect muscle function, peripheral sensory input, or the ability to perform the tasks involved in this study

3.3.3 Sample Size Calculation

An *a priori* power analysis was conducted using G*Power 3[197,198] statistical software to estimate the minimum amount of subjects needed to achieve the desired power. The current studies uses two specific aims with different statistical analyses, thus, two separate power analyses were performed each based on the statistical procedure needed for the two specific aims. Table 1 describes the statistical test parameters that were used in the sample size estimation. To the author's best knowledge the effect of jump distance on biomechanical characteristics has not been previously investigated using a repeated measures ANOVA. Previous studies have demonstrated that increasing jump distance does significantly increase ground reaction forces and tibial accelerations with Pearson correlation coefficients ranging from 0.357 – 0.933. Therefore, to estimate the needed sample size for the first specific aim a conservative effects size of 0.20 was used yielding an estimated sample size of fifty subjects. Previous research also has investigated the association between strength and proprioception with landing biomechanics and demonstrated R^2 values of 0.304 ($F = 10.259$, $p = 0.001$) with a sample size of fifty subjects.[50] Therefore, a effect size of 0.30 was chosen for the regression analysis of this study to ensure adequate power for the second specific aim. Based on the estimated sample size from each analysis the number of subjects needed for the current study is fifty-three.

Table 1. Sample Size Calculation

Statistical Test	Power	Alpha	Effect Size	# of Measures/ Predictors	Corr. Among Measures	Non-sphericity Corr.	Estimated Sample Size
Repeated Measures One-way ANOVA	0.80	0.05	0.20	4	0.30	1	50
Multiple Linear Regression	0.80	0.05	0.30	6	N/A	N/A	53

3.4 INSTRUMENTATION

3.4.1 Three-Dimensional Motion Analysis System

Lower extremity kinematics and kinetics will be measured using 3D coordinate data and ground reaction force data collected simultaneously. Three-dimensional trajectory data will be collected using a high-speed infrared optical capture system (Vicon Motion Systems Inc, Centennial, CO). This optical capture system is composed of eight wall mounted and two tripod mounted MX+13 high-speed infrared cameras (Vicon Motion Systems Inc, Centennial, CO.). These cameras track infrared light that is reflected off the 14mm retro-reflective markers that will be placed on the subject's lower extremities according to the Plug-in-Gait marker set.[189] Marker trajectory data collected by the camera system will be transferred to Vicon Nexus Software (Vicon Motion Systems Inc, Centennial, CO) at a sampling frequency of 250 Hz. The ten cameras will be oriented in a way that gives the system the best marker visibility throughout the task within a 4m long x 2m wide x 2.5m high capture volume. The camera system will be calibrated using the five-marker wand technique recommended by the manufacturer's guidelines. The Vicon motion analysis system has a reported accuracy of 117µm.[192] The orientation of the global coordinate system (**Figure 1**) will be defined by the position of the five-marker wand so that the origin is on the corner of the force place, positive x is toward the anterior direction of the subject, positive y is toward the left side of the subject, and positive z is directed upward.

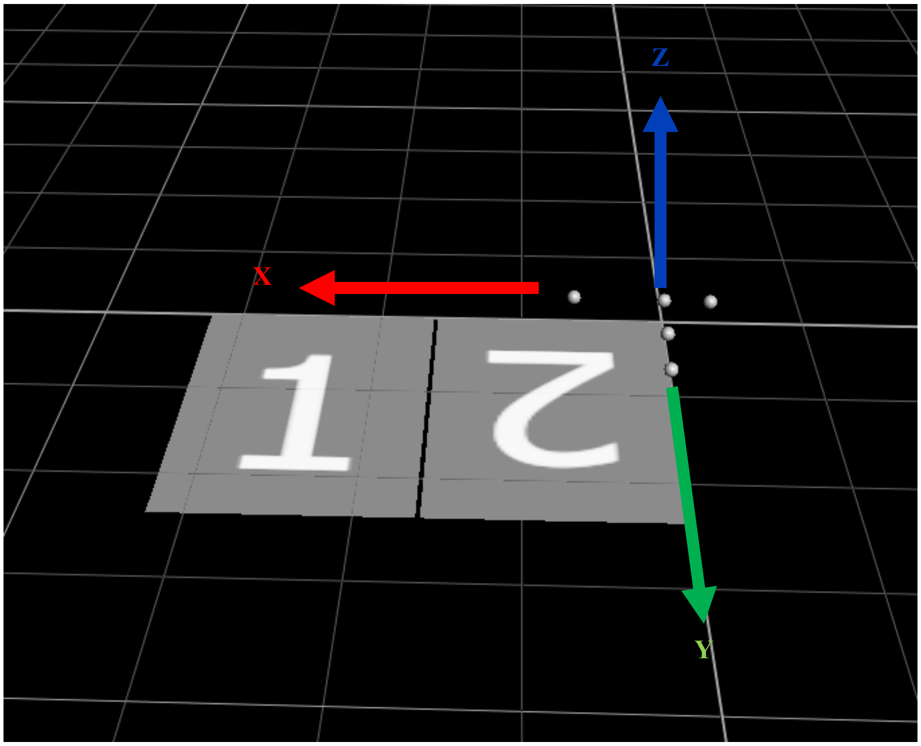


Figure 1. Global Coordinate System and Force Plate Orientation

3.4.2 Force Platform System

Ground reaction force data and joint kinetic calculations will use data collected by a 60cm x 40cm force platform (Type 9286BA, Kistler Instrument Corp., Amherst, NY) at a sampling frequency of 1500 Hz.[199] This specific force platform uses four piezoelectric three-component force sensors mounted at each corner of the platform and a built-in charge amplifier. Custom build flooring has been constructed around the force plates to allow for a flush floor surface. The force plate and motion-analysis data will be time synchronized using Vicon Nexus software (Vicon Motion Systems, Inc, Centennial, CO).[200] The orientation of the force plates will be measured and manually entered into the Nexus software so that center of pressure measures will be relative to the global coordinate system for kinetic calculations (**Figure 1**).

3.4.3 Surface Electromyography System

Muscle activity of the quadriceps and hamstrings will be measured using the Noraxon direct transmission system (DTS) multi-channel telemetric surface electromyography (sEMG) system (Noraxon U.S.A. Inc., Scottsdale, AZ)[201] along with Ambu® Blue Sensor N rectangular (30mm X 22mm X 1.6mm) pre-gelled Ag/AgCl, active, bipolar, self-adhesive surface electrodes (Ambu®, Denmark).[202] This is a 16-bit resolution system with an input range of ± 3.5mV and is composed of self-contained sEMG sensor transmitter units, a belt receiver unit, and a Noraxon 2400R G2 analog output desktop receiver unit. Surface EMG data will be sampled at 1500Hz. Each sEMG signal will be passed through a single-end 500-gain amplifier and a 10-500Hz low-pass filter within a self- contained Noraxon TeleMyo DTS sensor unit. The DTS unit also applies a notch filter at 50-60Hz to dampen noise related to internal electrical components. The sensor units transmit signals to the belt receiver unit, which then transmit signals to the desktop receiver unit. Raw sEMG signals will then be passed from the desktop receiver to a 32-channel 24-bit analog-to-digital board (Vicon Motion Systems LTD, Centennial, CO) to convert the analog signal to digital form. The converted sEMG signal will then be time synchronized and recorded using Nexus software (Vicon Motion Systems LTD, Centennial, CO).

3.4.4 Isokinetic Dynamometer

The Biodex System III isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY) will be used to measure both threshold to detect passive motion (TTDPM) and direction, peak torque

(PT), and time to peak torque (TTPT) for both hamstrings and quadriceps, and provide an isometric resistance to measure muscle activation during maximal voluntary isometric contractions of the hamstrings and quadriceps. This isokinetic dynamometer is a popular tool for the measurement of TTDPM[117,143,148,203] and direction, and TTPT.[148,204,205] Calibration of the Biodex System III dynamometer will be performed as outlined in the manufacturer's service manual.

3.4.5 Vertical Jump Target

A vertical jump target will be used to standardize the effort of the jump required after the initial landing. A VERTEC Jump Trainer (Sports Imports, Columbus, OH) will be used as a physical target positioned at 90% of the subject's measured maximal vertical jump. The VERTEC Jump Trainer has been used in previous research to standardize jump height during counter movement tasks.[181]

3.5 TESTING PROCEDURES

All testing will take place at the University of Pittsburgh's Neuromuscular Research Laboratory. Each subject will report to the laboratory for one testing session lasting approximately one hour and thirty minutes. Subjects will be asked to refrain from engaging in exercise or additional physical activity other than their daily living activities for the twenty-four hours prior to the testing session. Upon arrival to the laboratory, inclusion and exclusion criteria will again be confirmed by reviewing the subject-specific phone screen. Once inclusion and exclusion criteria are confirmed the investigator will discuss the study's aims and procedures and each subject will be given the opportunity to ask questions or voice any concerns that they may have. After all questions are answered the subject will sign an informed consent document as required by the IRB.

Before the beginning of laboratory testing each subject will be required to fill out the Tegner activity score questionnaire (Appendix A) designed by Tegner and Lysholm[206] to rate a person's level of participated activity. This questionnaire has been validated to compliment the symptomatic based scoring by quantifying the amount of recreational and occupational activity that post ACL-reconstruction individuals participate in.[206]

A specific testing order will be used for each subject with the intention to minimize the effect of higher- level tasks on proprioception testing and the effect of repeated trials on peripheral muscle fatigue. A specific testing order as outlined below will be used:

- Threshold to detect passive motion and direction
- Dynamic warm-up
- Biomechanical assessment of landing characteristics
- Knee flexion and extension strength and time to peak torque

For the purposes of this dissertation, only the dominant knee of each subject will be tested and analyzed. The dominant leg will be operationally determined as the preferred kicking leg when kicking a soccer ball. All testing will be completed in compression shorts and shirt and the subject's personal athletic shoes, except for threshold to detect passive motion and direction for which shoes were removed.

3.5.1 Threshold to Detect Passive Motion and Direction

Prior to TTDPM testing the Biodex System III will be calibrated as specified by manufacturer's guidelines. Specific subject positioning will be used to minimize variability between subjects and to ensure minimal tactile cueing during the test. The subject will be seated upright in the Biodex chair and mechanical adjustments to the chair will be made to standardize patient positioning. The fore-aft position of the seat back will be adjusted so that the popliteal fossa of the test limb is approximately four centimeters from the edge of the chair to minimize tactile sensation from contact with the chair while the knee joint is rotating. The chair and dynamometer position will be adjusted to align the femoral condyle with the axis of rotation of the dynamometer to ensure consistent joint rotation of the knee. Two shoulder straps and a waist belt will be tightened to keep the subject in the same position throughout testing but still comfortable to the subject.

It is also important to minimize any additional input or potential cueing from the visual, auditory, or tactile senses. Therefore, the subject will be fitted with a blindfold, foam earplugs, and over-the-ear headphones that produce white noise during testing to eliminate cueing from these senses and potential confounding of the measurement. To reduce any additional tactile cueing from the moving device a cotton tube sock will be placed over the subject's lower leg and foot before being placed in a pneumatic sleeve that will be inflated to 40 mmHg during testing. Range of motion limits of the device will be set by extending dynamometer attachment jig and the subject's knee to approximately zero degrees of knee flexion pressing the "Set Away" button on the dynamometer. Similarly, the knee will be moved toward flexion until just before the subject's leg comes in contact with the chair, about ninety degrees, and then pressing the "Set Towards" button.

During testing the subject’s leg will be positioned to forty-five degrees of knee flexion as measured with a goniometer. The “get-position” button will be selected within the Biodex Researcher’s Toolkit® software to get the dynamometer measured angle and will be written down as the “zero” reference angle and typed into the “go-to” reference that will be used to instruct the device to return this position after each test. Isometric mode will be selected in the software and the speed values will be zeroed. The researcher will manually select the randomized direction of rotation from a pre-allocated sequence. The subject will be instructed to “press the remote button once you are able to sense motion at your knee and you can distinguish the direction of rotation. Once the white noise begins the device will begin to move sometime between zero and one minute.” To initiate the testing, white noise will be turned on and the researcher will increase the dynamometer speed to 0.25 °/sec in the appropriate direction. Once the subject presses the remote button the dynamometer will stop moving and the end position will be recorded by pressing the “get-position” button. The test limb will then be returned to the starting position by selecting the “go-to” button. Each subject will be allowed one practice trial prior to measured tests. Each direction will have three successful trials collected and the order of tests will be randomized and balanced using a Latin Square technique.[207]

Figure 2: Threshold to Detect Passive Motion and Direction

3.5.2 Dynamic Warm-up

Before completing the maximal vertical jump and jump-landing tasks, each subject will complete a dynamic warm-up in order to maximize jump performance.[208,209] The content of the dynamic warm-up includes ten walking lunges, ten reverse lunges, ten single-leg Romanian dead lifts, 10 straight leg kicks with each leg, and high knees for a distance of ten meters.

3.5.3 Maximum Vertical Jump Height

Just after the dynamic warm-up and prior to the stop-jump testing participants will have their maximum vertical jump height tested. The subject will line up under the vanes and while keeping both feet flat on the ground they will reach up as high as possible and push the vanes away. This measured height will be the subject's standing reach distance. Next the subject will again begin standing directly under the vanes. Without taking a step the subject will perform a counter-movement jump and try to touch and push the highest vane possible. This measure will be their raw maximum vertical jump height. This value will be normalized to the subject's standing reach height by subtracting the value of the standing reach height from the raw maximum vertical jump height. Jump height measurements will be performed three times with the highest jump height used as the maximum vertical jump height.

3.5.4 Biomechanical Assessment of Landing Characteristics

3.5.4.1 Subject Preparation

Preparation for biomechanical assessment of landing will begin with anthropometric measurements including weight, height, leg length (standing distance from anterior superior iliac spine to medial malleolus), knee width, and ankle width. These measurements will be entered in the Vicon Nexus software to create a custom model from the 3D coordinate data.[190] This data will also be used during inverse dynamic calculations to determine knee joint forces and moments.

After all anthropometric measurements have been taken the quadriceps and hamstring area will be prepped for EMG electrode placement on the subject's dominant limb. Placement and skin preparation of the medial and lateral quadriceps and medial and lateral hamstring electrodes will be in accordance with SENIAM guidelines for EMG placement.[210] While the subject is positioned in a long sitting position the electrodes for vastus-lateralis (VL) will be placed at one-third of the line between the anterior superior iliac spine (ASIS) and the patella. The electrodes over the vastus-medialis (VM) will be placed at the 80% position on the line between the ASIS and the medial joint line (i.e. if the distance is measured to be 50cm than the electrodes will be placed at 40cm from the ASIS). The electrodes on the hamstring musculature will be identified with the subject lying prone and their knee positioned in approximately forty-five degrees on flexion. The location for the biceps-femoris (BF) electrode will be identified 50% of the distance following a line between the ischial tuberosity and the lateral epicondyle of the tibia with the thigh slightly externally rotated. The medial hamstrings group (MH) location will be identified as 50% of the line between the ischial tuberosity and the medial epicondyle of the tibia while the thigh is slightly internally rotated. Once the electrode sites have been identified the skin will be prepped by shaving any hair, lightly abrading, and wiping the placement site down with alcohol. The electrodes will then be place parallel to the line of action of the muscle group as shown in **Figure 3**. The remote sensor units will be connected to each of the electrodes and submaximal contractions (MVIC) of each muscle group will be used to confirm placement of electrodes. Confirmation of each electrode placement will be done by visual inspection of the measured signal in the Nexus software while the subject performs as 50% effort muscle contraction. Each remote sensory unit and electrodes will be secured to the skin using double-sided tape, Transpore® tape (3M, St. Paul, MN), and pre-wrap to minimize sensor movement artifact in the EMG signal.

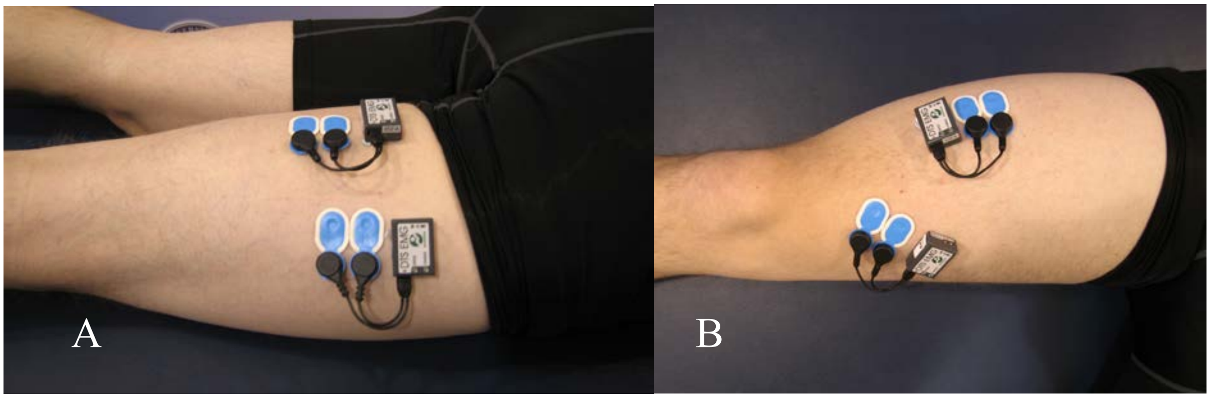


Figure 3: Hamstrings (A) and Quadriceps (B) EMG Electrode Placement

Maximal voluntary isometric contractions will be collected using the Biodex system III. The subject will be placed in a seated position with the for-aft chair adjustment set to allow the popliteal fossa three centimeters of space from the front edge of the seat. The rotational axis of the knee will be aligned with the dynamometer axis of rotation and the knee flexion extension attachment pad will be place on the distal lower leg, just above the malleoli. The subject will be secured to the chair using one waist strap and two shoulder straps. Verbal instruction will be given to each subject that include “on ‘GO’ we will ask that you extend your knee by kicking out as hard as you can and maintain it until I say relax.” During the contraction, five seconds of EMG data will be recorded. A hamstring MVIC will be conducted in the same position except the subject will be asked to “flex their knee by pulling your heel back as hard as you can until I say relax.”

Once EMG sensors have been attached and MVIC trials have been collected sixteen 14mm retroreflective markers will be placed on the subjects lower extremities according to the lower extremity Plug-in Gait biomechanical model (**Figure 4**).[189] Markers will be placed bilaterally on the following anatomical locations: ASIS, posterior superior iliac spine (PSIS), lateral thigh, lateral femoral epicondyle, lateral shank, lateral malleolus, posterior aspect of the heel, and second metatarsal head.

Figure 4. Lower Extremity Plug-in-Gait Marker Placement

3.5.4.2 Stop-Jump Task

The camera system will be calibrated using the manufacturers recommended guidelines and the global coordinate system will be defined prior to the testing session with the subject. Once subject-preparation has concluded a calibration trial will be collected with the subject standing on the force plate in the anatomic neutral position with their arms abducted to ninety degrees and palms facing forward. Verbal instruction will be given to “Point your toes forward, place your feet directly under your hips, keep your knee and hips as straight as possible, and hold still.” More specific segment position instructions will be given if needed upon visual inspection. A three second calibration trial will be collected while the subject remains still in this position. This trial will then be labeled and processed in the Nexus software to establish segmental coordinate systems specific to the subject’s biomechanical model.

Subjects will begin by standing at a distance of 20%, 40%, 60%, or 80% of their height from the edge of the force platform. The jump distance order will be randomized for each subject prior to the testing session using a random number generator and Latin square.[207] A six-inch hurdle will be placed half the distance from the starting line to the force plate to standardize the jump height to the force plate. To standardize the jump after the initial landing the VERTEC Jump Trainer will be positioned with the target vanes directly above the center of the force platform with the bottom vane equal to the subject's maximal vertical jump height. Each subject will be given verbal instruction and a visual demonstration of the task. Subjects will be asked to "begin with both toes on the line, after a count down of 'three-two-one-jump' perform and double-leg broad jump forward over the hurdle landing with one foot on each force platform. Immediately after landing, perform a maximal vertical jump to touch the bottom vane on the VERTEC Jump Trainer." One practice trial at each jump distance will be required so that each subject has a minimum of five practice trials. Trials will be excluded and repeated if the subject does not take off with both legs, each foot does not completely land on the force plate, does not clear the hurdle, or misses the target vane during the vertical jump. Kinematic, force platform, and EMG data will be visually inspected after each trial and repeated if insufficient data (such as EMG signal drop-out) was collected.

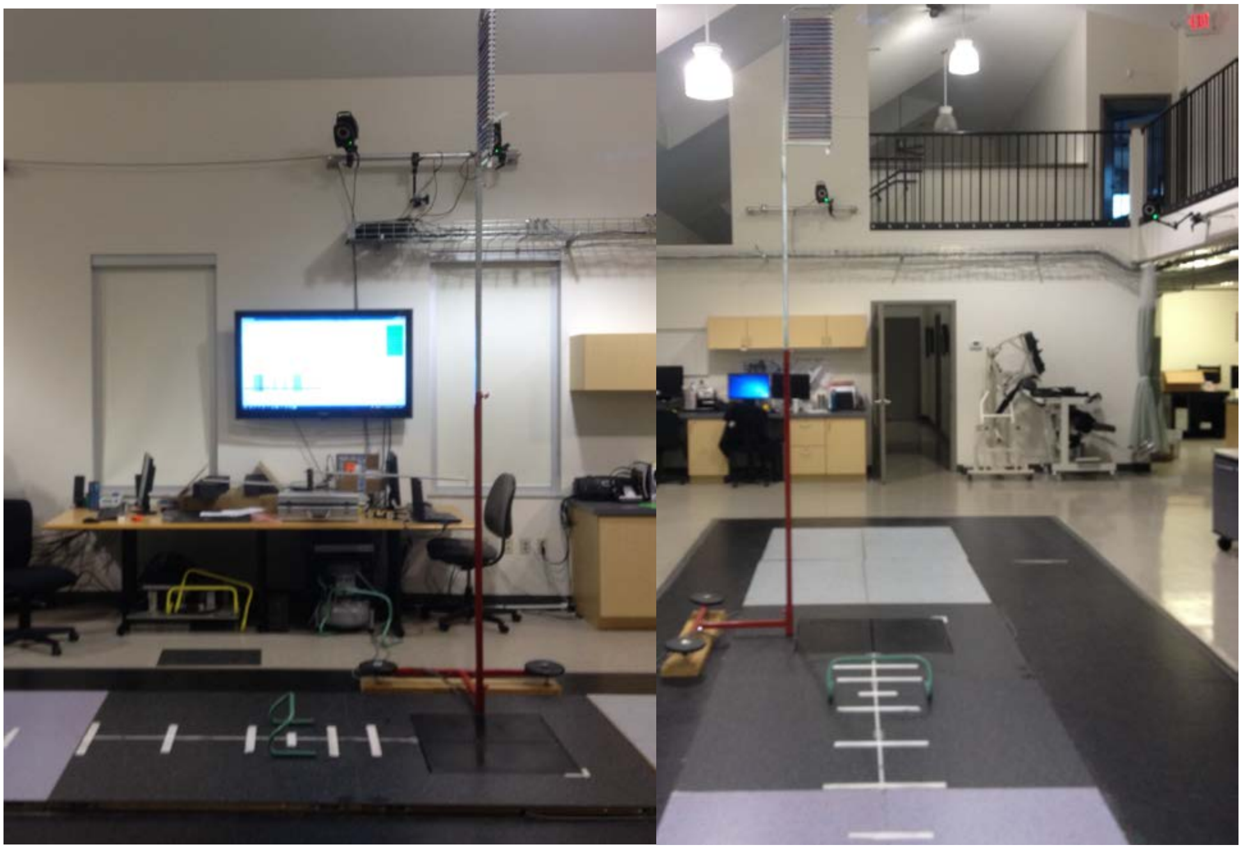


Figure 5: Capture Volume Setup

3.5.5 Knee Flexion and Extension Isokinetic Strength and Time to Peak Torque

Knee flexion and extension strength and time to peak torque of the dominant knee will be measured simultaneously on the Biodex system III isokinetic dynamometer. Subjects again will be seated in the Biodex chair in the same position as MVIC trials. Isokinetic strength testing will be completed at 240°/s for both flexion and extension between zero and sixty degrees of knee flexion. Our laboratory has demonstrated that isokinetic strength testing of knee extension and flexion between zero and sixty degrees is a reliable measure (ICC (2,1) = 0.99, SEM = 7.5ms).[181]

Subjects will be positioned in the Biodex chair as described during the MVIC testing. The range of motion limits on the Biodex dynamometer will then be set to zero and sixty degrees of knee flexion. The tester will visually set the subject's knee in zero degrees of flexion for the "away" limit and confirm the joint position with a goniometer. This procedure will be done again at sixty degrees of knee flexion to set the "towards" limit. The tester will then place the subject's knee at forty-five degrees of knee flexion and pause the dynamometer to record the limb weight. After verification of the subject's body position and dynamometer settings the subject will be given two sets of three-repetition practice/warm-up trials, one at 50% effort and one at 100% effort. The tester will instruct the subject to begin with their knee bent as far back as possible (ninety degrees of knee flexion) and begin reciprocal contractions of "pushing out and pulling as hard and as fast as you can" after a countdown of "3 – 2 – 1 – GO." The practice / warm-up trials will include one set of three repetitions at 50% of the subjects' perceived maximal effort followed by one set of three repetitions at 100% of the subject perceived maximal effort. After a one-minute resting period the subject will be asked to perform one set of five maximal repetitions, instructed as "as hard and as fast as you can" which will be used as measured trials.

3.6 DATA REDUCTION

3.6.1 Threshold to Detect Passive Motion and Direction

Threshold to detect passive motion and direction is a measure of joint excursion. During testing the starting position is constant for each trial at forty-five degrees of knee flexion. The subject will press a trigger that stops the dynamometer and the final joint angle is recorded. The measure of TTDPM is the difference between the ending joint angle and the beginning joint angle (as described in equation below). This measure will be calculated and averaged together for three successful trials toward knee extension and three trials toward flexion.

$$\text{TTDPM} = \theta_{\text{end}} - \theta_{\text{Begin}}$$

3.6.2 Landing Kinematics, Kinetics, and Muscle Activation

All marker trajectory, ground reaction force, and EMG data will be recorded using the Vicon Nexus software. Raw marker trajectory data will be filtered using a cross-validation Woltring filtering routine (quantic spline filter) within the Nexus software.[211] Ground reaction force data will be filtered using a low-pass, zero-lag fourth-order Butterworth filter with cutoff frequency of in Matlab (R2014a, Mathworks Inc., Natick, MA).[47, 128] The Plug-In-Gait model is a rigid segment biomechanical model, based on the Helen Hayes biomechanical model, and will be used to calculate joint angles, forces, and moments within the Nexus software. The estimation of hip, knee, and ankle joint centers and the definition of segmental coordinate systems will use subject-specific anthropometric data according to the procedures described by Kadaba et al.[190] and Davis et al.[212] Three-dimensional joint angle data will be calculated with Euler angle rotational decomposition using the right-hand rule in a sequence of X, Y, Z. Joint forces and moments will be calculated using inverse dynamics as described by Davis et al.[212]

Further Data reduction will be completed using a custom written Matlab (R2014a, Mathworks Inc., Natick, MA) script. Initial contact will be defined as the time point where the vertical ground reaction force exceeds a threshold of 5% of the subject’s body weight. Knee flexion and abduction angles will be identified at this time point. Peak knee flexion, knee abduction, and hip flexion will be defined at the maximal joint excursion in the specified direction during the landing phase. Landing phase will be defined as the time between initial contact and the change in knee joint power from negative to positive. Peak knee flexion and abduction moment will be defined as the maximal vector moment in the respective plane of motion throughout landing phase. Peak PATSF will be defined as the maximal knee joint force in the X (anterior) direction during the landing phase, normalized to the subject's body weight, and reported as N/kg. Peak vertical and posterior GRF during landing will be normalized to the subject's body weight and reported as % body weight.

Raw EMG data will be synchronized and recorded with the Nexus software, exported to an ASCII file, and processed using a custom written Matlab script. A fourth-order low-pass Butterworth filter with a cut-off frequency of 12 Hz will be used to filter the raw EMG signal. The mean EMG signal (V) from middle four seconds of the respective MVIC trial (Quadriceps or Hamstrings MVIC) will be used to normalize the EMG signal throughout the stop-jump trials and reported as %MVIC. The integrated EMG (iEMG) signal of each channel during the 150ms leading up to initial contact and the 150ms following initial contact will be expressed as the area under the curve (%MVIC x s).

3.6.3 Knee Flexion and Extension Strength and Time to Peak Torque

Knee extension strength will be defined as the average peak torque normalized to body weight (%body weight) during the five reciprocal trials toward knee extension. Knee flexion strength will be calculated the same way using the normalized average peak torque toward knee flexion. Time to peak torque will be defined as the time from the initiation of motion in the respective direction to the recorded peak torque for each repetition. Knee extension and flexion time to peak torque will be the average time to peak torque toward knee extension or flexion, respectively.

3.7 STATISTICAL ANALYSIS

All statistical analyses were performed in STATA 13 (Statacorp LP, College Station, TX). Descriptive statistics were calculated for all variables. Plots were generated and each variable was examined for outliers. The first specific aim of this dissertation was to examine the effect of jump distance on biomechanical landing characteristics. Repeated measures ANOVA analysis was used to examine the with-in subject differences between jump distance. First, normality was assessed for all kinematic, kinetic, GRF, and muscle activation. All normally distributed variables were also tested to determine if the assumption of sphericity was met. If sphericity it is not met the Greenhouse-Geisser adjusted ANOVA results were be used. For all variables that are not normally distributed, the Freidman's ANOVA was used to test for within-in subject difference by jump distance. Post-hoc pairwise analyses used the Bonferroni p-value adjustment.

The second specific aim of the dissertation was to examine the effect of jump distance on the relationship between biomechanical landing characteristics and the sensorimotor system. Separate sets of multiple linear regression equations by jump distance were created for each of the previously mentioned biomechanical landing characteristics. Bivariate analyses included a correlation matrix of the independent variables and potential collinearity was assessed. The full model was fit and a backward stepping technique was be used to remove non-significant predictors from each regression equation. Residual analyses were completed to examine linearity, heteroscedasticity, outliers, and potential leverage points. Statistical significance for both specific aims was set to 0.05. This procedure was repeated for each jump distance, creating a separate regression equation for each jump distance and each dependent variable.

4.0 RESULTS

The purpose of this study was to assess the effect of jump distance on biomechanical risk factors for ACL injury and muscle activation. Landing biomechanics related to ACL injury were assessed during a double-leg stop-jump maneuver completed at four jump distances (20%, 40%, 60%, and 80% of the subject's body height) using a video-based motion analysis system. Repeated measures analysis of variance (ANOVA) was used to test the within subject differences of landing biomechanical characteristics and muscle activation among different jump distances. Secondly, this study aimed to evaluate the relationship between sensorimotor characteristics at the knee and biomechanical risk factors for ACL injury. Univariate analyses of the independent and dependent variables are presented first. Bivariate statistics are then presented to assess the pairwise relationship between each independent and dependent variable. Multiple linear regression models were created and tested for each dependent variable by jump distance.

4.1 SUBJECTS

Fifty-six female subjects volunteered to participate in this study, however three did not meet study eligibility criteria due to previous history of ACL injury. This left a total of fifty-three subjects who were consented and participated in the study. Two subjects were not able to successfully complete the stop-jump task at a distance of 80% of their body height and were excluded from the repeated measures ANOVA of Hypothesis 1. However, data for these subjects were used for multiple linear regression models at the 20%, 40%, and 60% distances. Subjects came from a wide variety of activity levels and sport backgrounds. For inclusion, subjects met a minimum activity level of exercising at least three days per week for at least thirty minutes each session, however a questionnaire was still used to quantify activity related to the knee. The Tegner activity score questionnaire (**Appendix A**) was used to rate each subject's level of activity relative to the knee.[206]

Table 2. Subject Demographic Summary

n = 53	Mean	SD	Min	Max
Age (years)	23.2	4.3	18	31
Height (cm)	166.6	7.5	152.6	183
Weight (kg)	64.97	9.49	48.7	97.8
Activity level (0-10)	6.2	1.6	3	9

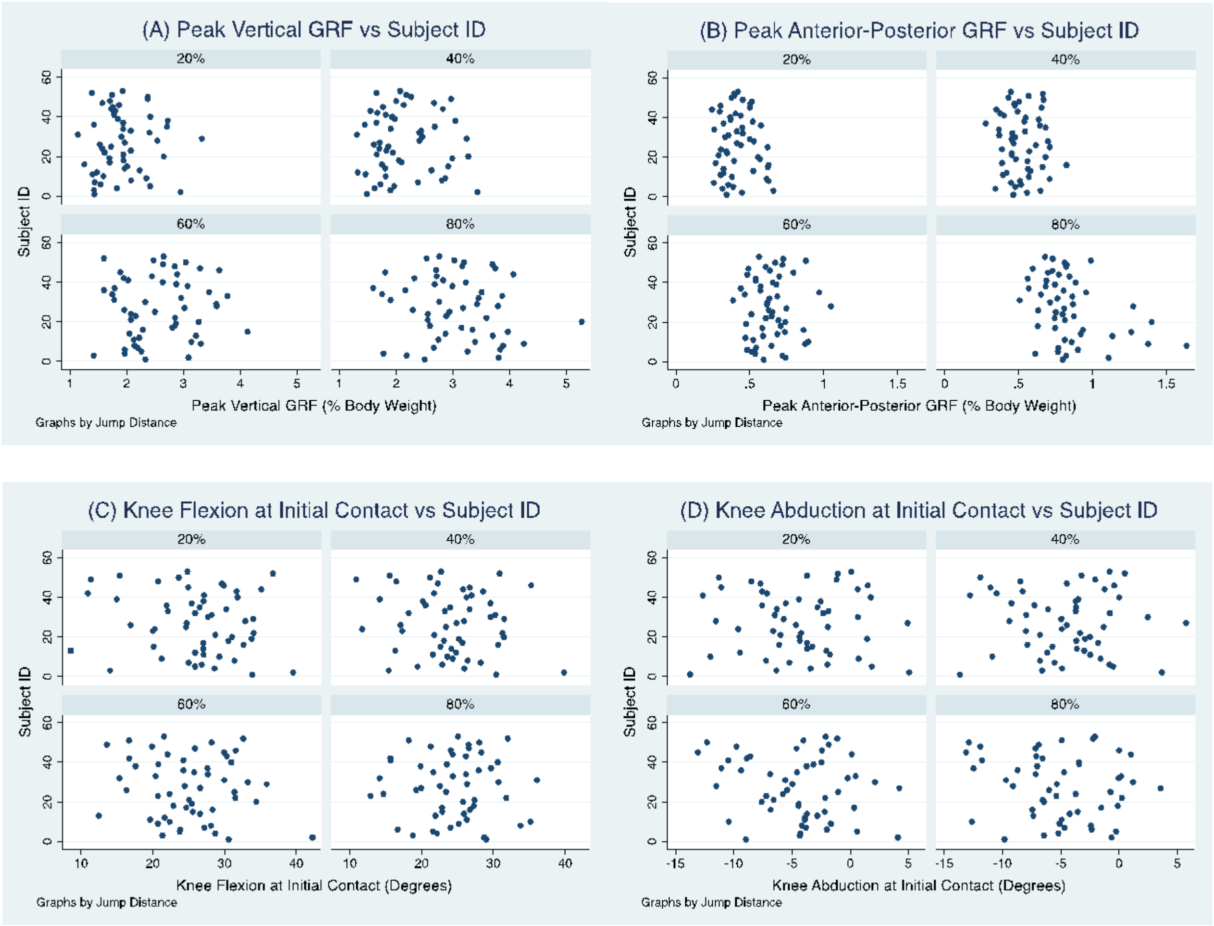
Abbreviations: number of subjects (n), standard deviation (SD), Minimum (Min), Maximum (Max), centimeters (cm), kilograms (kg)

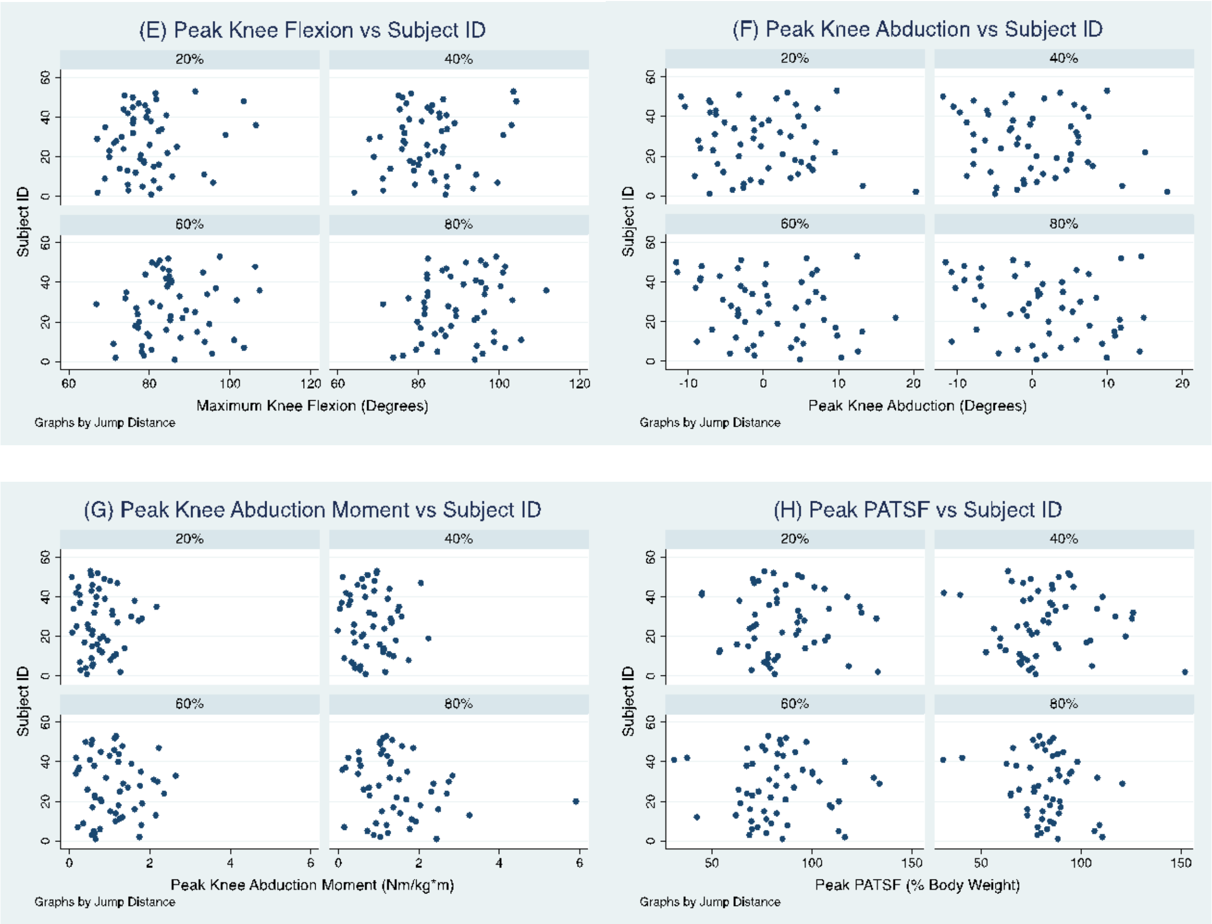
4.2 WITHIN SUBJECT DIFFERENCES IN LANDING BIOMECHANICS AND MUSCLE ACITIVTY BETWEEN JUMP DISTANCES

The first specific aim of this study was to assess if each jump distance would elicit different responses in biomechanical risk factors for ACL injury and different muscle activation levels of the quadriceps and hamstrings. Repeated measures ANOVA were used to test for differences in landing kinematics, kinetics, and muscle activation.

4.2.1 Potential Outliers

Figure 6 includes scatter plots of each biomechanical variable against subject ID for each jump distance. Based on visual inspection of the data there were some potential outliers that warranted investigation. Peak vertical ground reaction force (**Figure 6A**) displayed a high point at a jump distance of 80% for subject 21. Based on further data review within the motion analysis software there was no evidence that this data point was incorrect. Also associated with this increase in vertical ground reaction force is an increase in knee abduction moment at a jump distance of 80% for the same subject (**Figure 6H**). No other extreme points were identified that warranted further review.





(E) Peak Knee Flexion vs Subject ID

(F) Peak Knee Abduction vs Subject ID

(G) Peak Knee Abduction Moment vs Subject ID

(H) Peak PATSF vs Subject ID

Figure 6. Scatter Plots of Biomechanical Variables vs Subject ID by Jump Distance

4.2.2 Normality Test Results

Normality testing was completed to determine appropriate testing methods. Shapiro-Wilk test results (**Table 3**) determined that peak vertical ground reaction force, peak anterior-posterior ground reaction force, peak knee flexion, peak knee abduction moment, peak PATSF and all muscle activation measures were not normally distributed for at least one jump distance. However, evaluation of histograms for each of the variables showed that only peak knee abduction moment had obvious deviation from normality. Because ANOVA has been demonstrated to be a robust test for deviations from normality,[213] standard repeated measures

ANOVA was used to analyze the within subject differences between jump distances except for peak knee abduction moment and muscle activation measures. The only muscle activation measure that did not have any obvious deviations from normality based on histogram plots was reactivity of the vastus lateralis, therefore, repeated measure ANOVA tests were used.

Table 3. Dependent Variable Shapiro-Wilk Normality Test Results

	20% p-value	40% p-value	60% p-value	80% p-value
Biomechanics				
Peak vGRF (%BW)	0.0177*	0.0045*	0.2815	0.4894
Peak apGRF (%BW)	0.0230*	0.3577	0.0834	0.0000*
Knee Flexion at IC (degrees)	0.0950	0.7951	0.9463	0.7976
Knee Abduction at IC (degrees)	0.8330	0.8853	0.5388	0.2644
Peak Knee Flexion (degrees)	0.0002*	0.0362*	0.1369	0.4975
Peak Knee ABD (degrees)	0.0849	0.3514	0.3899	0.1555
Peak Knee ABDmom (Nm/kg*m)	0.0111*	0.1954	0.0400*	0.0000*
Peak PATSF (%BW)	0.1777	0.0541	0.0832	0.0056*
EMG				
Preactivity VL (%MVIC)	<0.0001*	<0.0001*	<0.0001*	<0.0001*
Preactivity VM (%MVIC)	<0.0001*	<0.0001*	<0.0001*	<0.0001*
Preactivity MH (%MVIC)	0.0034*	0.0001*	0.0034*	0.0018*
Preactivity LH (%MVIC)	0.0010*	<0.0001*	0.0006*	0.0013*
Reactivity VL (%MVIC)	0.0328*	0.0469*	0.0862	0.0221*
Reactivity VM (%MVIC)	0.0025*	0.0017*	0.0022*	0.0005*
Reactivity MH (%MVIC)	0.0005*	0.0059*	0.0004*	0.0009*
Reactivity LH (%MVIC)	<0.0001*	0.0016*	0.0010*	0.0152*

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), proximal anterior tibial shear force (PATSF), threshold to detect passive motion (TTDPM), peak torque (PT), flexion (flex), extension (ext)

*Significant ($p < 0.05$)

4.2.3 Repeated Measures Between Jump Distances

Summary statistics for each variable at each jump distance are reported in **Table 4**. Repeated measures ANOVA revealed significant differences among jump distances for all variables except knee flexion and abduction at initial contact, and peak proximal anterior tibial shear force (**Table 5**). Post-hoc pairwise testing (**Table 6**) revealed vertical and anterior-posterior ground reaction force significantly increased with each increase in jump distance, which demonstrated increased landing demand as jump distance increased.

Among the biomechanical characteristics only peak knee flexion was significantly different between jump distances of 20% and 40%. It steadily increased as jump distance increased from 79.4° at 20% to 89.8° at 80% ($p < 0.0001$) (**Figure 7E**). Peak knee abduction angle also showed significant increases as jump distance increased but only between 20% vs, 60%, 20% vs. 80%, 40% vs. 60%, and 40% vs. 80% ($p = 0.0162$, $p = 0.0021$, $p = 0.0121$, $p = 0.0025$, respectively) (**Figure 7F**). Peak knee abduction moment during landing remained consistent between the 20% and 40% jump distances (0.661 and 0.770 Nm/kg*m, respectively) but significantly increased to 1.114 and 1.226 Nm/kg*m at 60% and 80% jump distances, respectively (**Figure 6G**).

There was no significant difference in preactivity and reactivity of the quadriceps or hamstrings between jump distances of 20% and 40% body height. However, there was a steady increase in preactivity of the quadriceps and hamstrings as jump distance increased from 40% to 60% to 80% body height ($p < 0.0001$, **Figure 7A-D**). Muscle reactivity also exhibited increases with increases in jump distance (**Figure 7E-H**). Significant differences in reactivity were seen between jump distances of 20% and 60% ($p = 0.0027 – 0.008$) and between 20% and 80% jump distances ($p =0.0038 – 0.0057$). Reactivity of the lateral hamstring was the only muscle that did not show a significant increase between jump distances of 20% and 80% body height (50% compared to 53.5 %MVIC, p = 0.0097).

Table 4. Descriptive Statistics for the Kinematic, Kinetic, and Muscle Activation Variables for each Jump Distances

(A)	20% Height Jump Distance				
	Mean	SD	Median	25th%tile	75th%tile
Biomechanics					
Peak vGRF (%BW)	192.30	45.09	184.51	159.04	207.87
Peak apGRF (%BW)	42.102	11.237	39.486	33.272	50.285
Knee Flexion at IC (degrees)	26.022	6.725	27.028	22.114	30.523
Knee Abduction at IC (degrees)	-4.378	-4.348	-4.331	-1.866	-7.155
Peak Knee Flexion (degrees)	79.437	8.406	78.313	74.571	82.273
Peak Knee ABD (degrees)	-0.284	-6.490	-1.222	4.352	-6.120
Peak Knee ABDmom (Nm/kg*m)	0.752	0.468	0.661	0.456	1.017
Peak PATSF (%BW)	85.783	20.423	82.163	71.478	96.238
EMG					
Preactivity VL (%MVIC)	14.462	9.420	11.436	8.181	18.029
Preactivity VM (%MVIC)	12.247	8.263	10.159	6.854	14.299
Preactivity MH (%MVIC)	7.688	4.518	7.158	4.030	10.032
Preactivity LH (%MVIC)	13.763	8.994	11.007	7.296	18.931
Reactivity VL (%MVIC)	62.959	35.181	54.949	37.306	84.858
Reactivity VM (%MVIC)	55.488	34.332	45.352	30.828	77.158
Reactivity MH (%MVIC)	35.382	22.520	30.770	20.785	53.615
Reactivity LH (%MVIC)	63.863	46.990	47.998	34.476	83.832

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

Table 4 (Continued)

(B)	40% Height Jump Distance				
	Mean	SD	Median	25th%tile	75th%tile
Biomechanics					
Peak vGRF (%BW)	213.282	55.170	196.739	171.063	246.188
Peak apGRF (%BW)	52.142	11.740	50.249	44.881	61.602
Knee Flexion at IC (degrees)	24.040	5.806	24.160	21.151	26.939
Knee Abduction at IC (degrees)	-4.590	-4.092	-3.733	-1.811	-7.272
Peak Knee Flexion (degrees)	82.548	9.014	82.213	76.487	86.727
Peak Knee ABD (degrees)	-0.019	-6.654	-0.586	-4.992	5.100
Peak Knee ABDmom (Nm/kg*m)	0.840	0.522	0.770	0.403	1.223
Peak PATSF (%BW)	82.617	21.874	77.974	70.974	92.974
EMG					
Preactivity VL (%MVIC)	16.073	11.744	13.536	8.177	17.845
Preactivity VM (%MVIC)	13.847	10.680	11.172	6.752	15.505
Preactivity MH (%MVIC)	8.422	5.320	7.502	4.581	9.896
Preactivity LH (%MVIC)	15.280	12.408	11.711	7.218	19.183
Reactivity VL (%MVIC)	65.982	36.850	57.286	41.131	84.509
Reactivity VM (%MVIC)	58.748	38.256	47.545	32.711	86.300
Reactivity MH (%MVIC)	37.077	23.581	30.223	20.223	50.284
Reactivity LH (%MVIC)	65.747	45.820	56.797	28.603	94.131

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

Table 4 (Continued)

(C)	60% Height Jump Distance				
	Mean	SD	Median	25th%tile	75th%tile
Biomechanics					
Peak vGRF (%BW)	258.100	63.680	249.979	207.668	304.185
Peak apGRF (%BW)	64.350	13.748	63.332	53.974	72.113
Knee Flexion at IC (degrees)	25.121	5.909	25.050	21.354	28.705
Knee Abduction at IC (degrees)	-4.499	-3.981	-4.166	-6.886	-2.117
Peak Knee Flexion (degrees)	85.445	9.030	84.679	78.997	91.312
Peak Knee ABD (degrees)	0.808	-6.856	-0.288	-3.435	5.735
Peak Knee ABDmom (Nm/kg*m)	1.089	0.628	1.114	0.604	1.455
Peak PATSF (%BW)	83.108	20.499	82.136	70.170	91.639
EMG					
Preactivity VL (%MVIC)	19.847	12.341	16.229	11.876	24.474
Preactivity VM (%MVIC)	17.071	11.531	13.377	9.061	22.099
Preactivity MH (%MVIC)	10.526	6.013	9.582	6.490	13.015
Preactivity LH (%MVIC)	18.606	11.565	15.768	10.069	23.465
Reactivity VL (%MVIC)	68.159	35.539	66.609	44.303	85.798
Reactivity VM (%MVIC)	59.972	36.262	48.677	35.978	77.358
Reactivity MH (%MVIC)	38.384	24.076	35.645	22.772	50.231
Reactivity LH (%MVIC)	67.978	45.331	57.259	34.003	96.046

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

Table 4 (Continued)

(D)	80% Height Jump Distance				
	Mean	SD	Median	25th%tile	75th%tile
Biomechanics					
Peak vGRF (%BW)	303.600	73.321	299.020	255.557	358.684
Peak apGRF (%BW)	84.205	22.458	80.390	71.273	88.668
Knee Flexion at IC (degrees)	24.493	5.328	25.070	21.581	27.948
Knee Abduction at IC (degrees)	-5.088	-4.103	-5.115	-7.189	-2.058
Peak Knee Flexion (degrees)	89.783	8.927	89.222	82.172	96.374
Peak Knee ABD (degrees)	1.406	-7.430	1.005	-4.527	7.493
Peak Knee ABDmom (Nm/kg*m)	1.435	0.974	1.226	0.879	1.823
Peak PATSF (%BW)	82.540	15.399	83.014	76.522	89.128
EMG					
Preactivity VL (%MVIC)	24.755	14.910	19.459	15.895	31.841
Preactivity VM (%MVIC)	21.332	14.145	16.336	13.153	28.054
Preactivity MH (%MVIC)	12.766	6.633	11.498	8.610	16.538
Preactivity LH (%MVIC)	22.829	13.533	19.708	14.517	26.816
Reactivity VL (%MVIC)	69.344	35.040	59.760	45.941	86.484
Reactivity VM (%MVIC)	60.375	34.477	46.391	35.897	84.519
Reactivity MH (%MVIC)	37.530	21.110	34.028	25.474	48.525
Reactivity LH (%MVIC)	66.227	38.885	53.522	37.393	94.623

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

Table 5. Repeated Measures ANOVA Across Jump Distances

ANOVA Results

Variable	F-value	Chi-Squared[b]	p-value
Biomechanics			
Peak vGRF (%BW)[a]	89.107	-	<0.0001*
Peak apGRF (%BW)[a]	136.077	-	<0.0001*
Knee Flexion at IC (degrees)[a]	2.721	-	0.0600
Knee Abduction at IC (degrees)[a]	2.043	-	0.1320
Peak Knee Flexion (degrees)	77.111	-	<0.0001*
Peak Knee ABD (degrees)[a]	7.043	-	0.0020*
Peak Knee ABDmom (Nm/kg*m)	-	50.671	<0.0001*
Peak PATSF (%BW)[a]	2.587	-	0.0670
EMG			
Preactivity VL (%MVIC)	-	106.788	<0.0001*
Preactivity VM (%MVIC)	-	106.788	<0.0001*
Preactivity MH (%MVIC)	-	106.788	<0.0001*
Preactivity LH (%MVIC)	-	106.788	<0.0001*
Reactivity VL (%MVIC)	3.570	-	0.0220*
Reactivity VM (%MVIC)	-	16.788	0.0010*
Reactivity MH (%MVIC)	-	16.788	0.0010*
Reactivity LH (%MVIC)	-	16.788	0.0010*

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

[a] Greenhouse-Geisser adjustment for sphearisity violation

[b] Chi-squared values reported for Friedman ANOVA results

*Significant difference

Table 6. Post-hoc Pairwise Analysis Between Jump Distances

(A)	20% vs 40%		20% vs 60%	
Variable	Test Statistic[b]	p-value	Test Statistic[b]	p-value
Biomechanics				
Peak vGRF (%BW)	-3.822	0.003*	-8.963	<0.0001*
Peak apGRF (%BW)	-7.813	<0.0001*	-11.502	<0.0001*
Knee Flexion at IC (degrees)	3.410	0.0013*	1.404	0.1662
Knee Abduction at IC (degrees)	0.879	0.3833	0.375	0.7092
Peak Knee Flexion (degrees)	-5.397	<0.0001*	-8.858	<0.0001*
Peak Knee ABD (degrees)	-0.757	0.4491	-2.404	0.0162*
Peak Knee ABDmom (Nm/kg*m)[a]	-1.686	0.0917	-4.387	<0.0001*
Peak PATSF (%BW)	2.179	0.2890	2.111	0.4130
EMG				
Preactivity VL (%MVIC)[a]	-2.032	0.0422	-5.821	<0.0001*
Preactivity VM (%MVIC)[a]	-2.094	0.0363	-5.830	<0.0001*
Preactivity MH (%MVIC)[a]	-2.085	0.0371	-5.785	<0.0001*
Preactivity LH (%MVIC)[a]	-1.881	0.0599	-5.750	<0.0001*
Reactivity VL (%MVIC)	-1.664	0.6650	-2.651	0.0610*
Reactivity VM (%MVIC)[a]	-2.280	0.0226	-2.730	0.0041*
Reactivity MH (%MVIC)[a]	-2.067	0.0387	-2.997	0.0027*
Reactivity LH (%MVIC)[a]	-2.111	0.0347	-2.713	0.0067*

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

[a] Z statistic and Bonferonni manually adjusted p-value criteria ($p<0.0083$) shown

[b] Test statistic is either t statistic or z statistic based on ANOVA test used, Bonferonni adjustment used for all p-values

* Significant difference

Table 6 (Continued)

(B)	20% vs 80%		40% vs 60%	
Variable	Test Statistic[b]	p-value	Test Statistic[b]	p-value
Biomechanics				
Peak vGRF (%BW)	-11.973	<0.0001*	-6.690	<0.0001*
Peak apGRF (%BW)	-15.583	<0.0001*	-9.476	<0.0001*
Knee Flexion at IC (degrees)	1.906	0.1180	-2.333	0.0235*
Knee Abduction at IC (degrees)	1.736	0.0880	-0.367	0.7150
Peak Knee Flexion (degrees)	-12.330	<0.0001*	-4.651	<0.0001*
Peak Knee ABD (degrees)	-3.074	0.0021*	-2.510	0.0121*
Peak Knee ABDmom (Nm/kg*m)[a]	-4.799	<0.0001*	-4.873	<0.0001*
Peak PATSF (%BW)	2.387	0.125	-0.377	1.0000
EMG				
Preactivity VL (%MVIC)[a]	-6.158	<0.0001*	-5.183	<0.0001*
Preactivity VM (%MVIC)[a]	-6.140	<0.0001*	-5.042	<0.0001*
Preactivity MH (%MVIC)[a]	-6.168	<0.0001*	-5.077	<0.0001*
Preactivity LH (%MVIC)[a]	-6.130	<0.0001*	-5.042	<0.0001*
Reactivity VL (%MVIC)	-2.522	0.0890	-1.205	1.0000
Reactivity VM (%MVIC)[a]	-2.765	0.0057*	-0.097	0.3324
Reactivity MH (%MVIC)[a]	-2.859	0.0043*	-1.226	0.2202
Reactivity LH (%MVIC)[a]	-2.587	0.0097	-1.306	0.1916

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

[a] Z statistic and Bonferonni manually adjusted p-value criteria (p<0.0083) shown

[b] Test statistic is either t statistic or z statistic based on ANOVA test used, Bonferonni adjustment used for all p-values

* Significant difference

Table 6 (Continued)

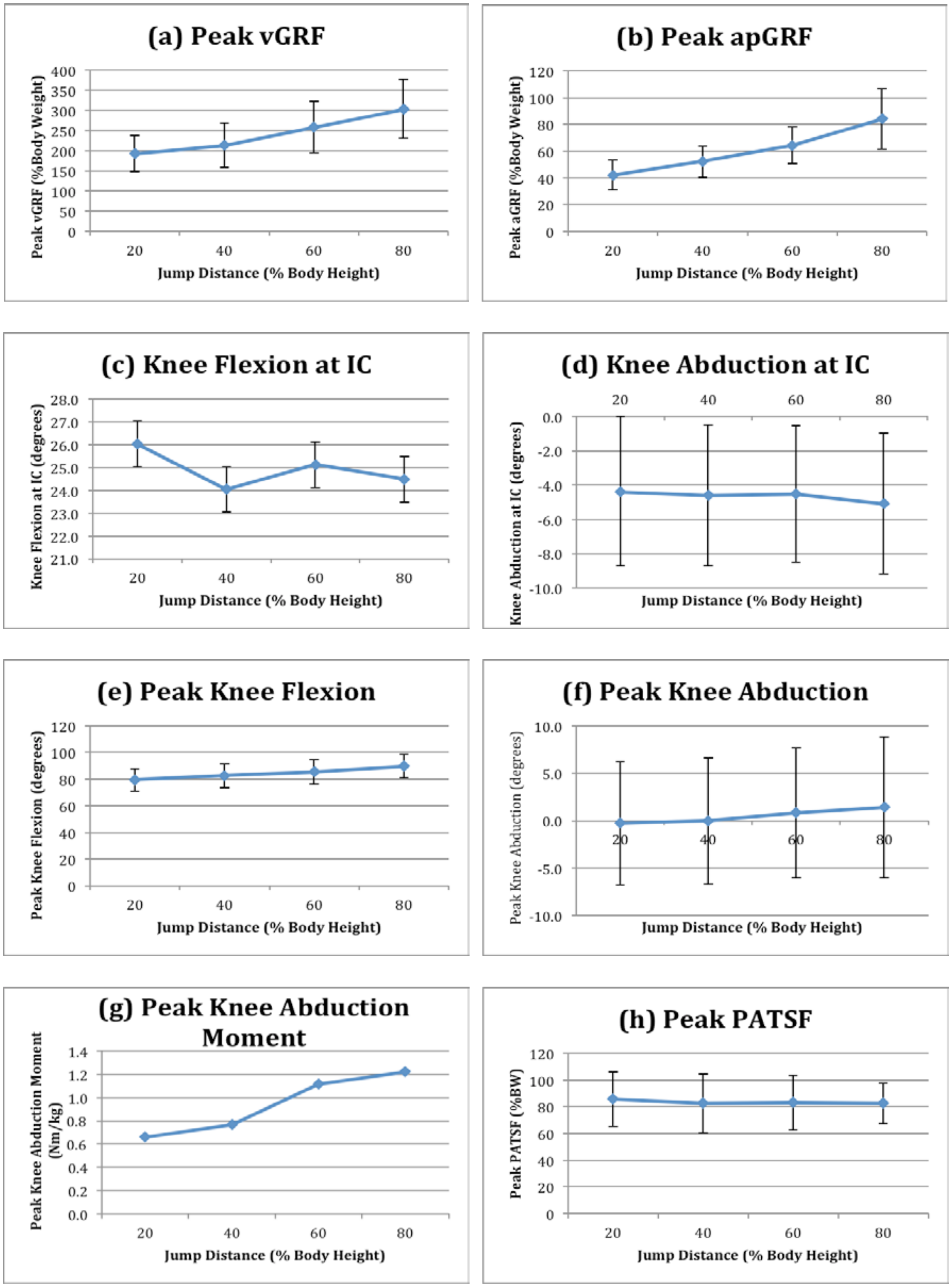
(C)	40% vs 80%		60% vs 80%	
Variable	Test Statistic[b]	p-value	Test Statistic[b]	p-value
Biomechanics				
Peak vGRF (%BW)	-12.685	<0.0001*	-5.931	<0.0001*
Peak apGRF (%BW)	-11.543	<0.0001*	-7.158	<0.0001*
Knee Flexion at IC (degrees)	-0.635	0.5282	0.893	0.3763
Knee Abduction at IC (degrees)	1.479	0.1455	2.120	0.0390*
Peak Knee Flexion (degrees)	-9.179	<0.0001*	-6.617	<0.0001*
Peak Knee ABD (degrees)	-3.028	0.0025*	-1.509	0.1313
Peak Knee ABDmom (Nm/kg*m)[a]	-5.127	<0.0001*	-3.384	0.0007*
Peak PATSF (%BW)	0.585	1.0000	1.265	1.0000
EMG				
Preactivity VL (%MVIC)[a]	-5.924	<0.0001*	-5.483	<0.0001*
Preactivity VM (%MVIC)[a]	-5.896	<0.0001*	-5.437	<0.0001*
Preactivity MH (%MVIC)[a]	-6.037	<0.0001*	-5.549	<0.0001*
Preactivity LH (%MVIC)[a]	-5.840	<0.0001*	-5.643	<0.0001*
Reactivity VL (%MVIC)	-1.389	1.0000	-0.566	1.0000
Reactivity VM (%MVIC)[a]	-1.640	0.1009	-0.122	0.9030
Reactivity MH (%MVIC)[a]	-1.697	0.0898	0.028	0.9776
Reactivity LH (%MVIC)[a]	-1.772	0.0765	-0.037	0.9701

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), maximum voluntary isometric contraction (MVIC), vastus lateralis (VL), vastus medialis (VM), medial Hamstring (MH), lateral hamstring (LH), standard deviation (SD)

[a] Z statistic and Bonferonni manually adjusted p-value criteria ($p<0.0083$) shown

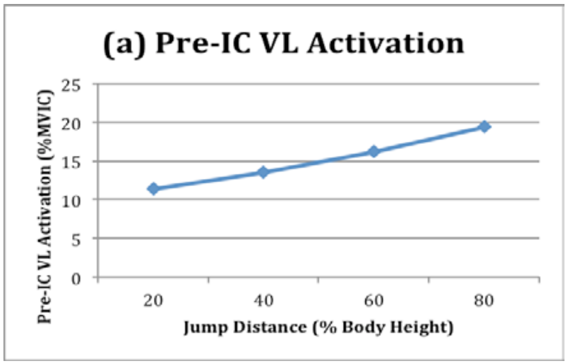
[b] Test statistic is either t statistic or z statistic based on ANOVA test used, Bonferonni adjustment used for all p-values

* Significant difference



*Median Values Reported

Figure 7. Biomechanical Characteristics Across Jump Distances



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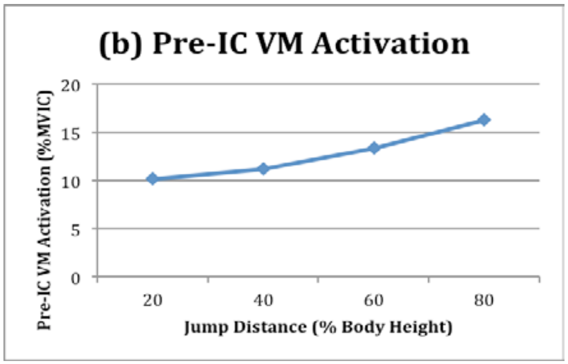
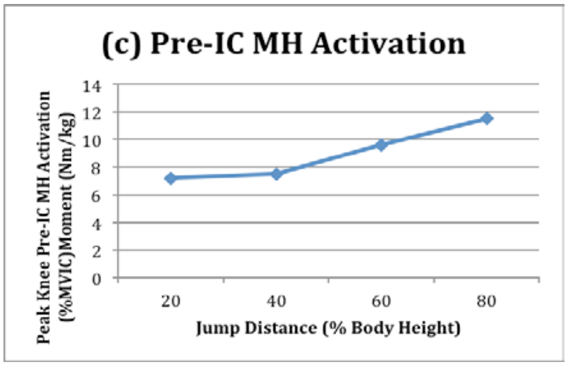


*Median Values Reported



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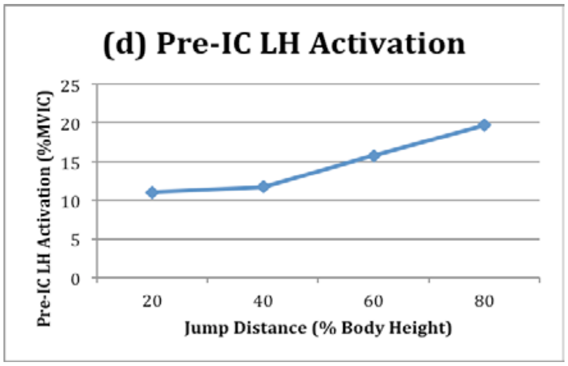

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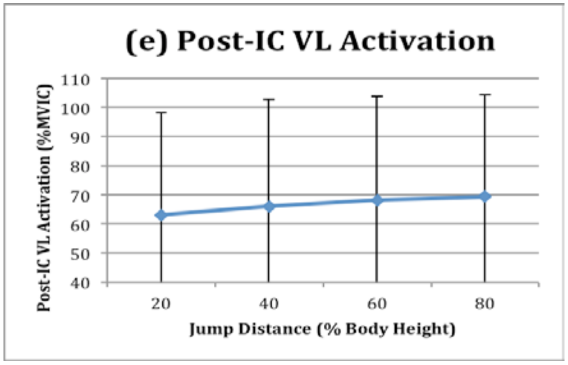
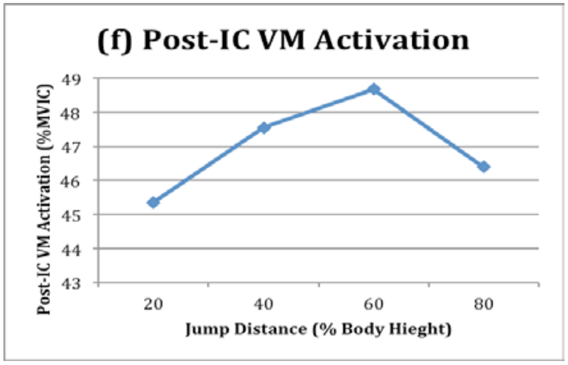
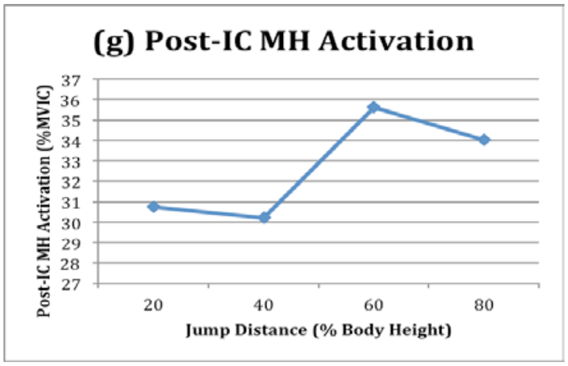
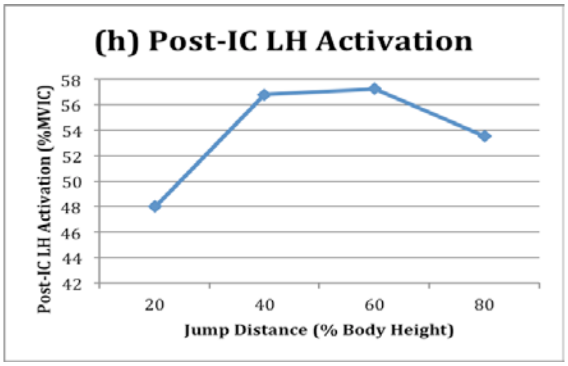




*Median Values Reported



*Median Values Reported



*Median Values Reported

Figure 8. Muscle Activation Across Jump Distances

4.3 RELATIONSHIP BETWEEN SENSORIMOTOR CHARACTERISTICS AND BIOMECHANICAL RISK FACTORS FOR ACL INJURY

The second specific aim of this study was to determine if components of the sensorimotor system could significantly predict the expression of biomechanical characteristics related to ACL injury and if this relationship changes with jump distance. Multiple linear regression analyses were used to assess the linear relationship between sensorimotor characteristics and biomechanical characteristics during landing and to determine sensorimotor characteristics that have a significant linear relationship with each of the biomechanical variables related to ACL injury. To appropriately assess these analyses, univariate statistics were first calculated to present point estimate and distributive characteristics of the dependent and independent variables. Bivariate analyses were then completed to assess correlation, simple regression, and assumptions of linearity, homoscedasticity, and potential outliers. Lastly, multiple linear regression models were used to obtain the final multiple linear regression models for each dependent variable at each jump distance.

4.3.1 Univariate Analysis

The dependent variables that were used in this specific aim were the eight biomechanical characteristics related to ACL injury; these variables included vertical ground reaction force, anterior-posterior ground reaction force, knee flexion angle at initial contact, knee abduction angle at initial contact, peak knee flexion angle during landing, peak knee abduction angle during landing, peak knee abduction moment during landing, and peak proximal anterior tibial shear force during landing. Descriptive statistics of the dependent variables are presented for each jump distance in **Table 4**. Normality testing revealed that only knee flexion at initial contact, knee abduction at initial contact, and peak knee abduction were normally distributed at all jump distances (**Table 7**). However, based on the distribution of the dependent variable shown in histograms (**Appendix B)** the only variable to show possible deviation from normality was peak knee abduction moment. All others did not show any obvious deviations from normality. Although Normality of the raw data is not an assumption of regression, the normality of the residuals will be assessed and addressed at a later point, after the model is applied.

Table 7. Normality of Dependent Variables using Shapiro-Wilk Test

	Jump Distance			
	20%	40%	60%	80%
	p-value	p-value	p-value	p-value
Peak vGRF (%BW)	0.0177	0.0045	0.2815	0.4894
Peak apGRF (%BW)	0.0230	0.3577	0.0834	0.0000*
Knee Flexion at IC (degrees)	0.0950	0.7951	0.9463	0.7976
Knee Abduction at IC (degrees)	0.8330	0.8853	0.5388	0.2644
Peak Knee Flexion (degrees)	0.0002	0.0362	0.1369	0.4975
Peak Knee ABD (degrees)	0.0849	0.3514	0.3899	0.1555
Peak Knee ABDmom (Nm/kg*m)	0.0111	0.1954	0.0400	0.0000
Peak PATSF (%BW)	0.1777	0.0541	0.0832	0.0056

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), proximal anterior tibial shear force (PATSF), threshold to detect passive motion (TTDPM), peak torque (PT), flexion (flex), extension (ext)

Transformation of peak knee abduction moment was attempted to account for the deviation from normality in the original data. Log transformation was unsuccessful in restoring normality, however, using the square root of peak knee abduction moment worked well. Shapiro-Wilk test results (p = 0.1755 – 0.9108) and histograms (**Figure 9**) showed no obvious deviations from normality.

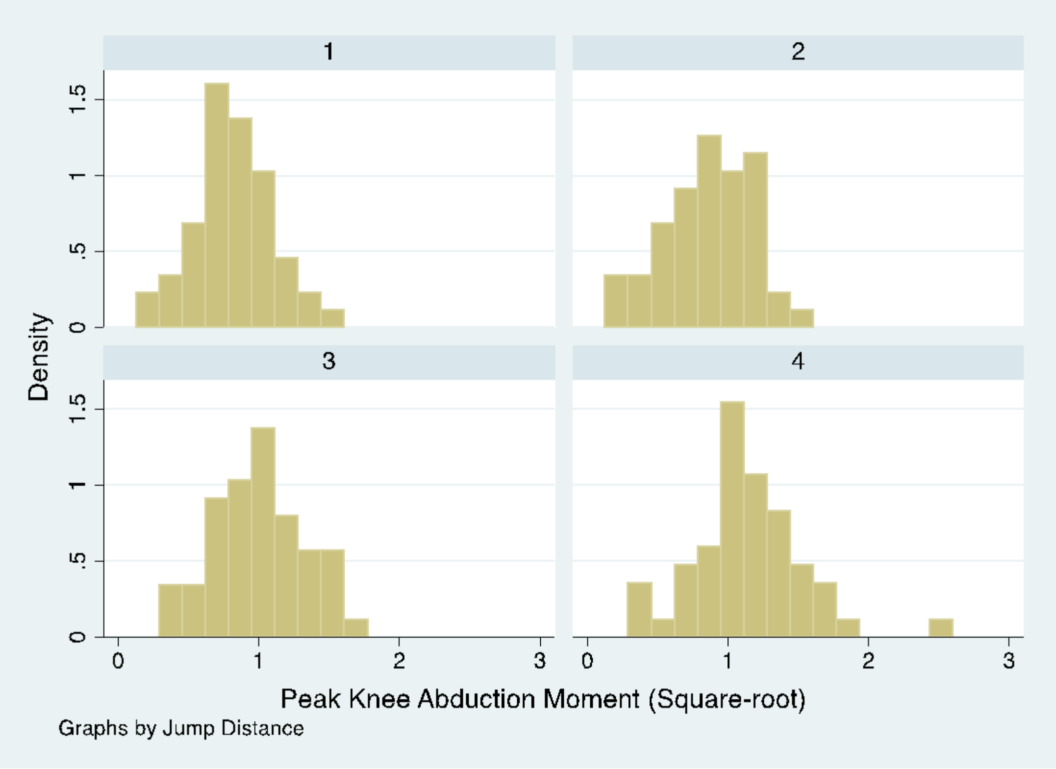


Figure 9. Histogram of Peak Knee Abduction Moment Square Root Transformation

Independent variables assessed for this specific aim included knee threshold to detect passive motion toward extension (TTDPMext), knee threshold to detect passive motion toward flexion (TTDPMflex), knee extension time to peak torque (TTPText), knee flexion time to peak torque (TTPTflex), knee extension peak torque (PText), knee flexion peak torque (PTflex). Descriptive statistics of these independent variables are presented in **Table 8**. Normality testing revealed that TTDPMext, TTDPMflex, and TTPTflex were not normally distributed (**Table 9**).

Table 8. Descriptive Statistics of Sensorimotor Characteristics

	Mean	SD	Median	IQR	25th%	75th%
TTDPMext (°)	2.48	2.77	1.62	1.78	1.00	2.78
TTDPMflex (°)	1.75	1.55	1.34	1.26	0.82	2.08
TTPText (ms)	121.89	17.98	120.00	20.00	110.00	130.00
TTPTflex (ms)	247.92	84.93	290.00	150.00	150.00	300.00
PText (%Body Weight)	109.75	28.17	110.22	37.80	91.20	129.00
PTflex (%Body Weight)	86.86	19.01	86.70	22.90	74.00	96.90

Abbreviations: threshold to detect passive motion toward extension (TTDPMext), threshold to detect passive motion toward flexion (TTDPMflex), knee extension time to peak torque (TTPText), knee flexion time to peak torque (TTPTflex), milliseconds (ms), knee extension peak torque (PText), knee flexion peak torque (PTflex), standard deviation (SD), inter quartile range (IQR)

Table 9. Normality of Sensorimotor Characteristics

	p - value
TTDPMext (°)	0.0000*
TTDPMflex (°)	0.0000*
TTPText (ms)	0.7629
TTPTflex (ms)	0.0000*
PText (%Body Weight)	0.9485
PTflex (%Body Weight)	0.0502

Abbreviations: threshold to detect passive motion toward extension (TTDPMext), threshold to detect passive motion toward flexion (TTDPMflex), knee extension time to peak torque (TTPText), knee flexion time to peak torque (TTPTflex), milliseconds (ms), knee extension peak torque (PText), knee flexion peak torque (PTflex)

*Significant ($p < 0.05$)

4.3.2 Bivariate Analysis

Two-way scatter plots were created for each of the dependent and independent variable pairs to assess linearity of the relationship, homoscedasticity, and any potential outliers within the data (**Appendix C**). Scatter plots of TTDPMext (**Appendix C.1**) did not show any signs of a non-linear relationship or violations of homoscedasticity but there was an outlier in the data. Subject 21 demonstrated extremely high TTDPMext and TTPMflex (**Appendix C.2**) values, however TTDPMext was much worse. These data points were verified in the hand written subject report and there was no reason to assume that the data from this subject was incorrect or should be omitted. Plots of the TTDPMflex comparison do not suggest any non-linear relationships but a few TTDPMflex data points are skewed to the right throughout each jump distance, which may influence homoscedasticity. TTPText (**Appendix C.3**) plots did not show any obvious signs of non-linear relationships, violations of homoscedasticity, or outliers. TTPTflex (**Appendix C.4**) did appear to be separated into three approximate groups. To account for this TTPTflex needed to be redefined into tertiles. The three groups appear to best be defined by three distinct criteria (TTPTflex G1 <50ms; TTPTflex G2 ≥50, ≤200; and TTPTflex G3 <200). The effect of this transformation will be assessed later, after the model is applied. Lastly, plots of PText (**Appendix C.5**) and PTflex (**Appendix C.6**) did not show any sign of non-linear relationships, violations of homoscedasticity, or outliers.

Pairwise correlations between each independent variable were assessed to determine the potential for any multicollinearity issues that may occur (**Table 10**). There were a few significant correlations between the pairs of independent variables TTDPM, TTPT, and PT. However, the correlation between TTDPMext and TTDPMflex was the only comparison that had a strong enough correlation that may have caused issues during the multiple linear regression modeling (r = 0.833, p = 0.000).

Table 10. Correlation of Independent Variables

	TTDPMext	TTDPMflex	TTPText	TTPTflex	PText
TTDPMflex	0.833[a]	1			
TTPText	0.047[a]	0.056[a]	1		
TTPTflex	0.106[a]	0.052[a]	0.423[a]	1	
PText	-0.240[a]	-0.177[a]	-0.508	-0.158	1
PTflex	-0.140[a]	-0.084[a]	-0.087	0.117	0.504

Abbreviations: Threshold to detect passive motion (TTDPM), peak torque (PT), flexion (flex), extension (ext)
[a]Spearman correlation coefficient

Pairwise correlations between each of these variables were also calculated to assess the strength of their linear relationship. These correlations are reported in **Appendix D**. Knee flexion angle at initial contact had by far the most number of significant correlations with sensorimotor characteristics at the knee. At a jump distance of 20% body height, knee flexion at initial contact was significantly correlated with TTDPMflex and TTPTflex (r = -.319, p = .020 and r = .400, p = .003, respectively). At a jump distance of 40% body height only TTPTflex was significantly correlated with knee flexion at initial contact (r = 0.328, p = 0.017). At a jump distance of 60% body height, knee flexion at initial contact was again significantly correlated with TTDPMflex and TTPTflex (r = -.273, p = .048 and r = .358, p = .009, respectively). At a jump distance of 80% body height, knee flexion at initial contact was significantly correlated with TTDPMext, TTDPMflex, PText, and PTflex (r = -.418, p = .002; r = -.397, p = .004; r = .324, p = .020; and r = .318, p = .023, respectively). Peak knee flexion also displayed significant correlations at jump distances of 40% body height with PText (r = 0.291, p = 0.035) and at 80% body height with TTDPMext (r = -.279, p = 0.047).

Simple linear regression models were completed to assess the linear relationship of each of the independent variables with the dependent variables (**Appendix E**). Jackknife residual plots (predicted values versus jackknife residuals) were created to evaluate potential outliers and homoscedasticity of the simple regression models (**Appendix F**). Overall the residual plots are central to zero for the jackknife residuals. However, there are significant outliers in the residual plots (> 3 or < -3) of peak vertical ground reaction force, peak anterior-posterior ground reaction force, peak knee abduction moment, and peak proximal anterior tibial shear force for simple linear regression models. These outliers were confirmed using Cook's D by demonstrating values greater than 1. Therefore robust regression was run on these specific models (**Appendix G**).

Similar to the correlation results, knee flexion at initial contact displayed significant linear relationships at all jump distances. At a jump distance of 20% body height, the only significant linear relationships with knee flexion at initial contact were with TTDPMflex and TTPTflex (R^2 = 0.1735 and R^2 = 0.1543). Peak knee flexion and PText was also shown to have a significant linear relationship (R^2 = 0.0844) at this jump distance. TTPTflex was significantly related to knee flexion at initial contact (R^2 = 0.1262) at a jump distance of 60% body height. At a jump distance of 80% body height, knee flexion at initial contact had a significantly linear relationship with TTDPMext, TTDPMflex, PText and PTflex (R^2 = 0.1539, R^2 = 0.1460, R^2 = 0.1050, R^2 = 0.1013, respectively). TTDPMext was also significantly related to peak knee flexion at a jump distance of 80% body height (R^2 = 0.0781). Peak knee abduction moment showed significant relationships with TTPTflex grouped variable at jump distances of 20% and 40% body height (**Appendix E.7**). Although the transformed variable (square root of the peak knee abduction moment) showed similar results (**Appendix E.9**), the transformed variable demonstrated lower MSE values and higher adjusted R^2 values for TTPTflex compared to the raw data, suggesting a better model fit.

Similarly, The transformation of TTPTflex was successful as is resulted in significant relationships, lower MSE values, and higher adjusted R^2 values as compared to the raw data.

4.3.3 Multiple Linear Regression Models

Separate multiple linear regression equations were constructed for each of the dependent variables at each of the four jump distances using a backwards-stepwise technique. Variance inflation factor (VIF) tables were created for each multivariate linear regression equation (full model and candidate model) to assess for multicollinearity. Histograms and scatterplots of residual vs. fitted values were created to assess normality of residuals and homoscedasticity.

4.3.3.1 Peak Vertical Ground Reaction Force

The candidate models at each of the jump distances for peak vertical ground reaction force are shown in **Table 11**. No independent variables were retained in the model at a jump distance of 20% body height. Although TTPTflex was retained for the model at distances of 40%, 60%, and 80%, none were statistically significant. Peak vertical ground reaction force did not have a significant linear relationship with the examined sensorimotor characteristics of the knee. No significant outliers or leverage points were detected using residual or Cooks D analysis.

Table 11. Peak Vertical Ground Reaction Force Multiple Linear Regression Results

(**A**) Vertical GRF at 40% Body Height

Source	SS	df	MS	Observations	53
Model	0.52046	1	0.52046	F (1 , 2)	1.73
Residual	15.31011	51	0.30020	Prob > F	0.1938
Total	15.83058	52	0.30443	R^2	0.0329
				Adjusted R^2	0.0139
				MSE	0.5479

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.23678	-1.32	0.194
Constant	2.18644	25.55	0.000

(**B**) Vertical GRF at 60% Body Height

Source	SS	df	MS	Observations	53
Model	1.45038	1	1.45038	F (1 , 2)	3.77
Residual	19.63963	51	0.38509	Prob > F	0.0578
Total	21.09001	52	0.40558	R^2	0.0688
				Adjusted R^2	0.0505
				MSE	0.62056

Predictor Variables	Coefficients	t	p-value
TTPTflex G3	0.37522	1.94	0.058
Constant	2.30560	13.90	0.000

(**C**) Vertical GRF at 80% Body Height

Source	SS	df	MS	Observations	51
Model	1.72048	1	1.72048	F (1 , 2)	3.35
Residual	25.15927	49	0.51345	Prob > F	0.0733
Total	26.87976	50	0.53760	R^2	0.064
				Adjusted R^2	0.0449
				MSE	0.71656

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.4329997	-1.83	0.073
Constant	3.13796	27.35	0.000

4.3.3.2 Peak Anterior-Posterior Ground Reaction Force

The candidate models at each of the jump distances for peak anterior-posterior ground reaction force are shown in **Table 12**. No independent variables were retained in the model at a jump distance of 20% body height. TTPTflex was retained for the model at distance of 40% and 60%, but neither model was statistically significant. At a jump distance of 80% body height, TTDPMext, TTDPMflex, PText and TTPTflex (group 2) were retained in the model, however, the overall model was not statistically significant. Peak anterior-posterior ground reaction force did not have a significant linear relationship with the examined sensorimotor characteristics of the knee. No significant outliers or leverage points were detected using residual or Cooks D analysis.

Table 12. Peak Anterior-Posterior Ground Reaction Force Multiple Linear Regression Results

(**A**) Anterior-Posterior GRF at 40%

Source	SS	df	MS		
Model	0.02491	1	0.02491	Observations	53
Residual	0.69247	51	0.01358	F (1 , 2)	1.83
Total	0.71738	52	0.01380	Prob > F	0.1815
				R^2	0.0347
				Adjusted R^2	0.0158
				MSE	0.11652

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.05181	-1.35	0.182
Constant	0.53315	29.30	0.000

(**B**) Anterior-Posterior GRF at 60%

Source	SS	df	MS		
Model	0.06167	1	0.06167	Observations	53
Residual	0.92129	51	0.01806	F (1 , 2)	3.41
Total	0.98295	52	0.01890	Prob > F	0.0705
				R^2	0.0627
				Adjusted R^2	0.0444
				MSE	0.1344

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.08150	-1.85	0.070
Constant	0.66200	31.54	0.000

Table 12 (Continued)

(**C**) Anterior-Posterior GRF at 80%

Source	SS	df	MS		
Model	0.31288	4	0.07822	Observations	51
Residual	2.20896	46	0.04802	F (1 , 2)	1.63
Total	2.52184	50	0.05044	Prob > F	0.1831
				R^2	0.1241
				Adjusted R^2	0.0479
				MSE	0.21914

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.04052	-1.63	0.111
TTDPMflex	0.07380	1.64	0.107
PTflex	-0.26023	-1.50	0.141
TTPTflex G2	-0.14480	1.87	0.068
Constant	1.07466	6.29	0.000

4.3.3.3 Knee Flexion at Initial Contact

Multiple regression models created for knee flexion at initial contact were significant at each jump distance (**Table 13**). Sensorimotor characteristics TTDPMext, TTDPMflex, TTDPM, and TTPTflex (group 3) significantly accounted for 26.84% of the variance of knee flexion at initial contact at a jump distance of 20% body height (p = 0.0015). Only TTPTflex (group 3) was retained at a jump distance of 40% body height but significantly accounted for 8.16% of the variance of knee flexion at initial contact (p = 0.0381). TTDPMflex and TTPTflex (group 3) significantly accounted for 18.65% of the variance of knee flexion at initial contact at a jump distance of 60% body height (p = 0.0057). Lastly, TTDPMext, PText and TTPText significantly accounted for 26.72% of the variance of knee flexion at initial contact at a jump distance of 80% body height (p = 0.002). Variance inflation factors (VIFs) ranged from 1.00 to 5.13 showing no signs of multicollinearity in the final model. The residuals of all four models show no obvious deviation from normality (**Appendix I.3**). No significant outliers or leverage points were detected using residual or Cooks D analysis.

Final Models:

Knee flexion at initial contact at 20% jump distance = 0.992(TTDPMext) -2.877(TTDPMflex) + 4.991(TTPTflexG3) + 24.933

Knee flexion at initial contact at 40% jump distance = 3.728 (TTPTflexG3) + 21.297

Knee flexion at initial contact at 60% jump distance = -0.961(TTDPMflex) + 4.450(TTPTflexG3) + 23.532

Knee flexion at initial contact at 80% jump distance = -0.682(TTDPMext) + 0.073 (TTPText) + 6.463(PText) + 10.153

Table 13. Knee Flexion at Initial Contact Multiple Linear Regression Results

(**A**) Knee Flexion at Initial Contact at 20%

Source	SS	df	MS		
				Observations	53
Model	631.2286	3	210.40952	F (1 , 2)	5.99
Residual	1720.6583	49	35.11548	Prob > F	0.0015
Total	2351.8869	52	45.22859	R^2	0.2684
				Adjusted R^2	0.2236
				MSE	5.9258

Predictor Variables	Coefficients	t	p-value
TTDPMext	0.99220	1.48	0.145
TTDPMflex	-2.87654	-2.40	0.020
TTPTflex G3	4.99082	2.65	0.011
Constant	24.93253	12.83	0.000

Table 13 (Continued)

(**B**) Knee Flexion at Initial Contact at 40%

Source	SS	df	MS	Observations	53
Model	143.1404	1	143.14045	F (1 , 2)	4.53
Residual	1610.0951	51	31.57049	Prob > F	0.0381
Total	1753.2355	52	33.71607	R^2	0.0816
				Adjusted R^2	0.0636
				MSE	5.6188

Predictor Variables	Coefficients	t	p-value
TTPTflex G3	3.72754	2.13	0.038
Constant	21.29748	14.48	0.000

(**C**) Knee Flexion at Initial Contact at 60%

Source	SS	df	MS	Observations	53
Model	338.4986	2	169.24929	F (1 , 2)	5.73
Residual	1476.9288	50	29.53858	Prob > F	0.0057
Total	1815.4274	52	34.91206	R^2	0.1865
				Adjusted R^2	0.1539
				MSE	5.4349

Predictor Variables	Coefficients	t	p-value
TTPTflex G3	4.45016	2.62	0.012
TTDPMflex	-0.96143	-1.98	0.054
Constant	23.53174	1.72	0.000

(**D**) Knee Flexion at Initial Contact at 80%

Source	SS	df	MS	Observations	51
Model	379.2001	3	126.40005	F (1 , 2)	5.71
Residual	1040.0503	47	22.12873	Prob > F	0.002
Total	1419.2505	50	28.38501	R^2	0.2672
				Adjusted R^2	0.2204
				MSE	4.7041

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.68153	-2.78	0.008
PText	6.46293	2.32	0.025
TTPText	0.07343	1.70	0.096
Constant	10.15933	1.40	0.169

4.3.3.4 Knee Abduction at Initial Contact

None of the multiple linear regression models at any jump distance retained any of the sensorimotor characteristics.

4.3.3.5 Peak Knee Flexion

Multiple regression models created for peak knee flexion were significant only at 40% and 80% body height (**Table 14**). PText significantly accounted for 8.44% of the variance of knee flexion at initial contact at a jump distance of 20% body height (p = 0.0348). PText was also the only independent variable retained at 80% body height and accounted for 7.81% variance in peak knee flexion (p = 0.0471). The residuals from jump distances of 40%, 60%, and 80% showed no obvious deviation from normality but residuals at 20% may have been problematic (**Appendix I.4**). Transformation of peak knee flexion was performed using the reciprocal (1/peak knee flexion) (**Appendix I.4**). However, multiple linear regression model results from the transformed dependent variable did not change the results of the original data. Therefore, final models were constructed using the original dependent variable data. No significant outliers or leverage points were detected using residual or Cooks D analysis.

Final Models:

Peak knee flexion at 40% jump distance = 9.299 (PText) + 72.343

Peak knee flexion at 80% jump distance = -0.883(TTDPMext) + 91.991

Table 14. Peak Knee Flexion Multiple Linear Regression Results

(**A**) Peak Knee Flexion at 20%

Source	SS	df	MS	Observations	53
Model	264.1127	1	264.11273	F (1 , 2)	3.95
Residual	3410.0739	51	66.86419	Prob > F	0.0523
Total	3674.1866	52	70.65744	R^2	0.0719
				Adjusted R^2	0.0537
				MSE	8.1771

Predictor Variables	Coefficients	t	p-value
PText	8.00145	1.99	0.052
Constant	70.65502	15.50	0.000

(**B**) Peak Knee Flexion at 40%

Source	SS	df	MS	Observations	53
Model	356.6996	1	356.699587	F (1 , 2)	4.7
Residual	3868.3576	51	75.8501484	Prob > F	0.0348
Total	4225.0572	52	81.25110	R^2	0.0844
				Adjusted R^2	0.0665
				MSE	8.7092

Predictor Variables	Coefficients	t	p-value
PText	9.29876	2.17	0.035
Constant	72.34291	14.09	0.000

(**C**) Peak Knee Flexion at 60%

Source	SS	df	MS	Observations	53
Model	138.1470	1	138.14700	F (1 , 2)	1.72
Residual	4102.4066	51	80.43934	Prob > F	0.1959
Total	4240.5536	52	81.54911	R^2	0.0326
				Adjusted R^2	0.0136
				MSE	8.9688

Predictor Variables	Coefficients	t	p-value
PText	5.78688	1.31	0.196
Constant	79.09394	15.82	0.000

Table 14 (Continued)

(**D**) Peak Knee Flexion at 80%

Source	SS	df	MS		
Model	311.0770	1	311.07695	Observations	51
Residual	3673.5006	49	74.96940	F (1 , 2)	4.15
Total	3984.5776	50	79.69155	Prob > F	0.0471
				R^2	0.0781
				Adjusted R^2	0.0593
				MSE	8.6585

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.88297	-2.04	0.047
Constant	91.99094	56.57	0.000

4.3.3.6 Peak Knee Abduction

Linear regression models for peak knee abduction retained no independent variables at any jump distance.

4.3.3.7 Peak Knee Abduction Moment

Multiple linear regression model outputs from all jump distances except 80% body height were significant. At a jump distance of 20% body height TTPTflex (group 2) was the only independent variable retained and accounted for 11.31% on the variance of peak knee abduction moment (p = 0.0138). At a jump distance of 40% body height, TTDPMext, TTDPMflex, and TTPTflex (group 2) accounted for 21.67% of the variance of peak knee abduction moment (p = 0.0071). TTDPMext, TTDPMflex, and TTPTflex (group 2) were also retained at 60% body height but only accounted for 16.44% of the variance of peak knee abduction moment (p = 0.0308). VIFs ranged from 1.00 to 5.22 showing no obvious signs of multicollinearity. However, histograms of the residuals at 80% body height did seem to significantly skew to the right (**Appendix I.7**). This deviation from normality was identified as a potential problem earlier and transformations to the square root of peak knee abduction moment were performed and shown in **Table 16**. These models did not change the results of what independent variables were retained but it did result in lower MSE values and the model at 80% body height was significant. Additionally, the residuals at all four distances of the transformed models showed no obvious deviation from normality (**Appendix I.7**). Therefore, the final models were constructed using the results from the square root of peak knee abduction moment. Further analysis of potential outliers and leverage points using residual scatterplot analysis and Cook's D revealed that the final model at a jump distance of 60% body height did have a significant outlier. Robust regression model was run using the same predictor variables. This model still showed a significant relationship ($R^2 = 0.2622$. adjusted $R^2 = 0.2007$, $p = 0.0049$)

Final Models:

Square root (peak knee abduction moment) at 20% jump distance = -0.372(TTPTflexG2) + 0.836

Square root (peak knee abduction moment) at 40% jump distance = -0.079(TTDPMext) + 0.119(TTDPMflex) + 0.354(TTPTflexG2) + 0.945

Square root (peak knee abduction moment) at 60% jump distance = -0.058(TTDPMext) + 0.078(TTDPMflex) + 0.298(TTPTflexG2) + 0.783

Square root (peak knee abduction moment) at 60% jump distance (Robust Regression) = -0.059(TTDPMext) + 0.087(TTDPMflex) - 0.203(TTPTflexG1) + 0.189 (TTPTflexG2) + 0.889

Square root (peak knee abduction moment) at 80% jump distance = -0.255(TTPTflexG2) + 1.196

Table 15. Peak Knee Abduction Moment Multiple Linear Regression Results

(A) Peak Knee Abduction Moment at 20%

Source	SS	df	MS	Observations	53
Model	1.28605	1	1.28605	F (1 , 2)	6.5
Residual	10.08534	51	0.19775	Prob > F	0.0138
Total	11.37138	52	0.21868	R^2	0.1131
				Adjusted R^2	0.0957
				MSE	0.44469

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.37221	-2.55	0.014
Constant	0.83626	12.04	0.000

(**B**) Peak Knee Abduction Moment at 40%

Source	SS	df	MS	Observations	53
Model	3.06752	3	1.02251	F (1 , 2)	4.52
Residual	11.08756	49	0.22628	Prob > F	0.0071
Total	14.15509	52	0.27221	R^2	0.2167
				Adjusted R^2	0.1688
				MSE	0.47569

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.12665	-2.350	0.023
TTDPMflex	0.18072	1.86	0.069
TTPTflex G2	-0.50041	-3.11	0.003
Constant	0.94722	9.19	0.000

(**C**) Peak Knee Abduction Moment at 60%

Source	SS	df	MS	Observations	53
Model	3.36966	3	1.12322	F (1 , 2)	3.21
Residual	17.12179	49	0.34942	Prob > F	0.0308
Total	20.49145	52	0.39407	R^2	0.1644
				Adjusted R^2	0.1133
				MSE	0.59112

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.12811	-1.92	0.061
TTDPMflex	0.18275	1.53	0.133
TTPTflex G3	0.50903	2.70	0.009
Constant	0.71135	3.67	0.001

Table 15 (Continued)

(**D**) Peak Knee Abduction Moment at 80%

Source	SS	df	MS		
				Observations	51
Model	1.79776	1	1.79776	F (1 , 2)	1.93
Residual	45.65042	49	0.93164	Prob > F	0.1711
Total	47.44817	50	0.94896	R^2	0.0379
				Adjusted R^2	0.0183
				MSE	0.96522

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.44262	-1.39	0.171
Constant	1.53867	9.96	0.000

Table 16. Peak Knee Abduction Moment (Square Root) Multiple Linear Regression Results

(**A**) Peak Knee Abduction (Square Root) at 20%

Source	SS	df	MS		
				Observations	53
Model	0.432	1	0.43202	F (1 , 2)	6.31
Residual	3.494	51	0.06851	Prob > F	0.0152
Total	3.926	52	0.07550	R^2	0.11
				Adjusted R^2	0.0926
				MSE	0.26174

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.21573	-2.51	0.015
Constant	0.87220	21.34	0.000

(**B**) Peak Knee Abduction (Square Root) at 40%

Source	SS	df	MS		
				Observations	53
Model	1.379	1	0.45980	F (1 , 2)	6.99
Residual	3.157	51	0.06578	Prob > F	0.0005
Total	4.537	52	0.08895	R^2	0.3041
				Adjusted R^2	0.2606
				MSE	0.25647

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.07924	-2.73	0.01
TTDPMflex	0.11902	2.27	0.03
TTPTflex G2	0.35380	-4.06	0.00
Constant	0.94476	16.94	0.00

Table 16 (Continued)

(**C**) Peak Knee Abduction (Square Root) at 60%

Source	SS	df	MS	Observations	53
Model	1.032	1	0.34393	F (1 , 2)	4.17
Residual	4.041	51	0.08247	Prob > F	0.0104
Total	5.073	52	0.09756	R^2	0.2034
				Adjusted R^2	0.1546
				MSE	0.28718

Predictor Variables	Coefficients	t	p-value
TTDPMext	-0.05768	-1.78	0.082
TTDPMflex	0.07832	1.35	0.184
TTPT G3	0.29808	3.26	0.002
Constant	0.78284	8.31	0.000

(**D**) Peak Knee Abduction (Square Root) at 80%

Source	SS	df	MS	Observations	51
Model	0.598	1	0.59794	F (1 , 2)	4.3
Residual	0.681	49	0.13897	Prob > F	0.0433
Total	7.408	50	0.14815	R^2	0.0807
				Adjusted R^2	0.062
				MSE	0.37379

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-0.25527	-0.21	0.043
Constant	1.19553	20.03	0.000

4.3.3.8 Peak Proximal Anterior Tibial Shear Force

Multiple linear regression models constructed for peak proximal anterior tibial shear force did not result in any significant models at any jump distance (**Table 17**). Histograms were constructed for each of these models to examine the residuals but there was no obvious sign of deviation from normality (**Appendix I.8**). No significant outliers or leverage points were detected using residual or Cooks D analysis.

Table 17. Peak Proximal Anterior Tibial Shear Force Multiple Linear Regression Results

(**A**) Peak PATSF at 20%

Source	SS	df	MS	Observations	53
Model	706.178	1	706.17760	F (1 , 2)	1.72
Residual	20982.820	51	411.42784	Prob > F	0.196
Total	21688.998	52	417.09611	R^2	0.0326
				Adjusted R^2	0.0136
				MSE	20.284

Predictor Variables	Coefficients	t	p-value
PTflex	19.38343	1.31	0.196
Constant	68.94599	5.24	0.000

(**B**) Peak PATSF at 40%

Source	SS	df	MS	Observations	53
Model	1102.336	1	1102.33641	F (1 , 2)	2.36
Residual	23777.228	51	466.22016	Prob > F	0.1303
Total	24879.565	52	478.45317	R^2	0.0443
				Adjusted R^2	0.0256
				MSE	21.592

Predictor Variables	Coefficients	t	p-value
PTflex	24.21758	1.54	0.130
Constant	61.58111	4.40	0.000

(**C**) Peak PATSF at 60%

Source	SS	df	MS	Observations	53
Model	1100.664	1	1100.66400	F (1 , 2)	2.71
Residual	20750.846	51	406.87934	Prob > F	0.1062
Total	21851.510	52	420.22135	R^2	0.0504
				Adjusted R^2	0.0317
				MSE	20.171

Predictor Variables	Coefficients	t	p-value
PTflex	24.19920	1.64	0.106
Constant	62.08866	4.75	0.000

Table 17 (Continued)

(**D**) Peak PATSF at 80%

Source	SS	df	MS		
Model	835.625	1	835.62549	Observations	51
Residual	11020.288	49	224.90383	F (1 , 2)	3.72
Total	11855.913	50	237.11826	Prob > F	0.0597
				R^2	0.0705
				Adjusted R^2	0.0515
				MSE	14.997

Predictor Variables	Coefficients	t	p-value
TTPTflex G2	-9.54263	-1.93	0.060
Constant	84.78493	35.31	0.000

5.0 DISCUSSION

Current research suggests that similar tasks completed with different demands (i.e. jump distance) result in significantly different biomechanical characteristics during landing.[91] Therefore, there is a need to investigate the relationship between jump distance and biomechanical ACL risk factors to determine specific demands that are more relevant and biomechanically sensitive to specific landing characteristics. The purpose of this study was to assess the effect of jump distance on biomechanical risk factors for ACL injury and examine how the relationship between sensorimotor system characteristics and landing biomechanics change throughout jump distances.

Hypothesis 1: As jump distance increases the demand during landing will also increase, as expressed by a significant increase in vertical and posterior ground reaction forces. Secondly, as jump distance and landing demand increase there will also be a significant increase in the expression of ACL risk factors (increased peak knee abduction angle, knee abduction angle at initial contact, knee abduction moment, peak vertical and posterior ground reaction forces, and proximal anterior tibial shear force) and changes in knee joint loading patterns (increased peak knee flexion peak and knee flexion at initial contact, knee flexion moment, and proximal anterior tibial shear force). Lastly, it was hypothesized that as jump distance and landing demand increase muscle activation of the quadriceps and hamstrings during both preactivity and reactivity will also increase.

Hypothesis 2: Threshold to detect passive motion, time to peak torque, and peak torque will each independently contribute to the variance seen in knee flexion and abduction angles at initial contact, peak knee flexion and abduction angles, peak knee flexion moment, peak abduction moment, and peak proximal anterior tibial shear force. It was also hypothesized that threshold to detect passive motion, time to peak torque, and peak torque will together significantly contribute to the variance seen in knee flexion and abduction angles at initial contact, peak knee flexion and abduction angles, peak knee flexion moment, peak abduction moment, and peak proximal anterior tibial shear force.

5.1 SUBJECT CHARACTERISTICS

The subject recruitment aim of this study was to recruit healthy females with a variety of activity levels. Based on the reporting results from the Tegner activity scale this aim was successful. Subjects reported activity levels between three and nine reflecting low to very high activity levels related to the knee. Most commonly, previous studies investigating contributing factors to ACL injuries have focused on female team sports or a general healthy population.[31,177,214,215] The current population is similar to previous studies investigating landing biomechanics related to female ACL injury.[194,216,217] The only other study that has investigated the relationship between landing kinematics and sensorimotor characteristics at the knee used healthy male subjects.[148]

5.2 LANDING BIOMECHANICS DURING LANDING

Landing biomechanics were collected during a double-leg stop-jump task at four different jump distances. Peak ground reaction forces, knee kinematics and knee kinetics were collected during the initial landing phase of the stop-jump maneuver. Knee flexion and abduction angles were also collected at the time of initial contact with the force platform.

5.2.1 Peak Vertical and Anterior-Posterior Ground Reaction Forces

The average peak vertical ground reaction force seen in this study ranged from 192.3% to 303.6% body weight across the four different jump distances. Average peak anterior-posterior ground reaction forces ranged between 42.1% and 84.2% body weight. A study by Sell et al.[33] examined the effect of stop-jump direction on landing biomechanics and found fairly similar results dependent on jumping direction. Peak vertical ground reaction forces ranged from 145% to 292% body weight and peak anterior-posterior ground reaction forces ranged from 24% to 38% body weight.[33] Although vertical ground reaction forces and anterior-posterior ground reaction forces were higher in the current study, increased jump distance used in the current study likely produced increased landing demand and thus increased vertical and anterior-posterior ground reaction forces. The study by Sell et al.[33] used a constant jump distance of 40% body height whereas this study increased jump distance from 20% to 80% body height, effectively increasing landing intensity. A study by Norcross et al.[71] reported average peak vertical ground reaction forces during a double-leg drop-jump of 286% to 294% body weight.[71] These values are on the higher end compared to the range that the current study observed. However, the drop-jump task involves the subject beginning by jumping off of a 30-cm high box that is placed at 50% of the subject's body height away from the landing area.[71] This increased height likely instills a higher vertical landing demand. Average peak anterior-posterior ground reaction forces were seen up to 96% body weight, higher than the anterior-posterior ground reaction forces seen in the current study.[71] This increase in anterior-posterior ground reaction force is also likely a result of the same mechanism.

5.2.2 Knee Flexion at Initial Contact

The current study observed a range of average knee flexion angles at initial contact of 24° to 26°. These values are consistent with what has previously been reported in the literature in an active female population.[42,71] Norcross et al.[71] reported values of 23° during a double-leg drop-jump maneuver in females.[71] Yu et al.[42] reported knee flexion at initial contact values of 23.95° during a similar task as the one used in the current study.[42] The stop-jump protocol in Yu et al.[42] used a three step approach where the current study used a single broad-jump approach.[42] However, using this same landing protocol and a similar population that research group found only slightly different results. In 2007 Chappel et al.[31] reported that recreationally active females landed with 17° of knee flexion at initial contact using this three step approach stop-jump maneuver.[31]

5.2.3 Knee Abduction at Initial Contact

The current study found that knee abduction angles at initial contact in a healthy recreationally active female population were between -4.3° and -5.1°. This negative abduction value represents an average adduction (varus) knee angle at initial contact. Previous research has mixed results regarding this variable. Norcross et al.[71] reported an average knee abduction angle between 6.8°

and 7.7° (valgus).[71] This difference seen between the current study results and Norcross et al.[71] may be partially due to differences in data collection and biomechanical model methodology. The current study utilized a camera-based motion analysis system and the Plug-in-Gait biomechanical model with an Euler angle decomposition sequence of XYZ. Norcross et al.[71] used electromagnetic tracking and motion monitor with different Euler angle sequence (YXZ), calculating the motion in the frontal plane first.[71] Additionally, the task employed by Norcross et al.[71] was a drop-jump task. This combination of task difference and differences in the biomechanical modeling may explain the difference seen between these results. Chappell et al.[31] used a more similar task and biomechanical model and reported similar knee abduction angles at initial contact (-5.0°).[31]

5.2.4 Peak Knee Flexion

The current study found that average peak knee flexion ranged from 79.4° to 89.7° between the different jump distances. These values are similar to those reported in previous studies. Norcross et al.[71] reported average peak knee flexion values during a drop jump, ranging from 87.9° to 93.8°.[71] However, studies by Sell et al.[33] and Yu et al.[42] reported lower average peak knee flexion values (62.9° to 78.9° and 68.5° respectively). Additionally, Sell et al.[33] and Yu et al.[42] both used a stop-jump task, however, they did not control for the height of the initial jump.[33,42] The current study used a six-inch hurdle to ensure each subject was consistently jumping with a similar vertical height at each jump distance. This controlling factor may have produced more similar knee flexion responses to the drop-jump task used by Norcross et. al.[71] by ensuring a standard amount of vertical trajectory during the flight phase.

5.2.5 Peak Knee Abduction

The current study found that females landed with a range of average peak knee abduction angle between -0.2° and 1.4°, demonstrating a fairly neutral knee position during landing. However, this average neutral knee position varied by about 7° implying there were some subjects who landed in a knee abducted position (valgus) and some who landed in a knee adducted position (varus). Previous studies have reported average peak knee abduction values up to 18.1 degrees.[71] Sell et al.[33] used a similar stop-jump task and reported knee abduction angles ranging from 3.9° to 8.0° and Norcross et al.[71] reported average values ranging from 14.5° to 18.1°.[33,71] Although the current study had subjects who exhibited similar knee abduction angles to Sell et al.,[33] there were also subjects that demonstrated knee adduction angles that resulted in the central statistic being close to zero. Several aspects of the current study were different and may help explain this difference. The current study used a similar camera based system and biomechanical model as compared to Sell et al.,[33] however the marker sets that were used were slightly different and even slight variations in the calculated knee coordinate system may account for the differences between these two studies.[33] Both marker sets are susceptible to knee axis rotation error due to thigh marker position.[218] The difference between the current study findings and those of Norcross et al.[71] was much greater. This larger difference may be partially explained by the data collection method differences (electromagnetic tracking vs. passive marker based) but the difference in the task used is likely the most prevalent cause.[71] Our laboratory has recently determined that differences in landing tasks can significantly alter landing mechanics, including peak knee abduction angle.[219] Lastly, the differences between previous research and the current study could be due to task-based instruction. In the current study, subjects were not urged to make sure their feet were separated so that each foot was in the center of the adjacent force platforms. Although trials where each foot was not completely on the individual force platforms were discarded and recollected, instructions to separate feet during landing may unintentionally force the use of a wide stance during landing, which may result in a greater likelihood of increased knee abduction angles in subjects. The current study made every effort to not influence landing strategy through instructions.

5.2.6 Peak Knee Abduction Moment

The current study found that females landed with an average peak knee abduction moment ranging from 0.75 – 1.4 Nm/kg*m. These results were similar to the values reported by Hewett et al.[23] in their prospective risk factor analysis for ACL injuries.[23] Hewett et al.[23] reported average peak knee abduction moments of 15 Nm (0.41 Nm/kg*m) for the uninjured group and 35 Nm (0.97 Nm/kg*m) for the injured group.[23] At a jump distance of 20% body height subjects in the current study exhibited an average peak valgus moment that fell between the injured and uninjured group. The current study demonstrated an increase in average peak knee abduction moment to 1.4 Nm/kg*m as jump distance increased, higher than the average peak knee abduction moment reported for the injured group in the Hewett et al.[23] study. However, the higher value reported in the current study was expected because the current study used a jump distance up to 80% body height where Hewett et al.[23] used a jump distance of 50% body height during a drop-jump task.[23] Additionally, although the subjects who participated in the current study did not have a history of ACL injury, they were not assessed for injury risk.

5.2.7 Peak Proximal Anterior Tibial Shear Force

Results from the current study found an average peak PATSF ranging from 82.5 to 85.8% body weight. This range is consistent with previous studies reporting peak PATSF in recreationally active females.[33,71] Between different jump directions Sell et al.[33] reported a range of peak PATSF of 75 to 111% body weight.[33] Norcross et al.[71] reported an average peak PATSF of 87% body weight and Yu et al.[42] reported 79% body weight as an average peak PATSF.[42,71] Although each of these studies had variances between the task used, peak PATSF remained fairly similar in a comparable population of recreationally active females.

5.3 MUSCLE ACTIVATION DURING LANDING

Muscle activity of the vastus medialis, vastus lateralis, medial hamstring, and lateral hamstring was collected just prior to and directly after initial contact of the landing phase of the stop-jump maneuver. Muscle activity was collected during the 150ms time frame prior to landing (preactivity) and just after initial contact (reactivity).

5.3.1 Quadriceps Activation

The current study reported average preactivity of the vastus medialis muscle to be 12.2 to 21.3% MVIC*s. Average vastus medialis reactivity was found to be 55.5 to 60.4% MVIC*s. Average muscle preactivity of the vastus lateralis muscle was found to be 14.5 - 24.8% MVIC*s while average reactivity was 69.3 – 63.0% MVIC*s. Although preactivity values were similar, muscle activation values for reactivity were much higher than the values reported by Sell et al.[33] (5.7% to 13.9% MVIC*s).[33] The difference between these results is likely due to three reasons. Sell et al.[33] began the 150ms window at the time of peak posterior ground reaction force whereas the current study started the reactivity window the time of initial contact with the force platform.[33] Secondly, the current study utilized much greater jump distance that caused higher demand during landing, likely leading to increased muscle activation to accommodate increased ground reaction forces during landing. Lastly, the current study used a visual target to ensure that each trial the subject performed a high effort (90% max vertical jump) during the countermovement jump, possibly inducing higher activation levels. A study by Sigward et al.[159] reported quadriceps muscle activation during landing to be 150% MVIC.[159] However, this study utilized a single-leg cutting task, requiring much more muscle activation to control the initial loading.[159] During a double-leg drop jump Shultz et al.[64] reported the average quadriceps preactivity to be 18% and reactivity to be 90% MVIC.[64] These values were much more comparable to the current study, likely due to a similar landing demand, however, ground reaction force values were not provided to compare.

5.3.2 Hamstrings Activation

The current study reported average preactivity of the medial hamstring muscle as 7.7 to 12.8% MVIC*s. Average medial hamstring reactivity activity was 13.8 to 37.5% MVIC*s. Average preactivity of the lateral hamstring was found to be 13.8 - 22.8% MVIC*s while average reactivity was 63.9 – 66.2% MVIC*s. Again, similar differences between the current study results and previous literature can be seen. The current study reported much higher reactivity as compared to Sell et al.[33] who reported medial hamstring activation to the 6.8 to 8.1% MVIC*s.[33]

This difference is likely due to the same reasons described in the quadriceps muscle activation section above. Additionally, the study population used by Sell et al.[33] was younger (high-school basketball players).[33] It may be possible that with increased age and exposure to activity some adaptations may occur that affect muscle activation during landing. Shultz et al.[64] also reported similar results for the hamstrings at preactivity (18% MVIC) and reactivity (50% MVIC).[64] However, contrary to the finding in quadriceps activation, Sigward et al.[159] reported similar hamstring reactivity despite using a single-leg task (medial hamstrings = 40% MVIC and lateral hamstring = 55% MVIC).[159] This unexpected comparison may be due to the cutting task that was used by Sigward et al.[159] The task used in the current study may not have produced a stimulus that required increased hamstring activation similar to the increase in quadriceps activation.

5.4 SENSORIMOTOR CHARACTERISTICS

Sensorimotor characteristics were collected using three different techniques. Proprioception was collected for knee extension and flexion using threshold to detect passive motion. Neuromuscular performance was collected using two techniques, time to peak torque and peak torque. These tests were collected in both knee extension and flexion at a constant angular velocity of 240°/s.

5.4.1 Threshold To Detect Passive Motion

The current study found threshold to detect passive motion to be 2.48° toward knee extension and 1.75° toward knee flexion. A previous study by Lephart et al.[147] reported threshold to detect passive motion values in healthy active females to be 1.9° toward knee extension, similar but lower than the current findings.[147] The current study used a speed of 0.25°/s where Lephart et al.[147] used a speed of 0.5°/s, which may account for the slight difference in findings. A study by Nagai et al.[148] reported threshold to detect passive motion using the same speed as the current study but only in males.[148] They reported values of 1.7° toward extension and 1.4° toward flexion.[148] Previous literature has suggested gender differences in proprioception which may explain these differences.[183]

5.4.2 Time to Peak Torque

The current study found time to peak torque to be 121.89ms toward knee extension and 247.92ms toward knee flexion. The only other study that has used a similar measurement protocol for time to peak torque or peak torque at 240°/s was Clark.[181] In this study time to peak torque toward flexion was reported to be 231.9ms for females. This is very similar to what was found in the current study. However, time to peak toward extension was not reported. The current study also found a categorical distribution of time to peak torque towards flexion that required transforming the data into a categorical variable with three distinct groups. There was a wide range of time to peak torque values toward flexion and the sensitivity of the machine may not have been high enough. However, the transformed data still allowed comparisons to be made across jump distances and its relation to biomechanical risk factors for ACL injury.

5.4.3 Peak Torque

The current study found peak torque to be 109.75 %body weight toward knee extension and 86.86 %body weight toward knee flexion. To the author's best knowledge, this is the first study to report knee extension peak torque in healthy females at 240°/s. A study by Allison et al.[220] reported peak torque values at the knee to be 191.3 and 93.0 %body weight toward extension and flexion, respectively.[220] However, the study by Allison et al.[220] used a much slower dynamometer speed (60°/s)[220] where the current study used 240°/s. Although it cannot be determine if these two cohorts of healthy females would have performed similarly using the same dynamometer speed, the muscle strength force-velocity relationship theory does help explain the discrepancy in peak torque values seen between the two studies. Previous literature using the same dynamometer speed is not available to directly compare the values reported, however, these peak torque measures reported in the current study are consistent with previous measures of peak torque within our laboratory.

5.5 HYPOTHESIS TESTING AND FINDINGS

5.5.1 Effect of Jump Distance on Biomechanical Characteristics Related to ACL Injury

The first aim of this study was to determine if biomechanical risk factors and characteristics related to ACL injury change as jump distance increases from twenty to eighty percent of the subject's body height. Measurement of biomechanical characteristics during landing tasks is a widely used technique for assessing potential risk for ACL injury[23,168] and exploring other potential mechanistic relationships to the expressions of specific biomechanical characteristics.[31,70,216] Biomechanical evaluation can be a critical piece of injury prevention initiatives, specifically in the identification of risk factors[23,205] and assessing the effectiveness of an intervention at modifying these characteristics.[61,89,221] However, a critical piece of biomechanical assessments is the standardization of the task being analyzed. Our laboratory has demonstrated that simple variations in landing task parameters can significantly alter landing kinematics and kinetics.[219] t is not know if and/or how increased landing demand during the same task may change biomechanical characteristics related to ACL injury.

5.5.1.1 Hypothesis 1a: As jump distance increases the demand during landing will also increase as expressed by a significant increase in vertical and posterior ground reaction forces

It was first hypothesized that by increasing jump distance during the double-leg stop-jump an incremental increase the landing demand would be seen. For the purpose of this study an increase in landing demand was characterized by an observed increase in peak vertical and/or anterior-posterior ground reaction force. Previous research has demonstrated that increases in ground reaction forces correspond to increases in knee joint moments and powers during landing.[216]

Our hypothesis was supported by the results of the current study that showed increases in jump distances resulted in significant increases in both vertical and anterior-posterior ground reaction forces (**Table 5**). **Figure 7a and 7b** illustrate the incremental increase in peak ground reaction forces as jump distance increases from 20% to 80% body height. From 20% body height to 80% body height there was a 100% body weight increase in peak vertical ground reaction force and 42.1% increase in peak anterior-posterior ground reaction force. Sell et al.[91] reported similar a similar difference in peak anterior posterior ground reaction force from 20% to 80% body height jump distance (283.8N to 661.5N, $p < 0.001$).[91] Sell et al.[91] were also able to report a significant linear relationship between peak anterior-posterior ground reaction force and jump distance ($r = 0.682$, $p < 0.001$).[91] Although not an increase in jump distance, Dickin et al.[216] reported significant increases in peak vertical ground reaction forces with increasing drop height during a drop-jump maneuver from 75% body weight at 0.3m to 116% body weight at 0.5 meters.[216]

The incremental increase in ground reaction forces during landing demonstrates that the increase in jump distance during the double-leg stop-jump task used in the current study significantly increases landing demand for each 20% increase in jump distance. These findings support the current hypothesis and help to substantiate the following results because inferences on the overall aim of investigating the effect of increased landing demand on biomechanical characteristics would be impossible without first demonstrating that increased jump distance produced increased demand during landing.

5.5.1.2 Hypothesis 1b: As jump distance and landing demand increase there will also be a significant increase biomechanical characteristics related to ACL injury

The second hypothesis of the current aim was that as jump distance increased, biomechanical characteristics that have been shown to be related to ACL injury would also increase. An increase in biomechanical characteristics related to ACL injury is defined as a change in magnitude of the characteristic in a direction thought to be associated with increase risk of ACL injury. Biomechanical characteristics related to ACL injury include knee abduction angle at initial contact,[23] peak knee abduction angle,[23] knee abduction moment,[23] peak vertical ground reaction force,[23,71] peak anterior-posterior ground reaction force,[152] and peak proximal anterior tibial shear force.[33,152] This hypothesis was partially supported by the results of the current study.

Both peak vertical and peak anterior posterior ground reaction forces demonstrated significant increases with increased jump distance. Results from the current study showed that there was a significant effect of jump distance on peak knee abduction angle (**Table 5**). Post-hoc results for peak knee abduction angle revealed no significant change between 20% and 40% body height or between 60% and 80% jump distance, however, there was a significant increase from 40% to 60% body height. **Figure 7f** illustrates this increase in peak knee abduction angle used at higher jump distances. Although there was a significant increase from the smallest jump distance to the largest jump distance the actual difference in peak knee abduction angle was only 1.7° and on average, subjects maintained a fairly neutral knee position across all four distances (-0.28°, -0.02°, 0.8°, and 1.4°).

Knee abduction position at initial contact was not significantly affected by jump distance. Despite increase ground reaction forces subjects maintained the knee position at initial contact, not only in the frontal plane but also the sagittal plane. This suggests that as subjects were exposed to varying degrees of landing intensity their knee position at initial contact remained stable. In relation to the aims of this study, this finding may also suggest that comparison between knee positions at initial contact may be possible between studies that utilized different landing intensity. However, previous research has demonstrated that once the demand or goal of the task changes, lower extremity position is not consistent at initial contact.[219]

The most predictive characteristic for future ACL injury is peak knee abduction moment during landing.[23] During jump distances of 60% body height or greater subjects in the current study landed with greater peak knee abduction moment than those in the injured group of a prospective risk factor study for ACL injuries.[23] Despite the consistent knee adduction angle at initial contact, and a neutral peak knee abduction position the results of the current study demonstrated a significant effect of jump distance on peak knee abduction moment. Subjects in the current study went from a knee abduction moment that has been reported in healthy females to exceeding what a previous study demonstrated to be a risk factor for ACL injury, simply by increasing the intensity of the landing task.[23] A study by Myer et al.[222] found complimentary results that demonstrated the height of the subject's center of mass was predictive of peak knee abduction moment.[222] The higher the center of mass of the subject the higher the peak knee abduction moment experience during landing.[222] Although this study did not normalize jump distance or height, the relative demand did significantly contribute to the peak knee abduction moment experienced during landing.[222] This relates to the findings of the current study that found as jump distance increases so does the peak knee abduction moment during landing. Knee abduction moment, an identified risk factor of ACL injury, is significantly affected by landing intensity.

There was no significant effect of jump distance on peak proximal anterior tibial shear force in the current study. Both **Table 6** and **Figure 7h** illustrate how peak proximal anterior tibial shear force remained fairly consistent across jump distances. Although anterior-posterior ground reaction force increased with jump distance so did peak knee flexion. It is possible that subjects absorbed the ground reaction forces using increased knee flexion during landing, subjects and thus were able to maintain peak proximal anterior tibial shear force across jump distances. Previous studies have demonstrated the relationship between anterior-posterior ground reaction force and proximal anterior tibial shear force and the effect of increased knee flexion in the absorption of sagittal plane loading.[71,152]

Jump distance and subsequent landing intensity has a significant effect on biomechanics related to ACL injury except for knee abduction angle at initial contact and peak proximal anterior tibial shear force. Although the subjects in the current study did not exhibit risky kinematics during landing, the results clearly demonstrate that changes in landing intensity can significantly alter the magnitude of biomechanical characteristics related to ACL injury. Simply by changing the demand of the task used to evaluate biomechanical characteristics, subjects may be characterized differently as at risk or not at risk. These finding suggest that comparison between study results that utilize landing tasks of varying intensities or demand should be done with caution.

5.5.1.3 Hypothesis 1c: As jump distance and landing demand increase there will be significant changes in kinematic and kinetic measures related to knee joint loading

The third hypothesis of the current aim was that as jump distance increases there would be a significant increase in knee kinematics and kinetics related to knee loading. For the purposes of this study, an increase in knee loading was characterized by an increase in peak knee flexion angle and at initial contact, and peak proximal anterior tibial shear force. This hypothesis was partially supported by the results of the current study.

Results showed a significant effect of jump distance on peak knee flexion angle but not knee flexion angle at initial contact (**Table 5**). Post-hoc analysis showed a significant within subject increase of peak knee flexion between each jump distance increase. However, knee flexion at initial contact remained consistent across each of the jump distances. Although knee flexion excursion was not directly measured the increase in peak knee flexion angle with consistent knee flexion angle at initial contact corresponds to an increase in overall knee flexion excursion during landing as jump distance increases. Previous studies have demonstrated that increases in peak knee flexion with consistent knee flexion at initial contact result in increased loading and energy absorption during landing.[71,216]

As stated in the previous hypothesis discussion, increases in jump distances did not result in increases in peak proximal anterior tibial shear force. Peak proximal anterior tibial shear force remained similar across all jump distances. This estimated measure of directional knee force is thought to correspond to the forces applied in the direction of action of the ACL.[41] Perhaps subjects utilized the observed increase in knee flexion to absorb the increase in round reaction forces across jump distances, maintaining consistent proximal anterior tibial shear force. Previous studies have demonstrated that alterations in sagittal plain kinematics are associated with changes in proximal anterior tibial shear force.[71,152]

Increases in jump distance have a significant effect on peak knee flexion angle and possibly knee flexion excursion but not on other knee loading characteristics such as peak proximal anterior tibial shear force. It is possible that this specific population (healthy adult female population) utilizes increased knee flexion to accommodate increased landing intensity. Another aspect of the task that may be important is the post-landing goal of a vertical jump to reach and touch a target. This countermovement task immediately following the landing may have contributed to the use of increased knee flexion during landing as a performance modulator. It is unknown if by removing the countermovement aspect of the task and asking subjects to land if these results would remain the same.

5.5.1.4 Hypothesis 1d: As jump distance and landing demand increase muscle activation of the quadriceps and hamstrings pre-landing and post-landing activity will also increase

The fourth hypothesis of the current aim was that as jump distance increased there would also be an increase in quadriceps and hamstrings activation, both during preactivity and reactivity windows. This hypothesis was partially supported by the results of the current study. Jump distance had a significant effect on all muscle activation characteristics, both preactivity and reactivity.

Preactivity of both the quadriceps and hamstrings show a fairly linear increase with increasing jump distance (**Figure 8a-d**). Post-hoc analyses showed that both the quadriceps and hamstring muscles did not significantly increase from 20% body height to 40% body-height but each subsequent increase in jump distance after that did yield significant increases in preactivity of the quadriceps and hamstrings. Preactivity had increases from 20% to 80% body height of 8.1%, 6.1%, 4.3%, and 8.7% MVIC for the vastus lateralis, vastus medialis, medial hamstring, and lateral hamstring, respectively ($p < 0.001$). This systematic increase in both quadriceps and hamstring muscle activation may be the result of an increased preparatory co-activation strategy prior to landing. In the current study subjects were required to take at least one practice jump at each jump distance but could take as many as they would like. This familiarization with each jump distance, along with the task being planned in nature, may have been enough for subjects to develop a neuromuscular preparatory response for the task. Chappell et al.[31] investigated the preparatory response to landing and found that females had significantly greater quadriceps and hamstrings preactivity compared to males.[31] Although the current study only assessed this in females it appears that there may be a gender trend in preparatory muscle activation but it is unknown if males would have a similar preactivity response to jump distance as females.

Overall there was again a significant effect of jump distance on quadriceps and hamstrings muscle activation reactivity. However, the results do not support the same linear relationship that was seen in the preactivity (**Figure 8e-h**). Interestingly, the jump distance of 40% body height elicited the greatest amount of muscle activation for the vastus medialis, medial hamstring, and lateral hamstring. A jump distance of 80% was greatest in the vastus lateralis, however, this change was not statistically significant. Post-hoc analyses showed that the only significant differences in reactivity of all muscles were between jump distances of 20% and 60% body height. Vastus medialis and medial hamstring activation between jump distances of 20% and 80% body height were also significantly different. These results suggest that landings from shorter jump distances elicit significantly less muscle activation during landing than the longer distances. However, an unexpected finding was that the central tendency of the vastus medialis, medial hamstring, and lateral hamstring was the decrease in activation levels between jump distances of 60% and 80% body height. This may suggest that the subjects were no longer adjusting their neuromuscular response to landing above the demand of 60% body height jump distance.

Increases in jump distance do have a significant effect on muscle activation strategies. This effect seems to be more linear in preactivity, however, reactivity measures were much more variable. Jump distances of 60% body height elicit the greatest increase in neuromuscular demand during landing. Regardless, comparison between studies that utilize tasks of different landing intensity or demand is not suggested.

5.5.2 Relationship between the sensorimotor system characteristics and landing biomechanics

The second aim of this study was to determine if sensorimotor characteristics (proprioception, time to peak torque, and peak torque) can significantly predict the expression of biomechanical characteristics related to ACL injury and if this relationship changes with jump distance. The sensorimotor system is a critical component of performance and injury prevention because of its responsibility for the detection of joint perturbations and the execution of appropriate motor responses to execute tasks.[53,95,96] A previous study has suggested that there may be a relationship between landing biomechanical characteristics and measures of the sensorimotor system, however, this study utilized only men.[148] Additionally, it is unknown how landing intensity affects this relationship. If the components of the sensorimotor system do have a significant influence on the landing biomechanics, then significant relationships should be observed across all jump distances.

5.5.2.1 Hypothesis 2a: Sensorimotor characteristics at the knee will each independently significantly contribute to the variance of biomechanical risk factors for ACL injury.

The first hypothesis of the current aim and prerequisite for the second aim was that sensorimotor characteristics would have simple linear relationships and would independently explain the variability associated with biomechanical characteristics related to ACL injury. The results of the current study partially support this hypothesis.

Sensorimotor characteristics had no significant correlation with either peak vertical ground reaction force or peak anterior-posterior ground reaction force (**Table 11A-B**). Additionally, there were no significant relationship between sensorimotor characteristics and ground reaction force measures using simple linear regression (**Table 12A-B**). Sensorimotor characteristics do not independently explain the variance of peak vertical or anterior-posterior ground reaction forces during landing at any of the four jump distances.

Knee flexion angle at initial contact did show significant linear relationships with some sensorimotor characteristics across different jump distances. Threshold to detect passive motion and time to peak torque toward flexion both were significantly related to knee flexion at initial contact at jump distances of 20% to 60% body height. However, at 80% jump height, threshold to detect passive motion toward extension and flexion and peak torque toward extension and flexion were significantly related to knee flexion at initial contact. Simple linear regression showed similar results. Threshold to detect passive motion toward flexion significantly explained 10.17% and 7.04% of the variance of knee flexion at initial contact at jump distances of 20% and 60% body height, respectively. Time to peak torque toward flexion also significantly explained 15.43% and 12.62% of the variance of knee flexion at initial contact at jump distances of 20% and 60% body height, respectively. No sensorimotor characteristics were able to significantly explain the variance of knee flexion at initial contact at a jump distance of 40% body height. At a jump distance of 80% body height, threshold to detect passive motion toward extension and flexion and time to peak torque toward extension and flexion significantly explained 17.49%, 15.78%, 10.50%, and 10.13%, respectively. The highest independent relationship with knee flexion at initial contact occurred at a jump distance of 80% body height. Nagai et al.[148] also found that better threshold to detect passive motion and higher peak torque accounted for a significant amount of the variance in knee flexion at initial contact.[148]

Results also showed significant relationships between sensorimotor characteristics and peak knee flexion during landing. The only significant linear relationships with peak knee flexion were with peak torque toward extension at a jump distance of 40% body height and threshold to detect passive motion toward extension at 80% body height. These characteristics accounted for only 8.44% and 7.81% of the variance of peak knee flexion. Similarly, Nagai et al.[148] found that at a jump distance of 40% peak torque toward extension significantly accounted for 7.8% of the variance of knee flexion excursion. Although this study did not measure knee flexion excursion, the fact that this study demonstrated no significant change in knee flexion at initial contact, peak knee flexion would be related to total knee flexion excursion.[148] Additionally Shultz et al.[64] demonstrated that peak torque values were not independently related to knee flexion excursion during drop landings.[64] This discrepancy with the current study and Nagai et al.[148] may be due to the task used as both the current study and Nagai et al.[148] used a stop-jump task at 40% body height.[148] Overall, sensorimotor characteristics do not seem to have a strong relationship with peak knee flexion used during landing, however the relationship is significant.

Sensorimotor characteristics showed no significant linear relationship with frontal plane kinematics (knee abduction at initial contact and peak knee abduction) through either correlation or simple linear regression. To the author's best knowledge this is the first study to investigate the relationship between sensorimotor characteristics and frontal plane kinematics. The observed lack of independent linear relationship may be due to the evaluation of sensorimotor characteristics specific to sagittal plane knee motion and strength. As new research continues to establish a potential relationship between frontal plane knee motion with hip characteristics[223,224] it may be useful to evaluate this relationship using sensorimotor characteristic at the hip.

Peak knee abduction moment was not normally distributed so a transformation was done to achieve normally distributed residuals. In the case of peak knee abduction moment the transformation that was successful was using the square root. However, despite normally distributed residuals, sensorimotor characteristics were not correlated to the original or the transformed dependent variables. Simple regression analyses resulted in the same findings. Sensorimotor characteristics did not account for a significant amount of the variance in the original or transformed peak knee abduction moment. Similar to peak knee abduction and at initial contact, the lack of relationship between sensorimotor characteristics may be partially due to the sagittal plane biased sensorimotor characteristics at the knee.

The current study failed to demonstrate any significant linear relationship between sensorimotor characteristics and peak proximal anterior tibial shear force using correlations. Results also showed that individual sensorimotor characteristics were not able to account for a significant amount of peak knee abduction moment, original or transformed. Although proximal anterior tibial shear force is specific to the sagittal plane, similar to the sensorimotor characteristics, previous research has demonstrated that posterior ground reaction force is the best predictor of proximal anterior tibial shear force.[152] The current study did not assess the correlation between these two variables, however sensorimotor characteristics did not significantly predict posterior ground reaction forces.

Although a few biomechanical characteristics were not related to the individual characteristics, knee flexion at initial contact and peak knee flexion had a few significant relationships with individual sensorimotor characteristics. Further analyses using multiple linear regression allow the evaluation of a linear relationship between biomechanical characteristics and multiple sensorimotor characteristics.

5.5.2.2 Hypothesis 2b: Sensorimotor characteristics at the knee will together significantly contribute to the variance of biomechanical risk factors for ACL injury.

The second hypothesis of the current aim was that sensorimotor characteristics would together explain a significant amount of variability associated with biomechanical characteristics related to ACL injury. The results of the current study partially support this hypothesis as some dependent variables showed no significant relationship with sensorimotor characteristics while others did.

Multiple linear regression models for both peak vertical ground reaction force and peak anterior-posterior ground reaction force yielded no significant models. Considering these finding with those of the previous aim, the results suggest that sensorimotor characteristics are not related to peak ground reaction force measured during double–leg stop-jumps.

Multiple linear regression models did significantly predict knee flexion at initial contact at all four jump distances. Although models from each jump distance did not contain the same final predictor variables there was a noticeable pattern that sensorimotor characteristics toward flexion were significant predictors at jump distances of 20%, 40%, and 60% body height. The most common prediction among these jump distance was time to peak torque toward flexion. Model results suggest that subjects with increased time to peak torque values utilize greater knee flexion at initial contact. At a jump distance of 20% body height, threshold to detect passive motion was also a significant predictor suggesting that subjects with more acute proprioception (less positional error) also used greater knee flexion at initial contact. The only other study examining the relationship between landing biomechanics and sensorimotor characteristics also found that threshold to detect passive motion toward flexion, with peak flexion torque, significantly accounted for 27.4% of the variance of knee flexion at initial contact.[148] Although the current study did not find peak torque toward flexion to be a significant predictor from jump distances of 20% - 60% body height, Nagai et al.[148] used a single-leg stop-jump task and male subjects.[148] However, contrary to findings at closer jump distances, results from the current study demonstrated sensorimotor characteristics directed toward extension significantly accounted for the variance of knee flexion at initial contact at a jump distance of 80% body height ($R^2 = 0.2672$, MSE = 4.70). This model at 80% body height was the best model fit for knee flexion at initial contact based on R^2 and MSE values. This model demonstrated that more acute proprioception and greater peak toque toward extension resulted in greater knee flexion at initial contact. This is an interesting finding that may suggest that the challenge and focus of the sensorimotor system may shift during tasks of different difficulties.

Multiple linear regression models for peak knee flexion resulted in significant findings for models of 40% and 80% jump distance. Sensorimotor characteristics did not account for a significant amount of variance at jump distance of 20% and 60% body height. At a jump distance of 40% body height peak torque toward extension was able to account for 8.44% of the variance of peak knee flexion. At a jump distance of 80% body height, threshold to detect passive motion toward extension was able to account for 7.81% of the variance of peak knee flexion. Even when considering the MSE values (8.71 vs. 8.66 respectively) both models have a similar goodness of fit. These results suggested that sensorimotor characteristics directed toward extension are most related to peak knee flexion during landing. The study by Nagai et al.[148] also reported peak torque toward extension to be the only significant predictor of knee flexion excursion during landing.[148] The findings of the current study add that there may be a shift in sensorimotor characteristics demand and/or influence as landing demand increases.

Multiple linear regression models were not able to significantly predict peak knee abduction angle or knee abduction angle at initial contact. Additionally, no significant relationships were identified with sensorimotor characteristic using simple regression or correlation. These results suggest that sensorimotor characteristics do not have a linear relationship with frontal plane knee motion at initial contact or peak values. As stated previously, this lack of relationship could be due to the sagittal plane bias sensorimotor measures used in this study. However, a previous study by Wild et al.[225] reported that adolescent females with lower hamstring strength (< 45 Nm) displayed significantly more peak knee abduction angles.[225] This prepubescent female sample was between the ages of 10 and 13 years of age where the current study used healthy adult females with a mean age of 23.2 and a ranged between 18 to 31 years of age. It may be that prepubescent females are at greater risk of utilizing these dangerous kinematic characteristics unlike the sample from the current study who demonstrated a mean peak knee abduction angle of 1.4° at a jump distance of 80% body height. Additionally, subjects in the current study demonstrated mean hamstring strength (peak torque toward flexion) of 86.86

%body weight where the average value reported in the study by Wild et al.[225] for "lower hamstring strength" was 1.17 %body weight, higher than the mean hamstring strength reported in the current study.[225] However, the study by Wild et al.[225] used a dynamometer speed of 180°/s, much slower than the speed used in the current study.

Sensorimotor characteristics were able to account for a significant amount of the variance in peak knee abduction moment during all four jump distances. Time to peak torque toward flexion was a significant predictor in the final models at all four jump distances. This finding is consistent with findings reported by Myer et al.[222] who demonstrated that peak hamstring torque (peak torque toward flexion) was a significant predictor of knee abduction moment along with knee valgus motion, center of mass height and percent body fat.[222] The study by Myer et al.[222] used an isokinetic dynamometer speed similar to what was used in this current study (300°/s vs 240°/ sec). At jump distances of 40% and 60% body height, threshold to detect passive motion, both toward extension and flexion, were also significant predictors of peak knee abduction moment. Based on the results of the current study it appears that at very close or very far distances only peak torque toward flexion significantly accounts for the variance of peak knee abduction moment. However, at mid distances both peak torque toward flexion and proprioceptive measure were predictive of peak knee abduction moment. It may be that landing intensities that are too low do not stimulate the neuromuscular control system enough to require significant input from all systems but intensities that are too great may overload the system and/or require greater contribution by strength components of the neuromuscular control system. This suggests that evaluation of the neuromuscular control system as it relates to its effect on peak knee abduction moment, a prospective risk factor for ACL injury, should be performed at landing intensities of mid range (between 40% and 60% body height for a double-leg stop-jump task). It is still unknown if this relationship or application would hold true for other biomechanical evaluation tasks.

Lastly, results from the current study demonstrated that sensorimotor characteristics did not account for a significant amount of variance seen in peak proximal anterior tibial shear force. A previous study by Sell et al.[152] performed an analysis to determine predictors of proximal anterior tibial shear force.[152] That study did not include any of the same variables that were included in this particular analysis of the current study but they were able to determine that peak posterior ground reaction force, knee flexion moment, knee flexion angle, vastus lateralis activation, and sex significantly accounted for 86.09% of the variance of peak proximal anterior tibial shear force.[152] These findings along with those of the current study suggest that proximal anterior tibial shear force is best predicted by factors that are concurrent to the specific landing mechanics (other kinematic and kinetic measures) rather than measures of the current state of neuromuscular control and muscle performance (proprioception, time to peak torque, and peak torque).

5.6 STUDY LIMITATIONS

The current study has some limitations that are worth noting. First, the current study used a healthy adult female population and therefore the results cannot be extrapolated to an adolescent or prepubescent female population who may not have undergone the same maturation process at the currently used young adult population. We also cannot assume that male subjects would respond the same way to changes in landing intensity as females. Many studies have identified biomechanical differences and ACL injury risk factors between genders.[56,94,159,173,195,226,227]

Additionally, the population sampled in the current study was healthy and free from ACL injury. Athletes reported a variety of activity levels relative to the knee and they were not evaluated for risk of ACL injury, therefore we cannot make conclusions related to the specific effect landing intensity has on injury risk classification, nor was this the purpose of the current study. Similarly, the current study did not specifically recruit individuals with a deficit in sensorimotor system characteristics.

The chosen task itself also introduces some limitations on the generalizability of the findings from the current study. First, previous research has demonstrated that different landing tasks elicit different biomechanical characteristics relative to the specific demand of the task.[33,75,219,228] Therefore, we cannot assume the same changes in biomechanical characteristics would occur if another task were used. However, the task used in the current study was specifically chosen because of its use in previous ACL injury risk factor studies[31,42,152,194] and because it has been demonstrated to elicit greater knee abduction angles and moments compared to single-leg tasks.[31,42,152,194] Another limitation relative to the jump landing task was that it was a planned task. Previous studies have identified that the addition of a reaction component to a landing task significantly affects landing biomechanics.[33,94,160] It is possible that if the subjects in the current study were exposed to a reaction component such as jump direction it may have modulated the effect of landing demand or may have altered the contributions of the sensorimotor system on landing biomechanics. Additionally, the curent study normalized the jump distance by body height because this practice is a very common method to normalize task demand between subjects.[23,33,42,71] However, this method does not account for individual abilities. As seen in the current study, two subjects were not able to perform a distance of 80% of their body height. Therefore, the landing intensity between subjects may not be perfectly consistent. This is not a large concern in the current study because the current study utilized within subject analysis. However, studies using statistical analyses of the central statistic need to consider that adjusting the jump landing demand based on body height may result in different relative landing demands based on individual ability.

Three measures of sensorimotor characteristics were used to investigate the relationship between the sensorimotor system and landing biomechanics associated with ACL injury. These included threshold to detect passive motion, time to peak torque, and peak torque. These measures are all specific to sagittal plane of motion, assessing a person's ability to detect and perform only in the sagittal plane. These measures were chosen because of their use in previous research that has identified the effect of ACL injury on the sensorimotor system.[117,119,144,148,180,181] Some recent research suggests that the hip may also be a contributor to biomechanical measures associated with ACL injury[223,224] and may warrant further investigation into sensorimotor characteristics at the hip to help explain frontal plane kinematics that were not explained by the sensorimotor characteristic in the current study.

Lastly, the current study used a cross-sectional study design comparing biomechanical characteristics across different jump distances and between biomechanical and sensorimotor characteristics at a similar point in time. Results from this study cannot explain how time or fatigue may affect these relationships. The results of the relationship are relevant to each individual's current physical state. It is still unknown how changes in a person's physical state, such as fatigue, illness, or concurrent injury may affect these relationships. Previous research has demonstrated that physical state, such as fatigue, does significantly affect lower extremity landing biomechanics and sensorimotor characteristics.[216,229-232]

5.7 FUTURE RESEARCH

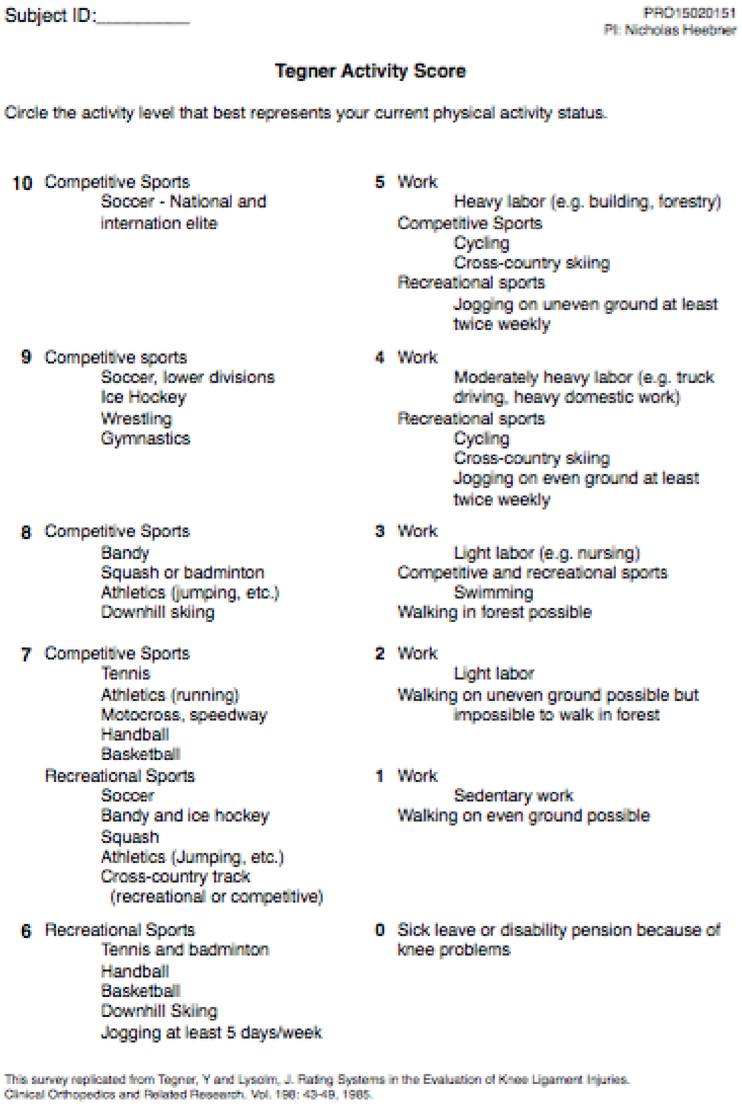
Based on the finding of the current study there are a few considerations that future studies should explore. First, future studies need to determine standardized methods for assessing biomechanical risk factors for ACL injury during landing tasks. Additionally, these studies should seek to determine what is the optimal method to standardize jump landing tasks among individuals with varying landing ability, such as using a percentage of a person's maximum broad jump. Future studies should also look to investigate if and how individual changes in biomechanics and a person's adaptability across different landing demands may influence injury risk. Individuals with a more adaptive sensorimotor system to control risk factors for injury, such as peak knee abduction moment, across different demands may be at a reduced risk of injury. Lastly, the current study demonstrates a lack of relationship between the sensorimotor system characteristics that were measured and knee abduction angle, both at initial contact and peak angles. Future studies should investigate how sensorimotor characteristics at the hip, such as hip internal rotation, might better influence frontal plane landing biomechanics related to ACL injury.

5.8 CONCLUSION

The purpose of this study was to determine the effects of jump distance on biomechanical risk factors for ACL injury and to examine the relationship between sensorimotor system characteristics and landing biomechanics throughout different jump distances. The results of this study demonstrated that increases in jump distance resulted in significant increases in landing demand and significant changes in landing biomechanics relevant to ACL injury. These findings first illustrate that studies utilizing tasks with different demands cannot compare landing biomechanics or make inference to injury risk based on findings from studies that have utilized different task demands. Additionally, this suggests that researchers and clinicians need to determine a standardized approach to measure landing biomechanics for the assessment of injury risk that ensures similar demand within and between subjects. Lastly, the results of the current study illustrate that as landing demand (jump distance) changes the contributions of the sensorimotor system also change. Based on these findings it is suggested that double-leg stop-jump tasks be performed at a jump distance of 40% - 60% body height. This range of distance did elicit biomechanical risk factors for ACL injury and sensorimotor characteristics were shown to have a significant contribution to these risk factors. Researchers and clinicians that attempt to use landing biomechanics to assess the integrity or function of the sensorimotor system as it relates to ACL injury risk or prevention need to be aware that sensorimotor system characteristics are most related to knee flexion at initial contact, peak knee flexion, and peak knee abduction moment and have the strongest relationship at mid-range landing demand. The contribution of the sensorimotor system is less related to these biomechanical characteristics at very short and very long jump distances.

APPENDIX A

TEGNER ACTIVITY SCORE QUESTIONNAIRE



Subject ID:________

PRO15020151
PI: Nicholas Heebner

Tegner Activity Score

Circle the activity level that best represents your current physical activity status.

- **10** Competitive Sports
    - Soccer - National and internation elite
- **9** Competitive sports
    - Soccer, lower divisions
    - Ice Hockey
    - Wrestling
    - Gymnastics
- **8** Competitive Sports
    - Bandy
    - Squash or badminton
    - Athletics (jumping, etc.)
    - Downhill skiing
- **7** Competitive Sports
    - Tennis
    - Athletics (running)
    - Motocross, speedway
    - Handball
    - Basketball

  Recreational Sports
    - Soccer
    - Bandy and ice hockey
    - Squash
    - Athletics (Jumping, etc.)
    - Cross-country track (recreational or competitive)
- **6** Recreational Sports
    - Tennis and badminton
    - Handball
    - Basketball
    - Downhill Skiing
    - Jogging at least 5 days/week
- **5** Work
    - Heavy labor (e.g. building, forestry)

  Competitive Sports
    - Cycling
    - Cross-country skiing

  Recreational sports
    - Jogging on uneven ground at least twice weekly
- **4** Work
    - Moderately heavy labor (e.g. truck driving, heavy domestic work)

  Recreational sports
    - Cycling
    - Cross-country skiing
    - Jogging on even ground at least twice weekly
- **3** Work
    - Light labor (e.g. nursing)

  Competitive and recreational sports
    - Swimming

  Walking in forest possible
- **2** Work
    - Light labor

  Walking on uneven ground possible but impossible to walk in forest
- **1** Work
    - Sedentary work

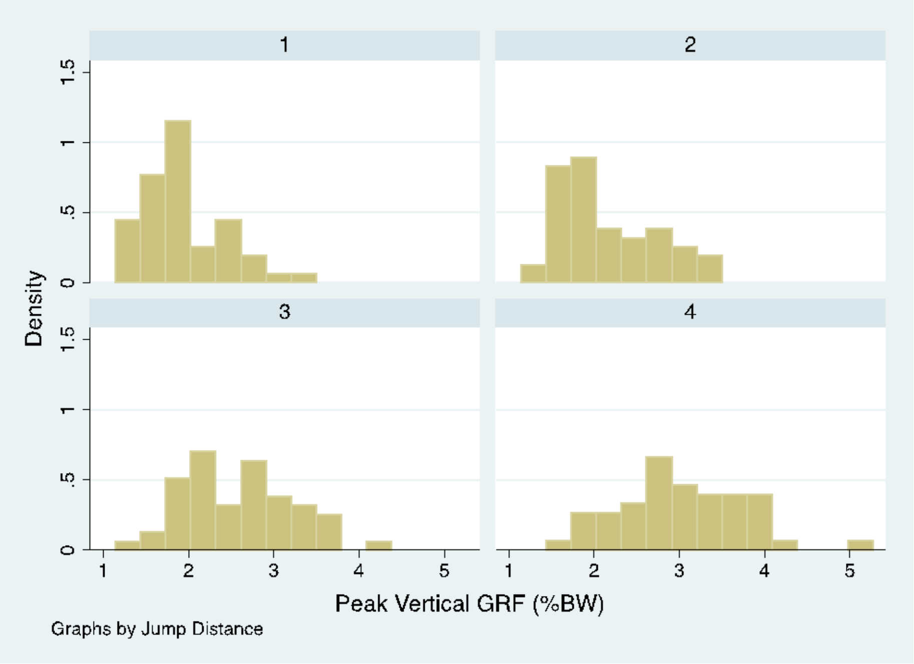
  Walking on even ground possible
- **0** Sick leave or disability pension because of knee problems

This survey replicated from Tegner, Y and Lysolm, J. Rating Systems in the Evaluation of Knee Ligament Injuries. Clinical Orthopedics and Related Research. Vol. 198: 43-49, 1985.

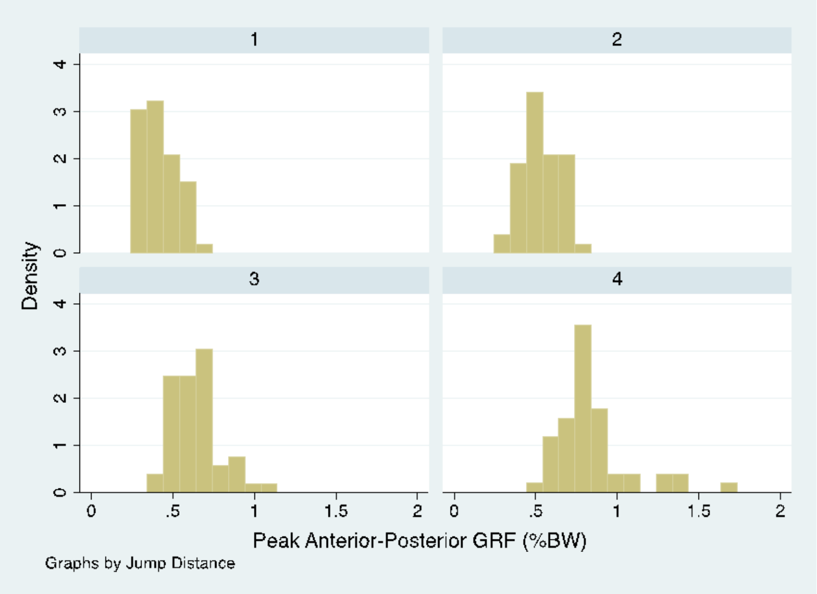
APPENDIX B

HISTOGRAMS OF BIOMECHANICAL RISK FACTORS OF ACL INJURY

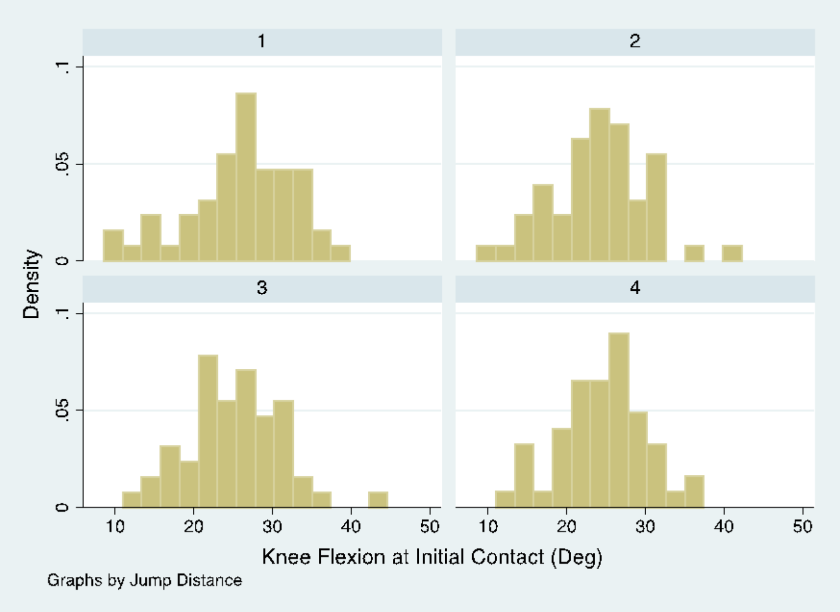
B.1 PEAK VERTICAL GROUND REACTION FORCE



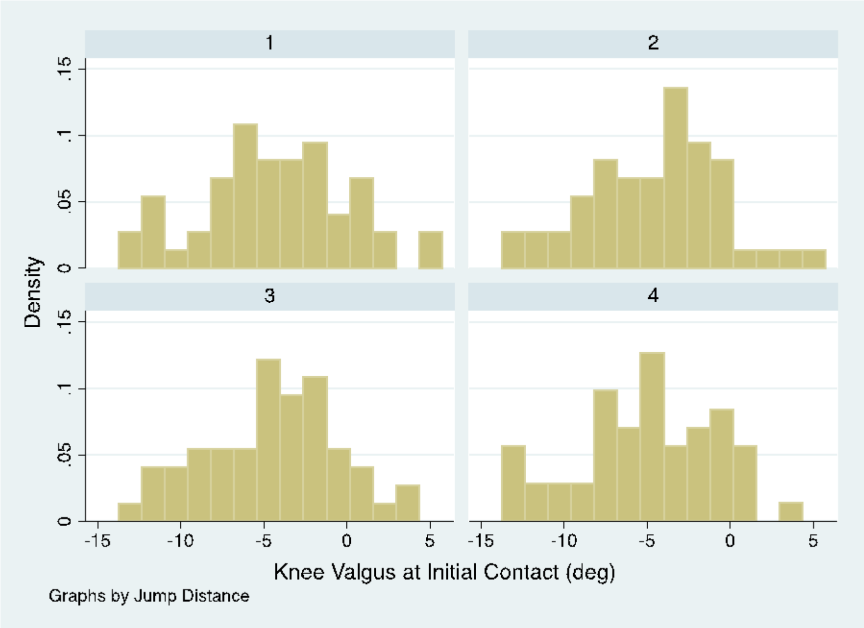
B.2 PEAK ANTERIOR-POSTERIOR GROUND REACTION FORCE



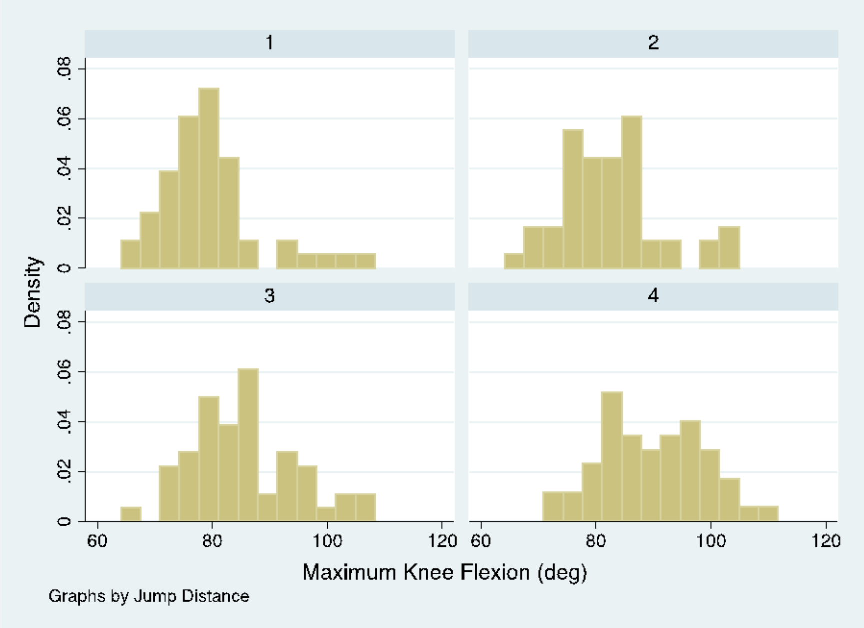
B.3 KNEE FLEXION AT INITIAL CONTACT



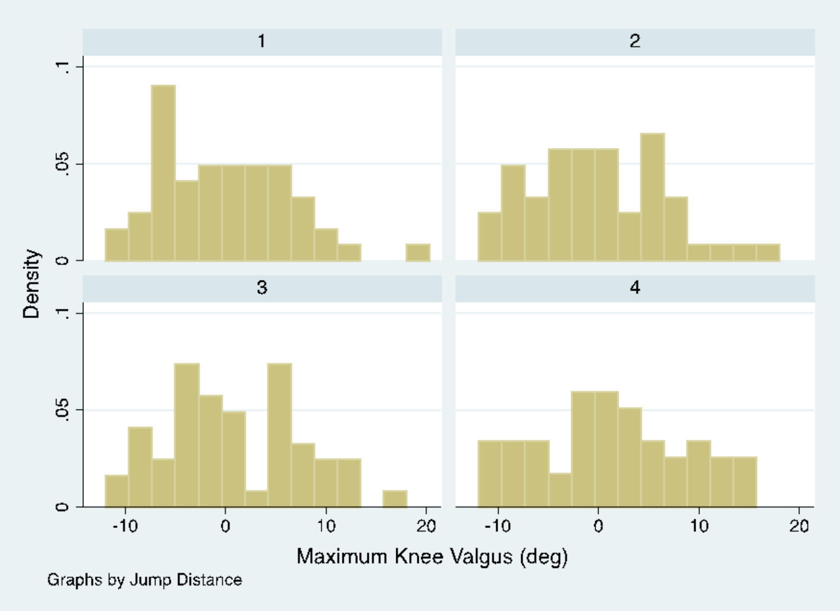
B.4 KNEE ABDUCTION AT INITIAL CONTACT



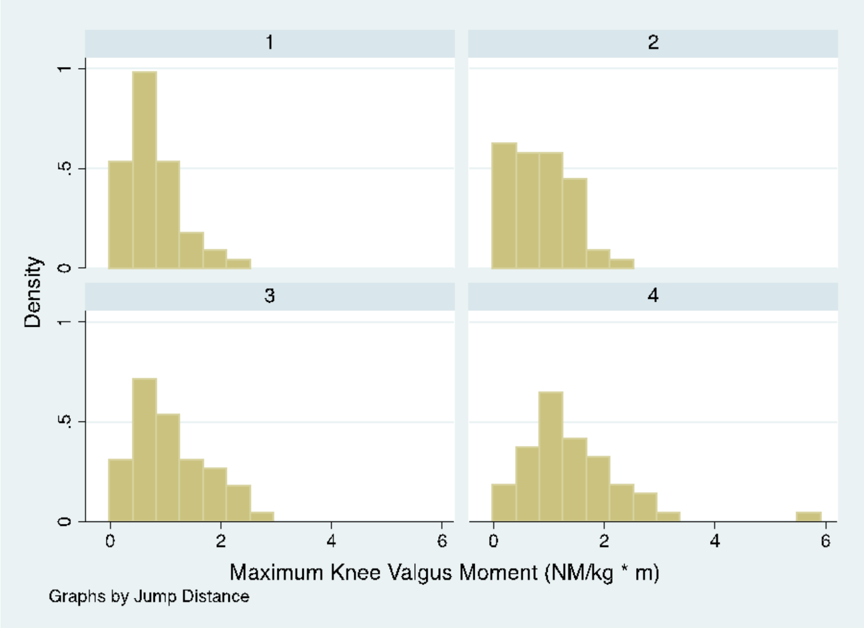
B.5 PEAK KNEE FLEXION



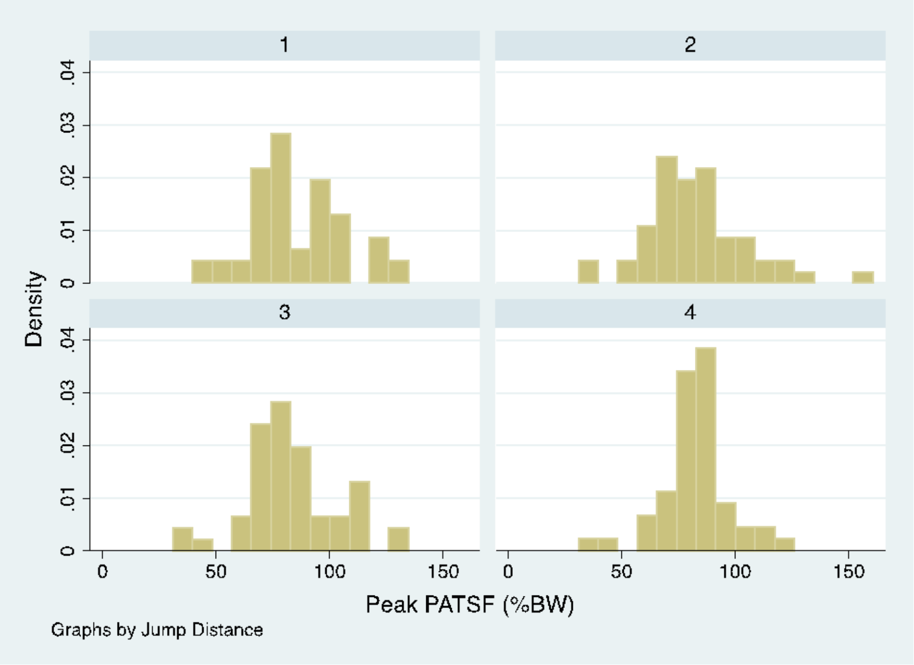
B.6 PEAK KNEE ABDUCTION



B.7 PEAK KNEE ABDUCTION MOMENT



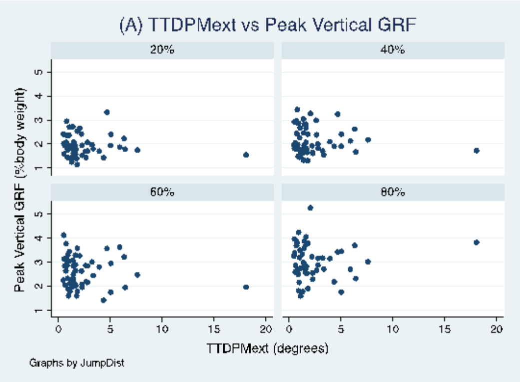
B.8 PEAK PROXIMAL ANTERIOR TIBIAL SHEAR FORCE

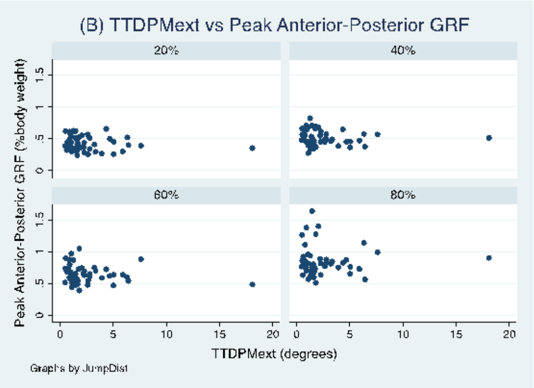


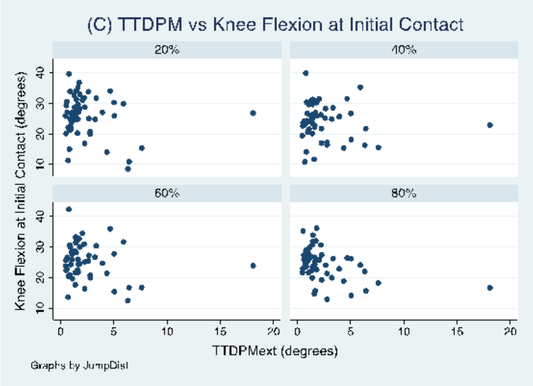
APPENDIX C

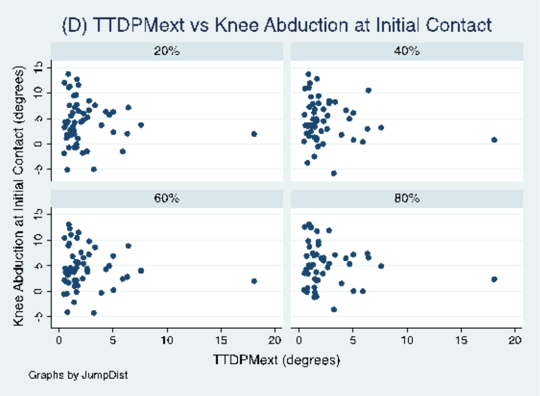
BIVARIATE ANALYSIS: PAIRWISE SCATTER PLOTS

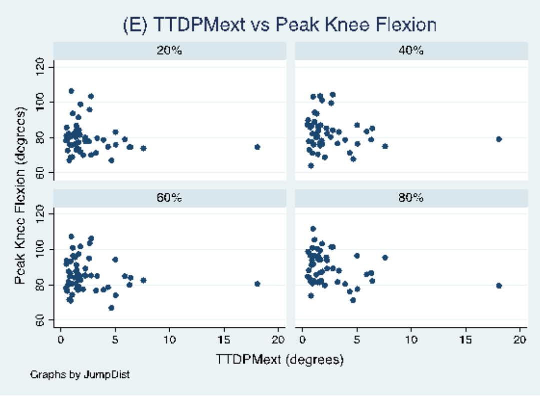
C.1 SCATTER PLOTS OF TTDPMEXT VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

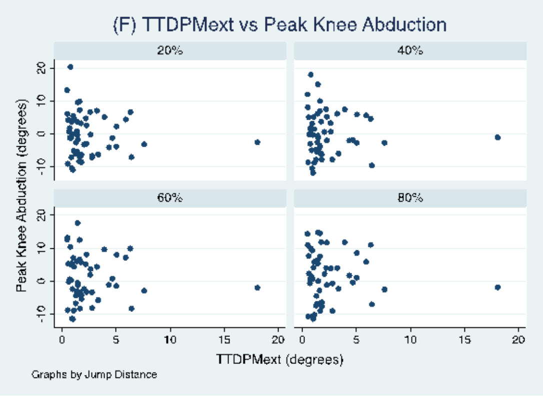


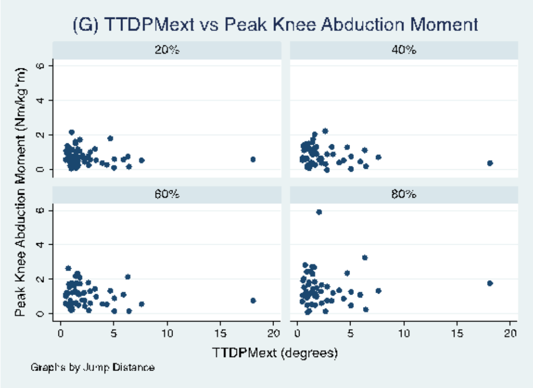


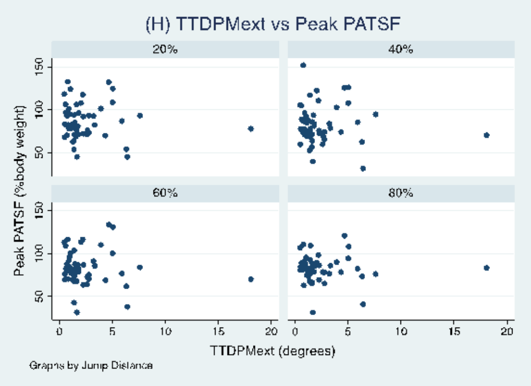

(C) TTDPM vs Knee Flexion at Initial Contact
20%
40%
60%
80%
Knee Flexion at Initial Contact (degrees)
10
20
30
40
0
5
10
15
20
TTDPMext (degrees)
Graphs by JumpDist


(D) TTDPMext vs Knee Abduction at Initial Contact
20%
40%
60%
80%
Knee Abduction at Initial Contact (degrees)
-5
0
5
10
15
0
5
10
15
20
TTDPMext (degrees)
Graphs by JumpDist

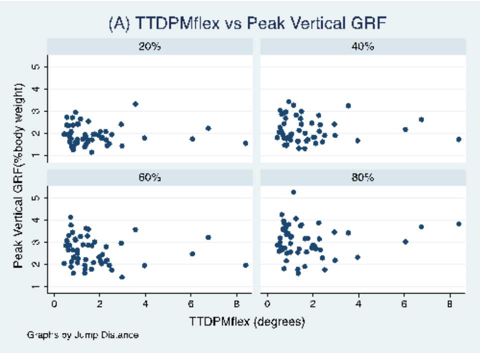

(E) TTDPMext vs Peak Knee Flexion
20%
40%
60%
80%
Peak Knee Flexion (degrees)
60
80
100
120
0
5
10
15
20
TTDPMext (degrees)
Graphs by JumpDist

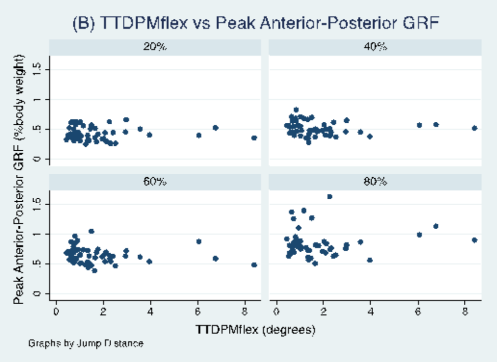

(F) TTDPMext vs Peak Knee Abduction
20%
40%
60%
80%
Peak Knee Abduction (degrees)
-10
0
10
20
0
5
10
15
20
TTDPMext (degrees)
Graphs by Jump Distance

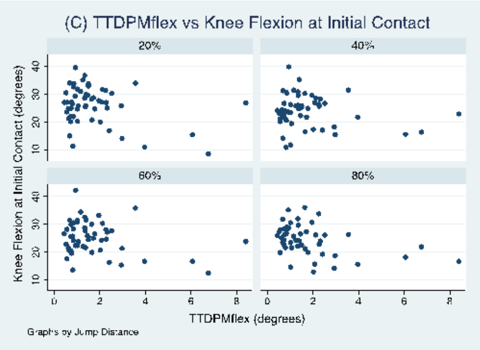

(G) TTDPMext vs Peak Knee Abduction Moment
20%
40%
60%
80%
Peak Knee Abduction Moment (Nm/kg*m)
0
2
4
6
0
5
10
15
20
TTDPMext (degrees)
Graphs by Jump Distance

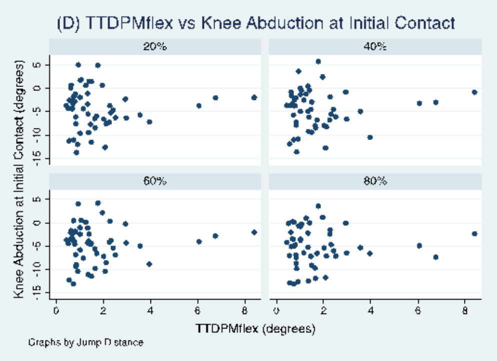

(H) TTDPMext vs Peak PATSF
20%
40%
60%
80%
Peak PATSF (%body weight)
50
100
150
0
5
10
15
20
TTDPMext (degrees)
Graphs by Jump Distance

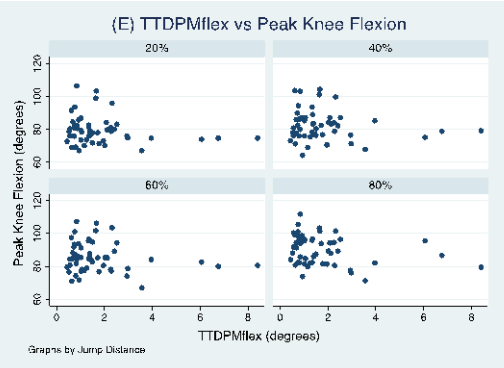
C.2 SCATTER PLOTS OF TTDPMFLEX VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

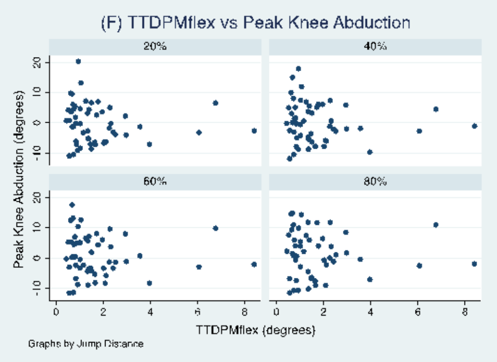


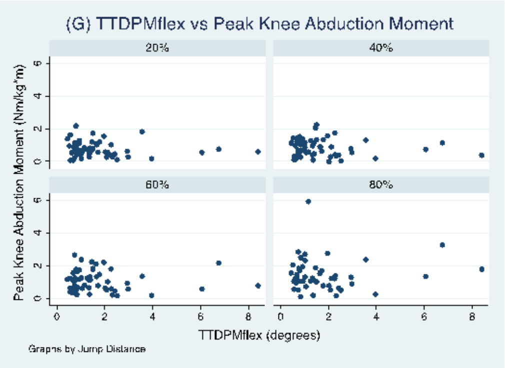


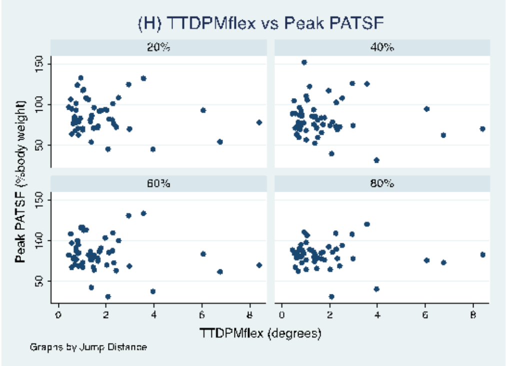




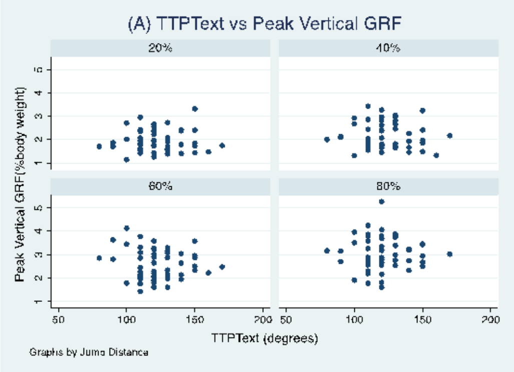


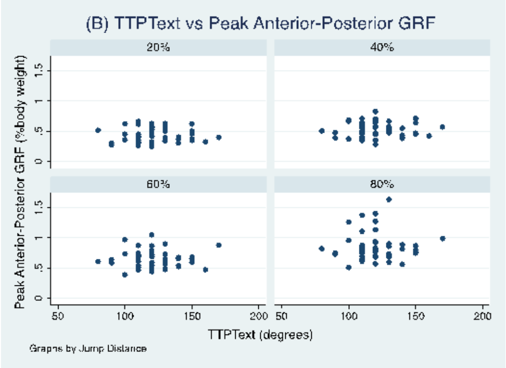


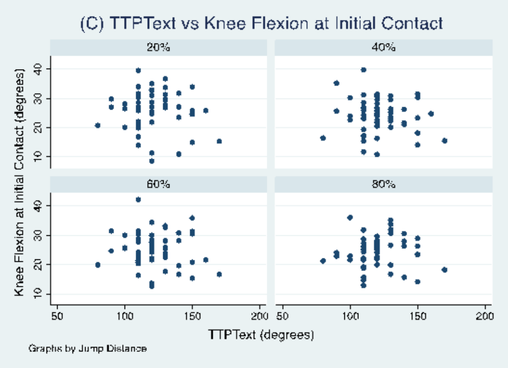


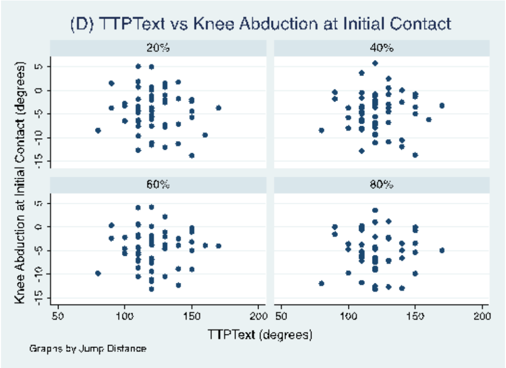


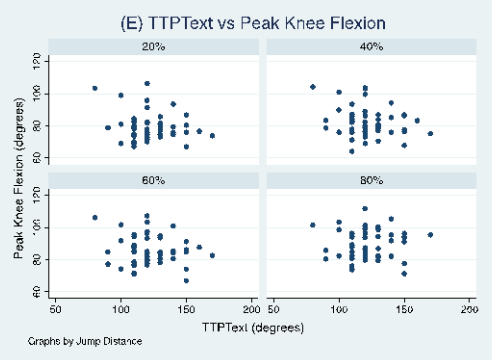
C.3 SCATTER PLOTS OF TTPTEXT VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

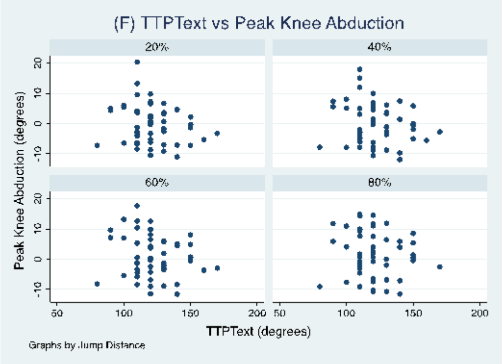


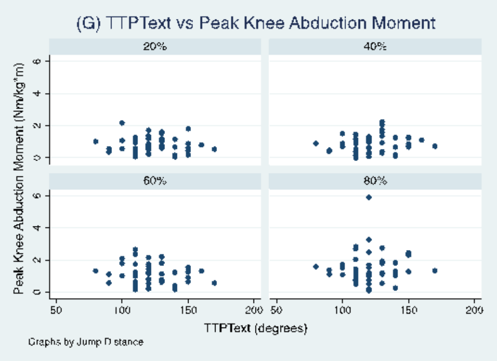


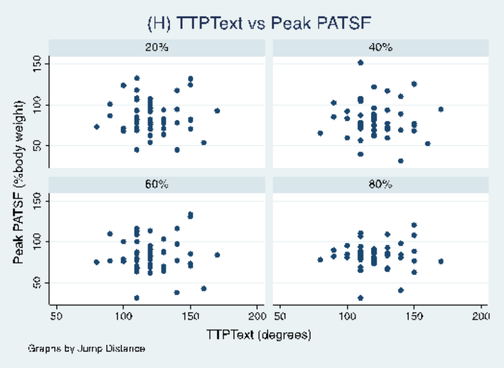




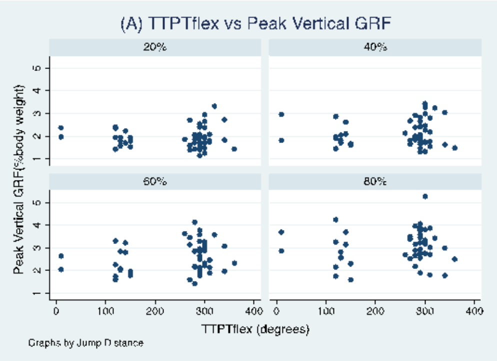


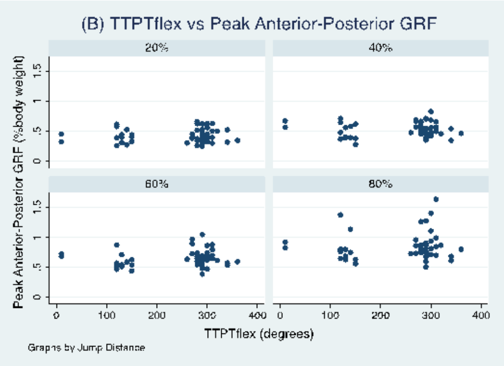


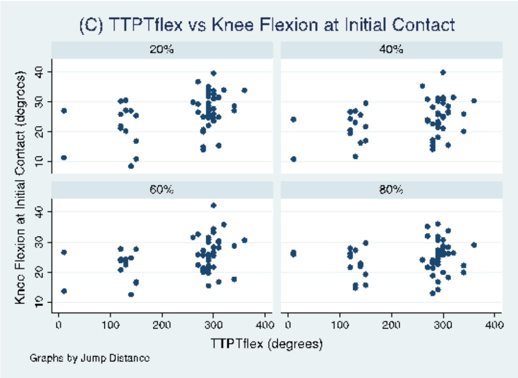


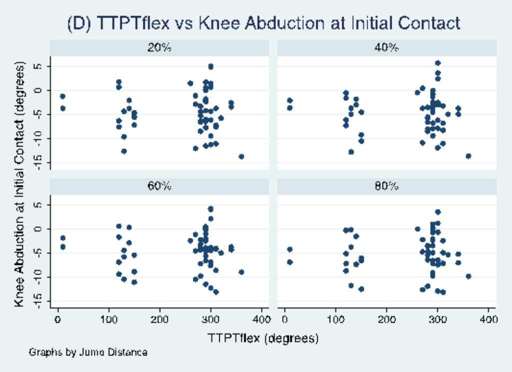


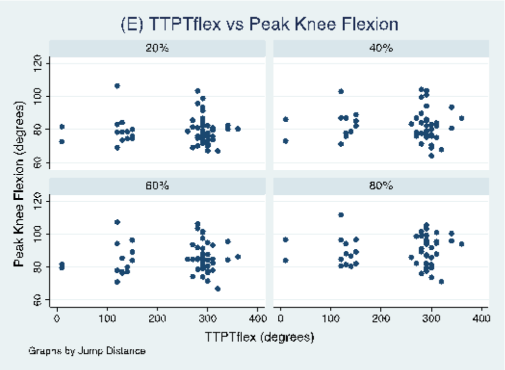
C.4 SCATTER PLOTS OF TTPTFLEX VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

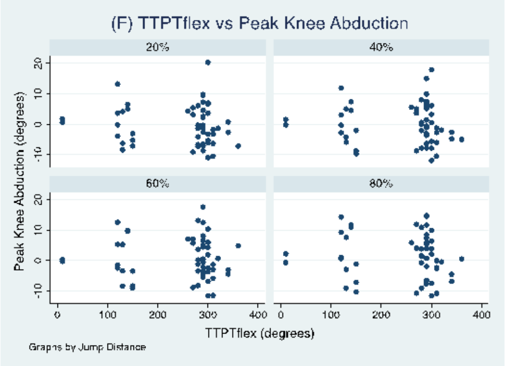


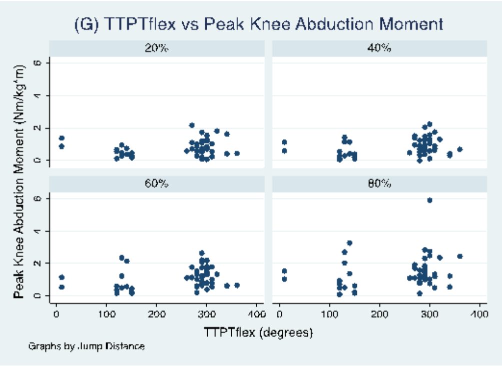


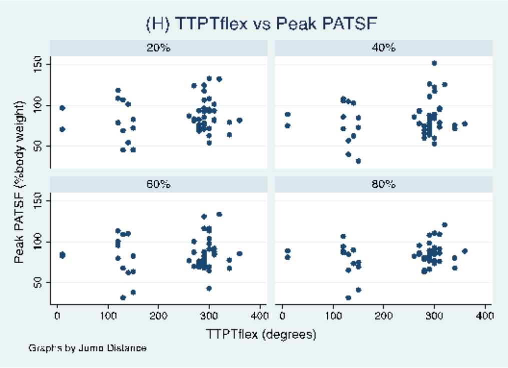




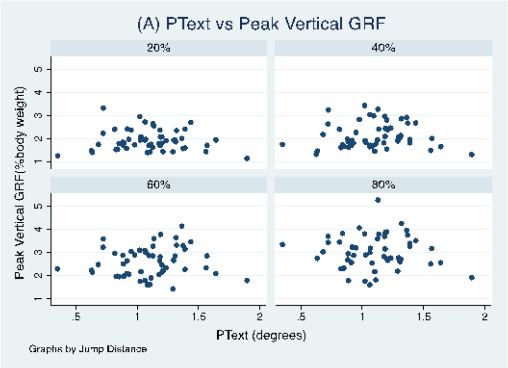


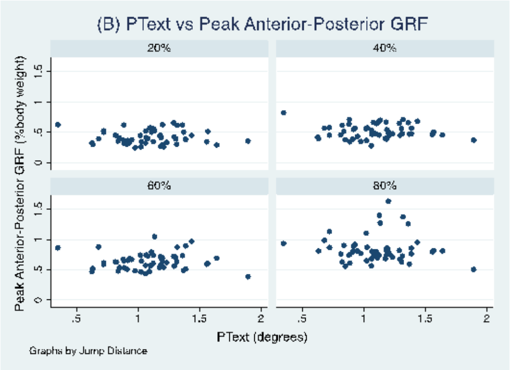


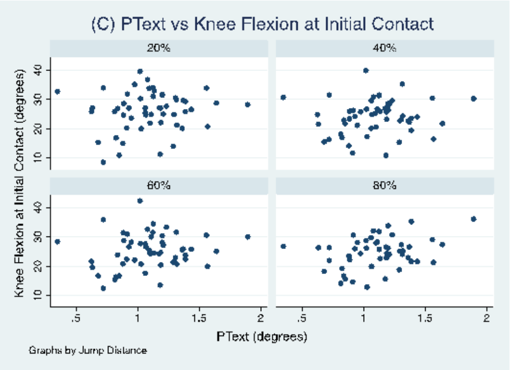


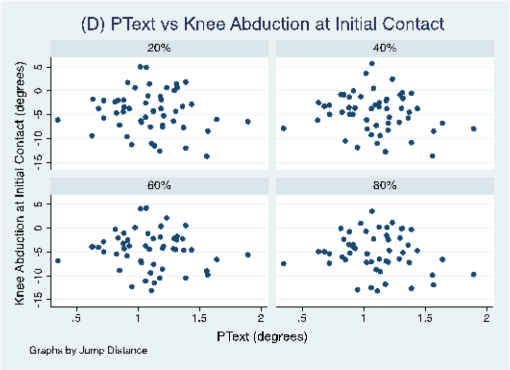


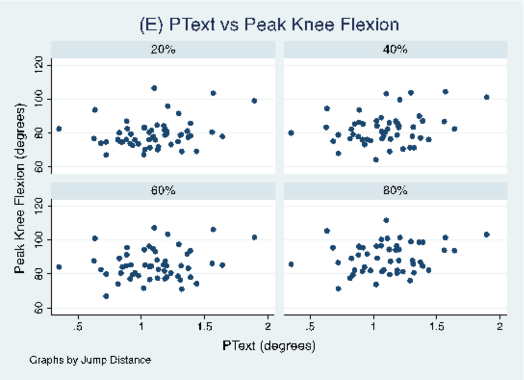
C.5 SCATTER PLOTS OF PTEXT VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

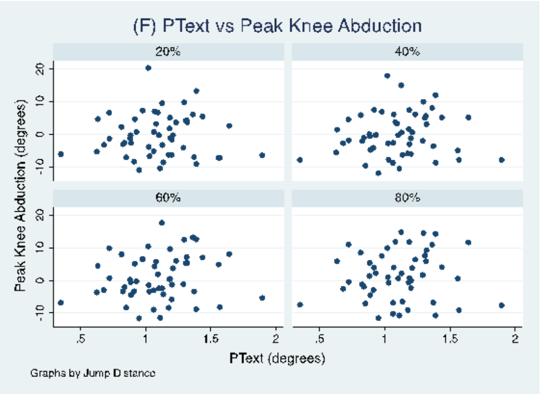

(A) PText vs Peak Vertical GRF

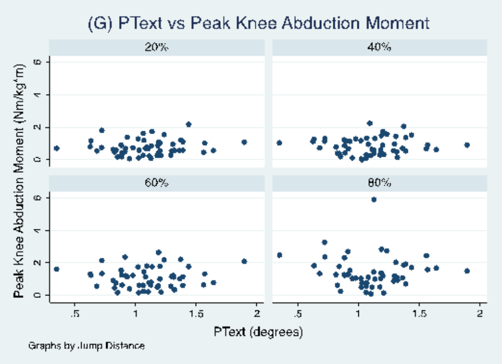

(B) PText vs Peak Anterior-Posterior GRF

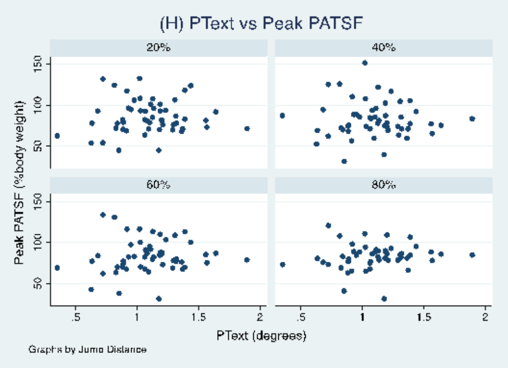

(C) PText vs Knee Flexion at Initial Contact


(D) PText vs Knee Abduction at Initial Contact

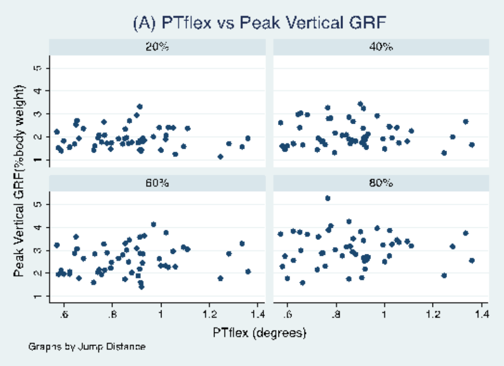

(E) PText vs Peak Knee Flexion

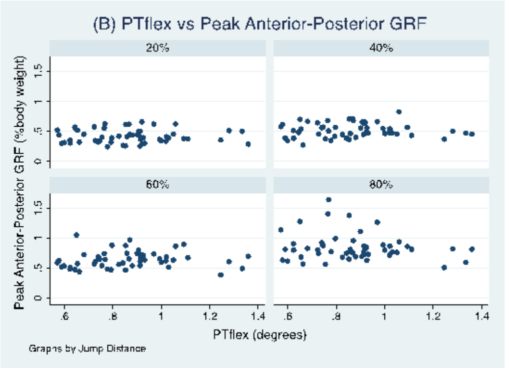

(F) PText vs Peak Knee Abduction

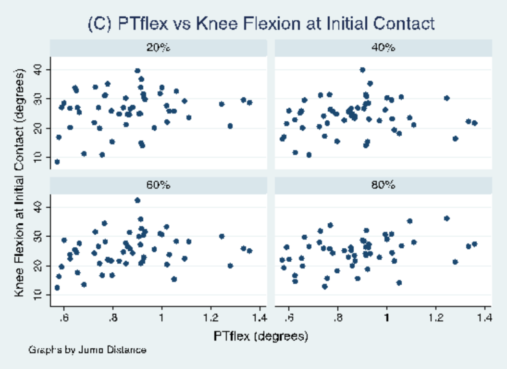


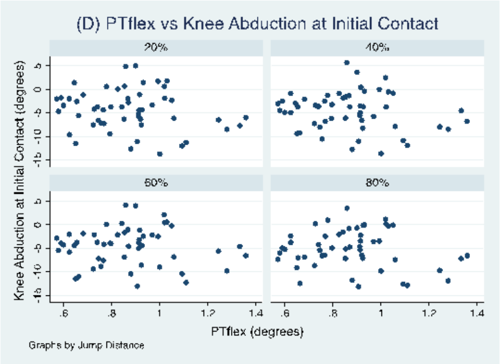


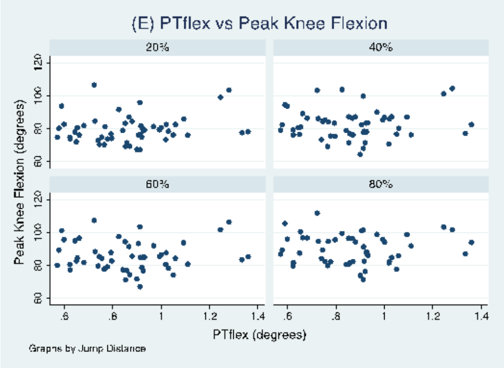
C.6 SCATTER PLOTS OF PTFLEX VERSUS DEPENDENT VARIABLES FOR EACH JUMP DISTANCE

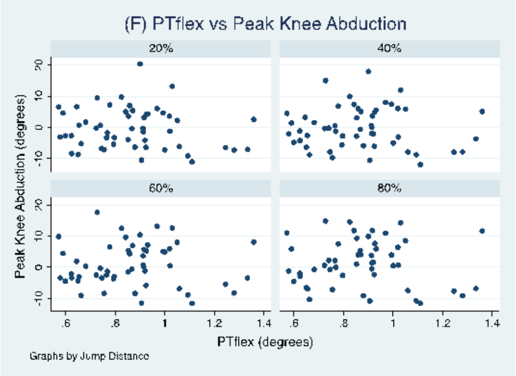


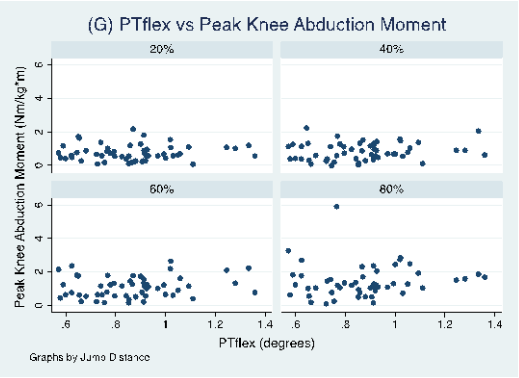


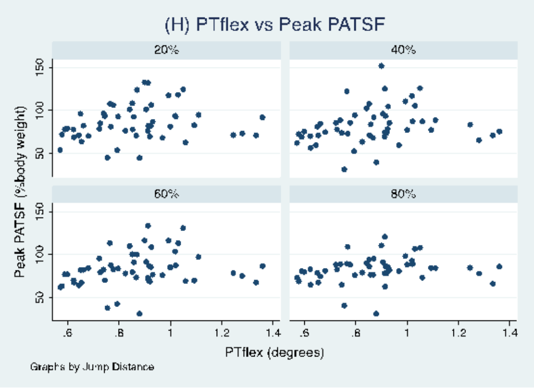





(E) PTflex vs Peak Knee Flexion


(F) PTflex vs Peak Knee Abduction


(G) PTflex vs Peak Knee Abduction Moment


(H) PTflex vs Peak PATSF

APPENDIX D

BIVARIATE ANALYSIS: PAIRWISE CORRELATION RESULTS

D.1 PEAK VERTICAL GRF VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.074	0.597	-0.075	0.594	-0.112	0.425	0.056	0.698
TTDPMflex	-0.047	0.740	-0.018	0.897	-0.149	0.288	0.056	0.699
TTPText	0.036	0.801	-0.030	0.830	-0.089	0.526	0.027	0.851
TTPTflex	-0.006	0.964	0.089	0.528	0.216	0.120	0.125	0.383
PText	-0.049	0.726	0.031	0.828	0.100	0.475	-0.053	0.714
PTflex	-0.073	0.605	-0.049	0.728	0.150	0.284	0.017	0.904

D.2 PEAK ANTERIOR-POSTERIOR GRF VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.110	0.433	-0.126	0.367	-0.170	0.223	-0.007	0.960
TTDPMflex	-0.025	0.858	-0.114	0.418	-0.143	0.306	0.081	0.572
TTPText	0.020	0.889	0.004	0.979	0.000	0.953	0.011	0.941
TTPTflex	0.078	0.578	-0.005	0.973	0.107	0.446	0.092	0.520
PText	-0.024	0.867	-0.087	0.534	0.008	0.957	-0.056	0.694
PTflex	0.003	0.984	-0.001	0.995	0.090	0.522	-0.159	0.264

D.3 KNEE FLEXION AT INITIAL CONTACT VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.172	0.218	-0.086	0.540	-0.166	0.234	-0.418	0.002*
TTDPMflex	-0.319	0.020*	-0.184	0.187	-0.273	0.048*	-0.397	0.004*
TTPText	-0.025	0.857	-0.114	0.418	-0.079	0.575	0.261	0.856
TTPTflex	0.400	0.003*	0.328	0.017*	0.358	0.009*	0.141	0.323
PText	0.124	0.377	0.111	0.429	0.153	0.273	0.324	0.020*
PTflex	0.195	0.163	0.145	0.302	0.246	0.076	0.318	0.023*

D.4 KNEE ABDUCTION AT INITIAL CONTACT VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	0.105	0.453	0.143	0.306	0.087	0.536	0.118	0.409
TTDPMflex	0.025	0.860	0.063	0.655	0.061	0.663	0.025	0.862
TTPText	-0.095	0.499	-0.009	0.949	-0.008	0.955	0.003	0.984
TTPTflex	-0.027	0.848	-0.007	0.962	0.022	0.875	0.064	0.655
PText	-0.108	0.442	-0.159	0.255	-0.085	0.544	-0.139	0.332
PTflex	-0.162	0.248	-0.150	0.283	-0.051	0.716	-0.121	0.400

D.5 PEAK KNEE FLEXION VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.161	0.250	-0.125	0.375	-0.149	0.288	-0.279	0.047*
TTDPMflex	-0.187	0.181	-0.153	0.276	-0.174	0.212	-0.256	0.070
TTPText	-0.196	0.159	-0.222	0.109	-0.129	0.359	0.005	0.970
TTPTflex	-0.016	0.911	-0.026	0.853	0.087	0.537	0.019	0.897
PText	0.268	0.052	0.291	0.035*	0.181	0.196	0.207	0.144
PTflex	0.188	0.177	0.080	0.571	0.035	0.801	0.011	0.938

D.6 PEAK KNEE ABDUCTION VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.077	0.585	-0.043	0.759	-0.042	0.764	-0.029	0.839
TTDPMflex	-0.095	0.501	-0.084	0.548	-0.060	0.667	-0.038	0.792
TTPText	-0.165	0.238	-0.166	0.236	-0.142	0.312	-0.107	0.454
TTPTflex	-0.081	0.564	-0.059	0.675	-0.012	0.933	-0.089	0.534
PText	0.038	0.786	0.100	0.477	0.161	0.250	0.077	0.593
PTflex	-0.122	0.384	-0.049	0.730	-0.007	0.963	-0.052	0.718

D.7 PEAK KNEE ABDUCTION MOMENT VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.156	0.264	-0.203	0.145	-0.150	0.283	0.018	0.903
TTDPMflex	-0.119	0.397	-0.116	0.410	-0.075	0.593	0.060	0.678
TTPText	0.017	0.905	0.165	0.239	-0.028	0.845	0.103	0.473
TTPTflex	0.165	0.237	0.265	0.056	0.261	0.059	0.195	0.170
PText	0.093	0.506	0.031	0.827	0.041	0.770	-0.059	0.682
PTflex	0.053	0.709	0.149	0.287	0.186	0.183	0.081	0.572

D.8 PEAK PATSF VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.055	0.698	-0.061	0.662	-0.084	0.551	-0.059	0.682
TTDPMflex	-0.126	0.370	-0.091	0.519	-0.130	0.354	-0.087	0.543
TTPText	-0.015	0.918	0.016	0.912	-0.003	0.984	0.042	0.772
TTPTflex	0.122	0.385	0.117	0.404	0.081	0.564	0.193	0.175
PText	0.058	0.679	-0.029	0.837	0.093	0.508	0.104	0.470
PTflex	0.180	0.196	0.211	0.130	0.224	0.106	0.183	0.199

D.9 PEAK KNEE ABDUCTION MOMENT (SQUARE ROOT) VS. SENSORIMOTOR CHARACTERISTICS BY JUMP DISTANCE

	20% Body Height		40% Body Height		60% Body Height		80% Body Height	
	r	p-value	r	p-value	r	p-value	r	p-value
TTDPMext	-0.142	0.311	-0.192	0.168	-0.150	0.283	0.026	0.858
TTDPMflex	-0.100	0.479	-0.107	0.445	-0.094	0.505	0.055	0.701
TTPText	0.013	0.929	0.166	0.236	-0.011	0.940	0.105	0.464
TTPTflex	0.153	0.273	0.282	0.041	0.316	0.021	0.245	0.083
PText	0.096	0.496	0.032	0.823	0.037	0.795	-0.039	0.784
PTflex	0.059	0.677	0.167	0.231	0.195	0.162	0.151	0.291

Abbreviations: vertical ground reaction force (vGRF), anterior-posterior ground reaction force (apGRF), body weight (BW), initial contact (IC), abduction (ABD), abduction moment (ABDmom), proximal anterior tibial shear force (PATSF), threshold to detect passive motion toward extension (TTDPMext), threshold to detect passive motion toward flexion (TTDPMflex), time to peak torque toward extension (TTPText), time to peak torque toward flexion (TTPTflex), peak torque toward extension (PText), peak torque toward flexion (PT flex), Pearson's rho (r)

* $p < 0.05$

APPENDIX E

SIMPLE LINEAR REGRESSION: MODEL RESULTS

E.1 PEAK VERTICAL GRF MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0055	0.28	(1 , 51)	0.4541	0.5967
TTDPMflex	53	0.0022	0.11	(1 , 51)	0.4549	0.7398
TTPText	53	0.0013	0.06	(1 , 51)	0.4551	0.8010
TTPTflex	53	0.0000	0.00	(1 , 51)	0.4553	0.9641
TTPTflex[a]	53	0.0131	0.33	(2 , 50)	0.4569	0.7188
PText	53	0.0024	0.12	(1 , 51)	0.4548	0.7262
PTflex	53	0.0053	0.27	(1 , 51)	0.4542	0.6053
40% Body Height						
TTDPMext	53	0.0056	0.29	(1 , 51)	0.5557	0.5938
TTDPMflex	53	0.0003	0.02	(1 , 51)	0.5571	0.8974
TTPText	53	0.0009	0.05	(1 , 51)	0.5569	0.8301
TTPTflex	53	0.0078	0.40	(1 , 51)	0.5550	0.5282
TTPTflex[a]	53	0.0382	0.99	(2 , 50)	0.5518	0.3779
PText	53	0.0009	0.05	(1 , 51)	0.5569	0.8278
PTflex	53	0.0024	0.12	(1 , 51)	0.5565	0.7282
60% Body Height	53					
TTDPMext	53	0.0125	0.65	(1 , 51)	0.6390	0.4245
TTDPMflex	53	0.0221	1.15	(1 , 51)	0.6359	0.2879
TTPText	53	0.0079	0.41	(1 , 51)	0.6405	0.5263
TTPTflex	53	0.0467	2.50	(1 , 51)	0.6279	0.1200
TTPTflex[a]	53	0.0687	1.85	(2 , 50)	0.6267	0.1676
PText	53	0.0101	0.52	(1 , 51)	0.6398	0.4748
PTflex	53	0.0225	1.17	(1 , 51)	0.6358	0.2840
80% Body Height						
TTDPMext	50	0.0031	0.15	(1 , 49)	0.7395	0.6980
TTDPMflex	50	0.0031	0.15	(1 , 49)	0.7395	0.6988
TTPText	50	0.0007	0.04	(1 , 49)	0.7404	0.8509
TTPTflex	50	0.0155	0.77	(1 , 49)	0.7349	0.3834
TTPTflex[a]	50	0.0657	1.69	(2 , 48)	0.7233	0.1960
PText	50	0.0028	0.14	(1 , 49)	0.7396	0.7135
PTflex	50	0.0003	0.01	(1 , 49)	0.7405	0.9041

[a] Grouped variable (three groups)

E.2 PEAK ANTERIOR-POSTERIOR GRF MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0121	0.62	(1 , 51)	0.1128	0.4332
TTDPMflex	53	0.0006	0.03	(1 , 51)	0.1134	0.8579
TTPText	53	0.0004	0.02	(1 , 51)	0.1134	0.8887
TTPTflex	53	0.0061	0.31	(1 , 51)	0.1131	0.5783
TTPTflex[a]	53	0.0069	0.17	(2 , 50)	0.1142	0.8408
PText	53	0.0006	0.03	(1 , 51)	0.1134	0.8669
PTflex	53	0.0000	0.00	(1 , 51)	0.1135	0.9844
40% Body Height						
TTDPMext	53	0.0160	0.83	(1 , 51)	0.1177	0.3670
TTDPMflex	53	0.0129	0.67	(1 , 51)	0.1178	0.4177
TTPText	53	0.0000	0.00	(1 , 51)	0.1186	0.9788
TTPTflex	53	0.0000	0.00	(1 , 51)	0.1186	0.9733
TTPTflex[a]	53	0.0548	1.45	(2 , 50)	0.1165	0.2445
PText	53	0.0076	0.39	(1 , 51)	0.1182	0.5343
PTflex	53	0.0000	0.00	(1 , 51)	0.1186	0.9945
60% Body Height						
TTDPMext	53	0.0290	1.52	(1 , 51)	0.1368	0.2229
TTDPMflex	53	0.0205	1.07	(1 , 51)	0.1374	0.3062
TTPText	53	0.0001	0.00	(1 , 51)	0.1388	0.9530
TTPTflex	53	0.0114	0.59	(1 , 51)	0.1380	0.4461
TTPTflex[a]	53	0.0665	1.78	(2 , 50)	0.1355	0.1789
PText	53	0.0001	0.00	(1 , 51)	0.1388	0.9574
PTflex	53	0.0081	0.42	(1 , 51)	0.1383	0.5217
80% Body Height						
TTDPMext	50	0.0001	0.00	(1 , 49)	0.2269	0.9603
TTDPMflex	50	0.0066	0.32	(1 , 49)	0.2261	0.5720
TTPText	50	0.0001	0.01	(1 , 49)	0.2269	0.9412
TTPTflex	50	0.0085	0.42	(1 , 49)	0.2259	0.5200
TTPTflex[a]	50	0.0244	0.60	(2 , 48)	0.2264	0.5524
PText	50	0.0032	0.16	(1 , 49)	0.2265	0.6942
PTflex	50	0.0254	1.28	(1 , 49)	0.2240	0.2639

[a] Grouped variable (three groups)

E.3 KNEE FLEXION AT INITIAL CONTACT MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0296	1.56	(1 , 51)	6.6896	0.2180
TTDPMflex*	53	0.1017	5.77	(1 , 51)	6.4364	0.0200
TTPText	53	0.0006	0.03	(1 , 51)	6.7886	0.8566
TTPTflex*	53	0.1543	9.69	(1 , 51)	6.225	0.003
TTPTflex*[a]	53	0.1543	4.56	(2 , 50)	6.3070	0.0151
PText	53	0.0154	0.80	(1 , 51)	6.7385	0.3766
PTflex	53	0.0379	2.01	(1 , 51)	6.6611	0.1627
40% Body Height						
TTDPMext	53	0.0074	0.38	(1 , 51)	5.8415	0.5400
TTDPMflex	53	0.0338	1.79	(1 , 51)	5.7631	0.1873
TTPText	53	0.0129	0.67	(1 , 51)	5.8253	0.4184
TTPTflex*	53	0.1073	6.13	(1 , 51)	5.5396	0.0166
TTPTflex[a]	53	0.1006	2.80	(2 , 50)	5.6460	0.0706
PText	53	0.0123	0.64	(1 , 51)	5.8270	0.4290
PTflex	53	0.0209	1.09	(1 , 51)	5.8016	0.3015
60% Body Height						
TTDPMext	53	0.0276	1.54	(1 , 51)	5.8834	0.2344
TTDPMflex*	53	0.0745	4.10	(1 , 51)	5.7398	0.0480
TTPText	53	0.0062	0.32	(1 , 51)	5.9478	0.5752
TTPTflex*	53	0.1280	7.49	(1 , 51)	5.5714	0.0085
TTPTflex*[a]	53	0.1262	3.61	(2 , 50)	5.6326	0.0343
PText	53	0.0235	1.23	(1 , 51)	5.8956	0.2727
PTflex	53	0.0603	3.27	(1 , 51)	5.7836	0.0763
80% Body Height						
TTDPMext*	50	0.1749	10.38	(1 , 49)	4.8887	0.0023
TTDPMflex*	50	0.1578	9.18	(1 , 49)	4.9391	0.0039
TTPText	50	0.0007	0.03	(1 , 49)	5.3800	0.8556
TTPTflex	50	0.0199	1.00	(1 , 49)	5.3279	0.3229
TTPTflex[a]	50	0.0541	1.37	(2 , 48)	5.2886	0.2634
PText*	50	0.1050	5.75	(1 , 49)	5.0913	0.0203
PTflex*	50	0.1013	5.53	(1 , 49)	5.1019	0.0228

[a] Grouped variable (three groups)

E.4 KNEE ABDUCTION AT INITIAL CONTACT MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0111	0.57	(1 , 51)	4.3668	0.4534
TTDPMflex	53	0.0006	0.03	(1 , 51)	4.3898	0.8595
TTPText	53	0.0090	0.46	(1 , 51)	4.3713	0.4986
TTPTflex	53	0.0007	0.04	(1 , 51)	4.3895	0.8476
TTPTflex[a]	53	0.0143	0.36	(2 , 50)	4.4030	0.6979
PText	53	0.0116	0.60	(1 , 51)	4.3655	0.4418
PTflex	53	0.0261	1.37	(1 , 51)	4.3334	0.2478
40% Body Height						
TTDPMext	53	0.0205	1.07	(1 , 51)	4.0895	0.3063
TTDPMflex	53	0.0040	0.20	(1 , 51)	4.1239	0.6546
TTPText	53	0.0001	0.00	(1 , 51)	4.1319	0.9490
TTPTflex	53	0.0000	0.00	(1 , 51)	4.132	0.9623
TTPTflex[a]	53	0.0198	0.51	(2 , 50)	4.1317	0.6065
PText	53	0.0254	1.33	(1 , 51)	4.0794	0.2547
PTflex	53	0.0226	1.18	(1 , 51)	4.0852	0.2830
60% Body Height						
TTDPMext	53	0.0076	0.39	(1 , 51)	4.0051	0.5358
TTDPMflex	53	0.0038	0.19	(1 , 51)	4.0127	0.6631
TTPText	53	0.0001	0.00	(1 , 51)	4.0202	0.9548
TTPTflex	53	0.0005	0.03	(1 , 51)	4.0193	0.8748
TTPTflex[a]	53	0.0237	0.61	(2 , 50)	4.0119	0.5490
PText	53	0.0073	0.37	(1 , 51)	4.0056	0.5436
PTflex	53	0.0026	0.13	(1 , 51)	4.0150	0.7163
80% Body Height						
TTDPMext	50	0.0139	0.69	(1 , 49)	4.1156	0.4094
TTDPMflex	50	0.0006	0.03	(1 , 49)	4.1433	0.8622
TTPText	50	0.0000	0.00	(1 , 49)	4.1445	0.9842
TTPTflex	50	0.0041	0.20	(1 , 49)	4.136	0.6548
TTPTflex[a]	50	0.0137	0.33	(2 , 48)	4.1587	0.7176
PText	50	0.0192	0.96	(1 , 49)	4.1046	0.3324
PTflex	50	0.0145	0.72	(1 , 49)	4.1144	0.3997

[a] Grouped variable (three groups), dummy coding was used for regression models

E.5 PEAK KNEE FLEXION MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0256	1.36	(1 , 51)	8.3771	0.2496
TTDPMflex	53	0.0348	1.84	(1 , 51)	8.3390	0.1813
TTPText	53	0.0385	2.04	(1 , 51)	8.3230	0.1593
TTPTflex	53	0.0002	0.01	(1 , 51)	8.4868	0.9113
TTPTflex[a]	53	0.0031	0.08	(2 , 50)	8.5589	0.9252
PText	53	0.0719	3.95	(1 , 51)	8.1771	0.0523
PTflex	53	0.0354	1.87	(1 , 51)	8.3361	0.1771
40% Body Height						
TTDPMext	53	0.0155	0.80	(1 , 51)	9.0311	0.3745
TTDPMflex	53	0.0233	1.22	(1 , 51)	8.9953	0.2755
TTPText	53	0.0495	2.65	(1 , 51)	8.8739	0.1094
TTPTflex	53	0.0007	0.03	(1 , 51)	9.0988	0.8529
TTPTflex[a]	53	0.0075	0.19	(2 , 50)	9.1580	0.8288
PText*	53	0.0844	4.70	(1 , 51)	8.7092	0.0348
PTflex	53	0.0063	0.32	(1 , 51)	9.0730	0.5712
60% Body Height						
TTDPMext	53	0.0221	1.15	(1 , 51)	9.0171	0.2877
TTDPMflex	53	0.0304	1.60	(1 , 51)	8.9790	0.2119
TTPText	53	0.0165	0.86	(1 , 51)	9.0429	0.3590
TTPTflex	53	0.0075	0.39	(1 , 51)	9.0841	0.5365
TTPTflex[a]	53	0.0145	0.37	(2 , 50)	9.1421	0.6933
PText	53	0.0326	1.72	(1 , 51)	8.9688	0.1959
PTflex	53	0.0013	0.06	(1 , 51)	9.1128	0.8013
80% Body Height						
TTDPMext*	50	0.0781	4.15	(1 , 49)	8.6585	0.0471
TTDPMflex	50	0.0654	3.43	(1 , 49)	8.7179	0.0702
TTPText	50	0.0000	0.00	(1 , 49)	9.0175	0.9700
TTPTflex	50	0.0003	0.02	(1 , 49)	9.0161	0.8973
TTPTflex[a]	50	0.0007	0.02	(2 , 48)	9.1077	0.9824
PText	50	0.0430	2.20	(1 , 49)	8.8215	0.1442
PTflex	50	0.0001	0.01	(1 , 49)	9.0171	0.9379

[a] Grouped variable (three groups)

E.6 PEAK KNEE ABDUCTION MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0059	0.30	(1 , 51)	6.5336	0.5854
TTDPMflex	53	0.0089	0.46	(1 , 51)	6.5235	0.5008
TTPText	53	0.0272	1.42	(1 , 51)	6.4632	0.2383
TTPTflex	53	0.0066	0.34	(1 , 51)	6.5313	0.5639
TTPTflex[a]	53	0.0024	0.06	(2 , 50)	6.6099	0.9407
PText	53	0.0015	0.07	(1 , 51)	6.5481	0.7863
PTflex	53	0.0149	0.77	(1 , 51)	6.5038	0.3836
40% Body Height						
TTDPMext	53	0.0019	0.10	(1 , 51)	6.7122	0.7585
TTDPMflex	53	0.0071	0.37	(1 , 51)	6.6946	0.5482
TTPText	53	0.0274	1.44	(1 , 51)	6.6258	0.2362
TTPTflex	53	0.0035	0.18	(1 , 51)	6.7068	0.6750
TTPTflex[a]	53	0.0005	0.01	(2 , 50)	6.7837	0.9874
PText	53	0.0100	0.51	(1 , 51)	6.6850	0.4771
PTflex	53	0.0024	0.12	(1 , 51)	6.7106	0.7295
60% Body Height						
TTDPMext	53	0.0018	0.09	(1 , 51)	6.9165	0.7636
TTDPMflex	53	0.0036	0.19	(1 , 51)	6.9100	0.6674
TTPText	53	0.0012	0.03	(1 , 51)	6.9873	0.9705
TTPTflex	53	0.0001	0.01	(1 , 51)	6.9222	0.9329
TTPTflex[a]	53	0.0012	0.03	(2 , 50)	6.9873	0.9705
PText	53	0.0258	1.35	(1 , 51)	6.8326	0.2501
PTflex	53	0.0000	0.00	(1 , 51)	6.9225	0.9630
80% Body Height						
TTDPMext	50	0.0008	0.04	(1 , 49)	7.5025	0.8392
TTDPMflex	50	0.0014	0.07	(1 , 49)	7.5003	0.7921
TTPText	50	0.0115	0.57	(1 , 49)	7.4624	0.4538
TTPTflex	50	0.0080	0.39	(1 , 49)	7.4758	0.5335
TTPTflex[a]	50	0.0027	0.07	(2 , 48)	7.5731	0.9363
PText	50	0.0059	0.29	(1 , 49)	7.4836	0.5925
PTflex	50	0.0027	0.13	(1 , 49)	7.4956	0.7180

[a] Grouped variable (three groups)

E.7 PEAK KNEE ABDUCTION MOMENT MODELS BY JUMP DISTANCE

	n	R^2	Adj R^{2*}	F	(df1, df2)	MSE	p-value
20% Body Height							
TTDPMext	53	0.0244	0.0053	1.28	(1 , 51)	0.4664	0.2640
TTDPMflex	53	0.0141	0.0052	0.73	(1 , 51)	0.4689	0.3966
TTPText	53	0.0003	-0.0193	0.01	(1 , 51)	0.4721	0.9049
TTPTflex	53	0.0273	0.0082	1.43	(1 , 51)	0.4657	0.2371
TTPTflex[a]	53	0.1279	0.0930	3.67	(2 , 50)	0.4454	0.0327
PText	53	0.0087	-0.0107	0.45	(1 , 51)	0.4701	0.5062
PTflex	53	0.0028	-0.0168	0.14	(1 , 51)	0.4715	0.7088
40% Body Height							
TTDPMext	53	0.0411	0.0223	2.19	(1 , 51)	0.5159	0.1452
TTDPMflex	53	0.0134	-0.0060	0.69	(1 , 51)	0.5233	0.4095
TTPText	53	0.0271	0.0080	1.42	(1 , 51)	0.5196	0.2389
TTPTflex	53	0.0700	0.0518	3.84	(1 , 51)	0.5081	0.0555
TTPTflex[a]	53	0.1244	0.0893	3.55	(2 , 50)	0.4979	0.0362
PText	53	0.0010	-0.0186	0.05	(1 , 51)	0.5266	0.8266
PTflex	53	0.0222	0.0031	1.16	(1 , 51)	0.5209	0.2867
60% Body Height							
TTDPMext	53	0.0226	0.0034	1.18	(1 , 51)	0.6267	0.2830
TTDPMflex	53	0.0056	-0.0139	0.29	(1 , 51)	0.6321	0.5933
TTPText	53	0.0008	-0.0188	0.04	(1 , 51)	0.6336	0.8446
TTPTflex	53	0.0682	0.0682	3.73	(1 , 51)	0.6119	0.0590
TTPTflex[a]	53	0.0994	0.0634	2.76	(2 , 50)	0.6075	0.0730
PText	53	0.0017	-0.0179	0.09	(1 , 51)	0.6333	0.7696
PTflex	53	0.0345	0.0156	1.82	(1 , 51)	0.6228	0.1829
80% Body Height							
TTDPMext	50	0.0003	-0.0201	0.02	(1 , 49)	0.9839	0.9029
TTDPMflex	50	0.0035	-0.0168	0.17	(1 , 49)	0.9823	0.6783
TTPText	50	0.0106	-0.0096	0.52	(1 , 49)	0.9788	0.4731
TTPTflex	50	0.0381	0.0185	1.94	(1 , 49)	0.9651	0.1699
TTPTflex[a]	50	0.0407	0.0007	1.02	(2 , 48)	0.9738	0.3689
PText	50	0.0035	-0.0169	0.17	(1 , 49)	0.9823	0.6818
PTflex	50	0.0066	-0.0137	0.32	(1 , 49)	0.9808	0.5715

[a] Grouped variable (three groups)

* Adjusted R2 reported to compare between raw and transformed data (E.10)

E.8 PEAK PATSF MODELS BY JUMP DISTANCE

	n	R^2	F	(df1, df2)	MSE	p-value
20% Body Height						
TTDPMext	53	0.0030	0.15	(1 , 51)	20.5910	0.6977
TTDPMflex	53	0.0158	0.82	(1 , 51)	20.4590	0.3703
TTPText	53	0.0002	0.01	(1 , 51)	20.6200	0.9176
TTPTflex	53	0.0148	0.77	(1 , 51)	20.4690	0.3851
TTPTflex[a]	53	0.0267	0.69	(2 , 50)	20.5480	0.5085
PText	53	0.0034	0.17	(1 , 51)	20.5870	0.6791
PTflex	53	0.0326	1.72	(1 , 51)	20.2840	0.1960
40% Body Height						
TTDPMext	53	0.0038	0.19	(1 , 51)	22.0450	0.6622
TTDPMflex	53	0.0082	0.42	(1 , 51)	21.9960	0.5191
TTPText	53	0.0002	0.01	(1 , 51)	22.0840	0.9118
TTPTflex	53	0.0137	0.71	(1 , 51)	21.9350	0.4036
TTPTflex[a]	53	0.0201	0.51	(2 , 50)	22.0810	0.6014
PText	53	0.0008	0.04	(1 , 51)	22.0780	0.8372
PTflex	53	0.0443	2.36	(1 , 51)	21.5920	0.1303
60% Body Height						
TTDPMext	53	0.0070	0.36	(1 , 51)	20.6260	0.5506
TTDPMflex	53	0.0168	0.87	(1 , 51)	20.5240	0.3543
TTPText	53	0.0000	0.00	(1 , 51)	20.6990	0.9836
TTPTflex	53	0.0066	0.34	(1 , 51)	20.6310	0.5641
TTPTflex[a]	53	0.0107	0.27	(2 , 50)	20.7930	0.7648
PText	53	0.0086	0.44	(1 , 51)	20.6100	0.5082
PTflex	53	0.0504	2.71	(1 , 51)	20.1710	0.1062
80% Body Height						
TTDPMext	50	0.0035	0.17	(1 , 49)	15.5280	0.6817
TTDPMflex	50	0.0076	0.38	(1 , 49)	15.4960	0.5425
TTPText	50	0.0017	0.08	(1 , 49)	15.5420	0.7719
TTPTflex	50	0.0372	1.89	(1 , 49)	15.2630	0.1749
TTPTflex[a]	50	0.0705	1.89	(2 , 48)	15.1520	0.1731
PText	50	0.0107	0.53	(1 , 49)	15.4710	0.4699
PTflex	50	0.0334	1.69	(1 , 49)	15.2930	0.1993

[a] Grouped variable (three groups)

E.9 PEAK KNEE ABDUCTION MOMENT (SQUARE ROOT) BY JUMP DISTANCE

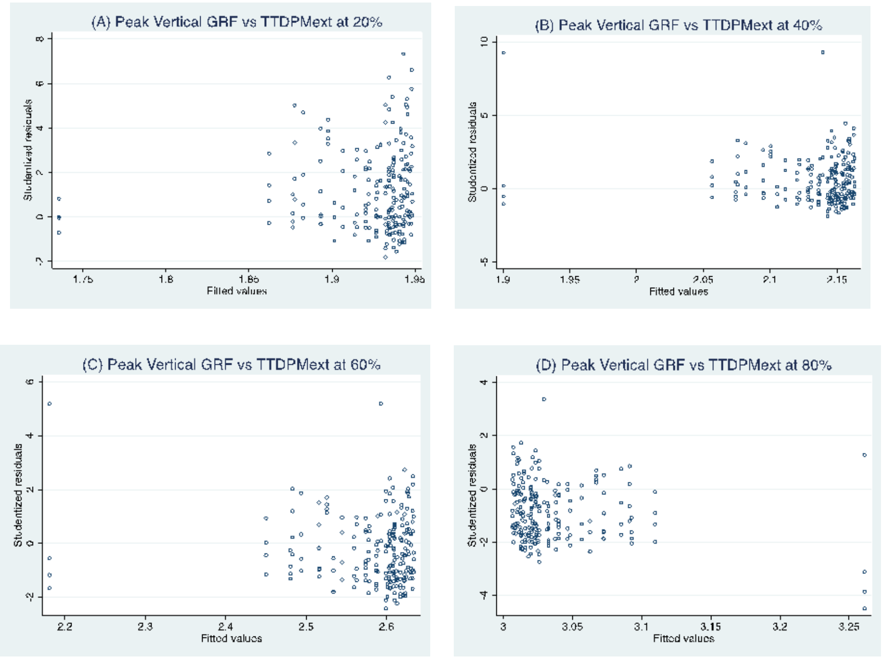
	n	R^2	Adj R^2	F	(df1, df2)	MSE	p-value
20% Body Height							
TTDPMext	53	0.0201	0.0009	1.05	(1 , 51)	0.2746	0.3107
TTDPMflex	53	0.0099	0.0095	0.51	(1 , 51)	0.2761	0.4785
TTPText	53	0.0002	-0.0194	0.01	(1 , 51)	0.2774	0.9286
TTPTflex	53	0.0235	0.0044	1.23	(1 , 51)	0.2742	0.2729
TTPTflex[a]	53	0.1272	0.0923	3.64	(2 , 50)	0.2618	0.0334
PText	53	0.0091	0.0103	0.47	(1 , 51)	0.2762	0.4958
PTflex	53	0.0034	0.0161	0.18	(1 , 51)	0.2770	0.6769
40% Body Height							
TTDPMext	53	0.0369	0.0180	1.95	(1 , 51)	0.3100	0.1681
TTDPMflex	53	0.0115	0.0079	0.59	(1 , 51)	0.3141	0.4451
TTPText	53	0.0274	0.0084	1.44	(1 , 51)	0.3115	0.2359
TTPTflex	53	0.1011	0.0831	5.62	(1 , 51)	0.2856	0.0216
TTPTflex[a]	53	0.1522	0.1183	4.49	(2 , 50)	0.2938	0.0161
PText	53	0.0010	-0.0186	0.05	(1 , 51)	0.3157	0.8228
PTflex	53	0.0280	0.0089	1.47	(1 , 51)	0.3114	0.2312
60% Body Height							
TTDPMext	53	0.0225	0.0034	1.18	(1 , 51)	0.3118	0.2833
TTDPMflex	53	0.0088	0.0107	0.45	(1 , 51)	0.3140	0.5051
TTPText	53	0.0001	0.0195	0.01	(1 , 51)	0.3154	0.9404
TTPTflex	53	0.0997	0.0820	5.65	(1 , 51)	0.2993	0.0213
TTPTflex[a]	53	0.1522	0.1183	4.49	(2 , 50)	0.2933	0.0161
PText	53	0.0013	0.0182	0.07	(1 , 51)	0.3152	0.7952
PTflex	53	0.0379	0.0191	2.01	(1 , 51)	0.3094	0.1623
80% Body Height							
TTDPMext	50	0.0007	-0.0197	0.03	(1 , 49)	0.3887	0.0858
TTDPMflex	50	0.0030	0.0173	0.15	(1 , 49)	0.3882	0.7009
TTPText	50	0.0110	0.0092	0.55	(1 , 49)	0.3867	0.4636
TTPTflex	50	0.0599	0.0407	3.12	(1 , 49)	0.3740	0.0834
TTPTflex[a]	50	0.0820	0.0437	2.14	(2 , 48)	0.3764	0.1284
PText	50	0.0015	0.0188	0.08	(1 , 49)	0.3885	0.7843
PTflex	50	0.0228	0.0028	1.14	(1 , 49)	0.3844	0.2900

APPENDIX F

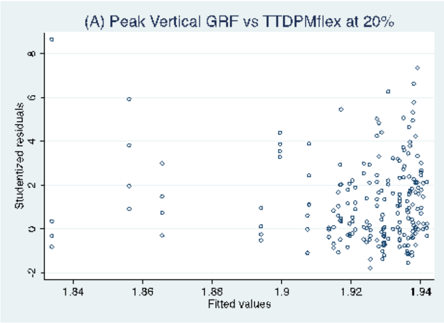
SIMPLE REGRESSION: JACKKNIFE RESIDUAL VS FITTED VALUES PLOTS

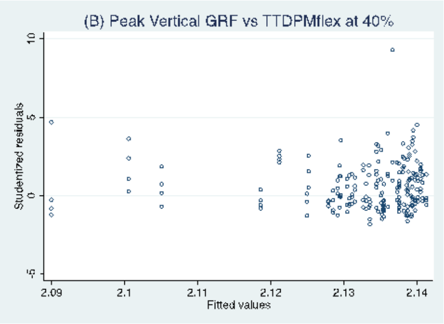
F.1 VERTICAL GROUND REACTION FORCE

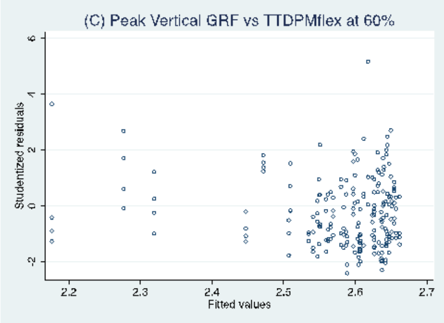
Peak Vertical Ground Reaction Force vs. TTDPMext

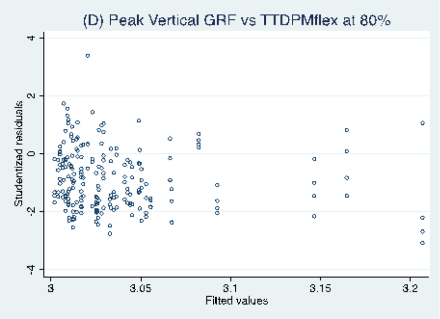


Peak Vertical Ground Reaction Force vs. TTDPMflex

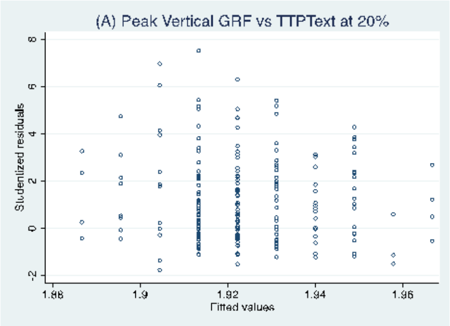


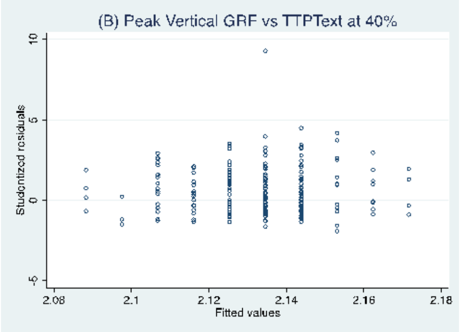


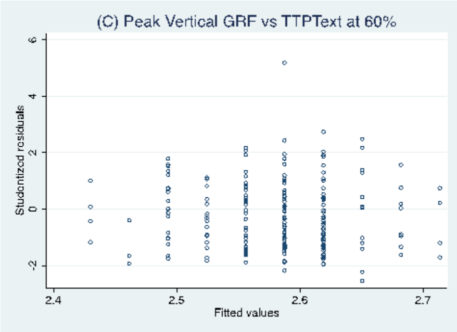


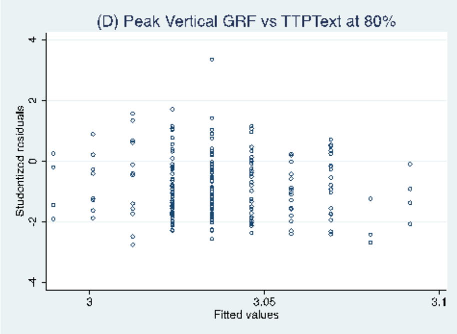


Peak Vertical Ground Reaction Force vs. TTPText

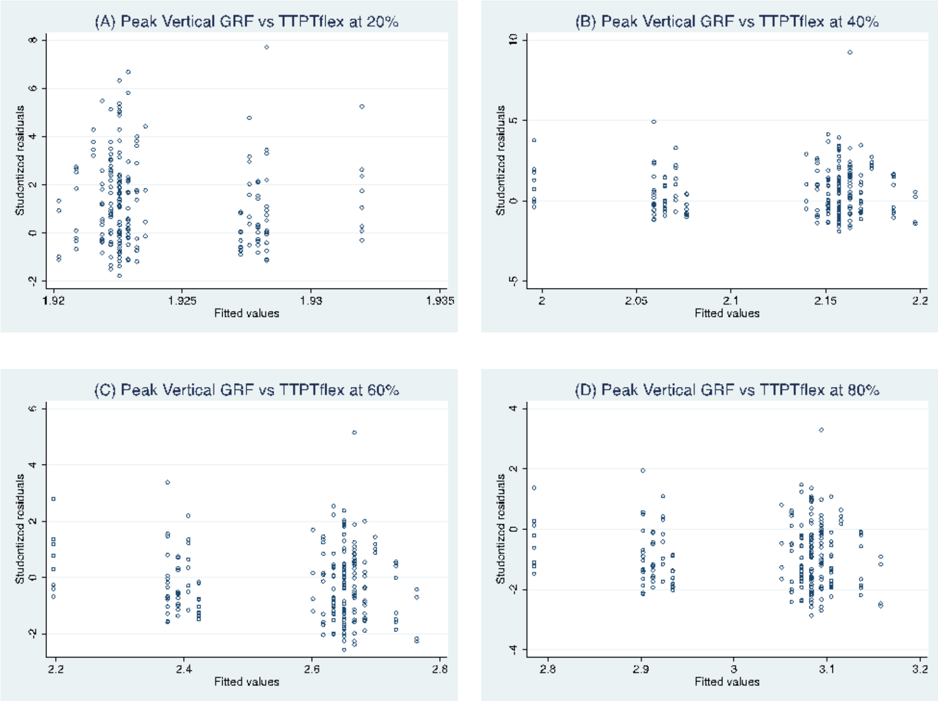




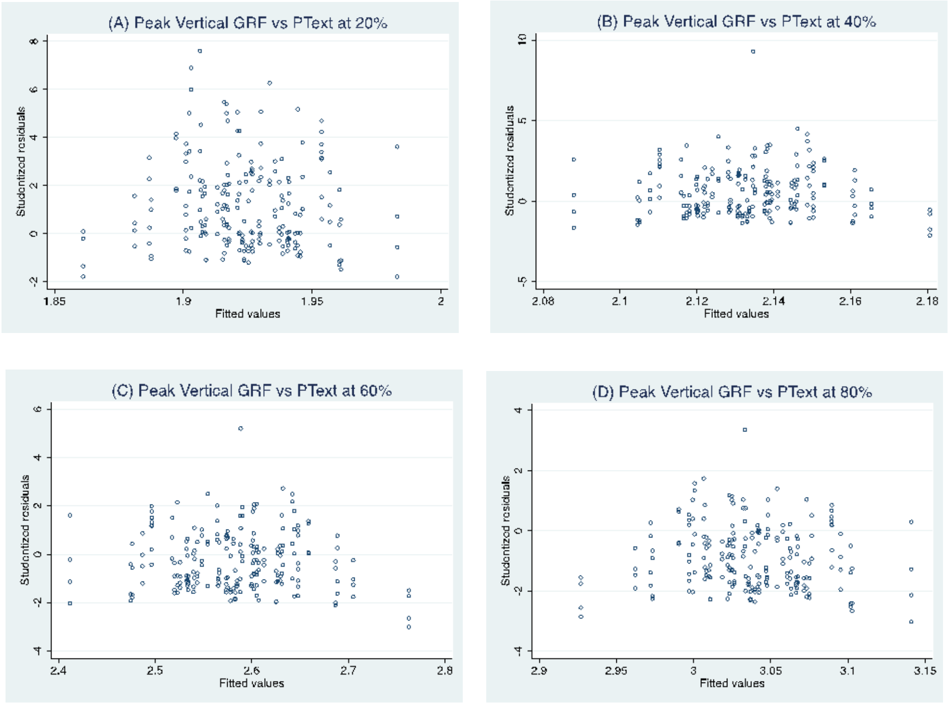




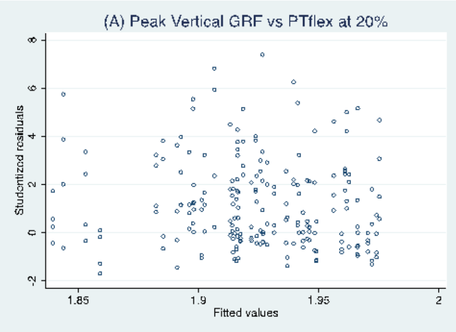
Peak Vertical Ground Reaction Force vs. TTPTflex

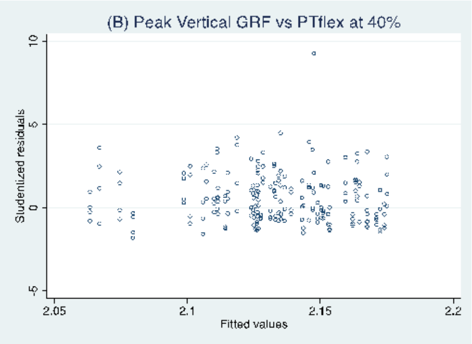


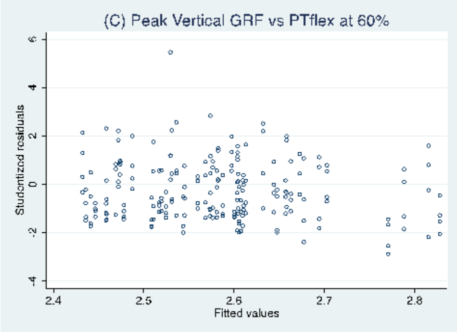
Peak Vertical Ground Reaction Force vs. PText

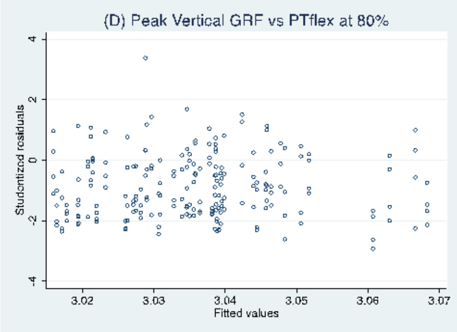


Peak Vertical Ground Reaction Force vs. PTflex


(A) Peak Vertical GRF vs PTflex at 20%

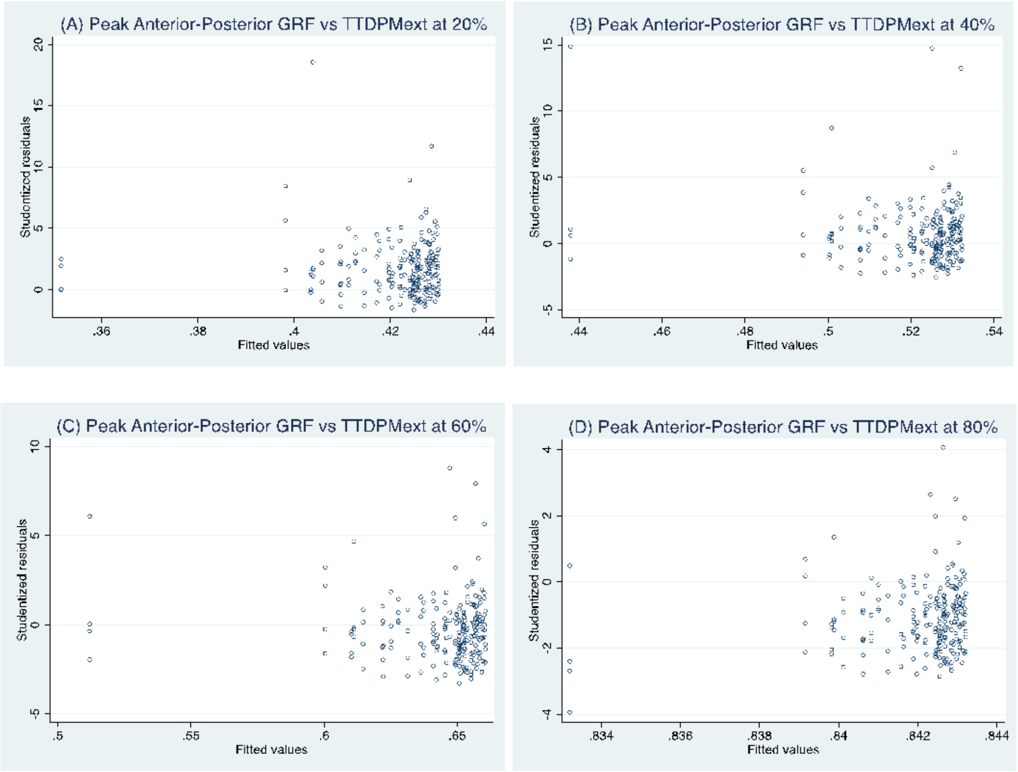

(B) Peak Vertical GRF vs PTflex at 40%


(C) Peak Vertical GRF vs PTflex at 60%

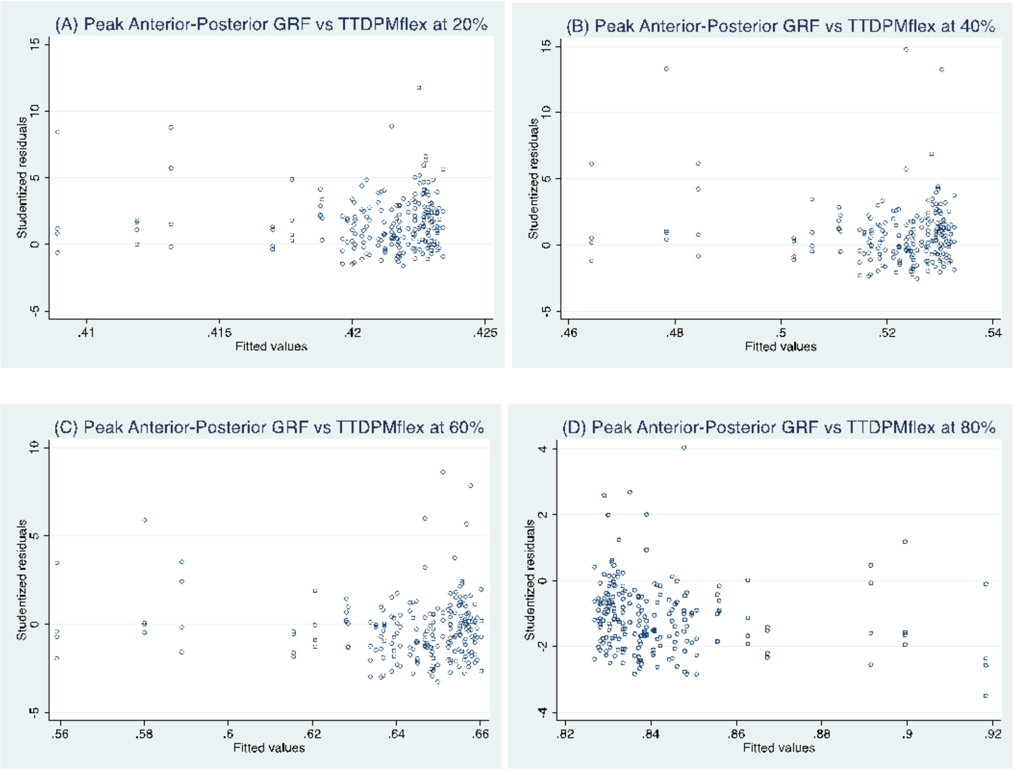

(D) Peak Vertical GRF vs PTflex at 80%

F.2 ANTERIOR-POSTERIOR GROUND REACTION FORCE

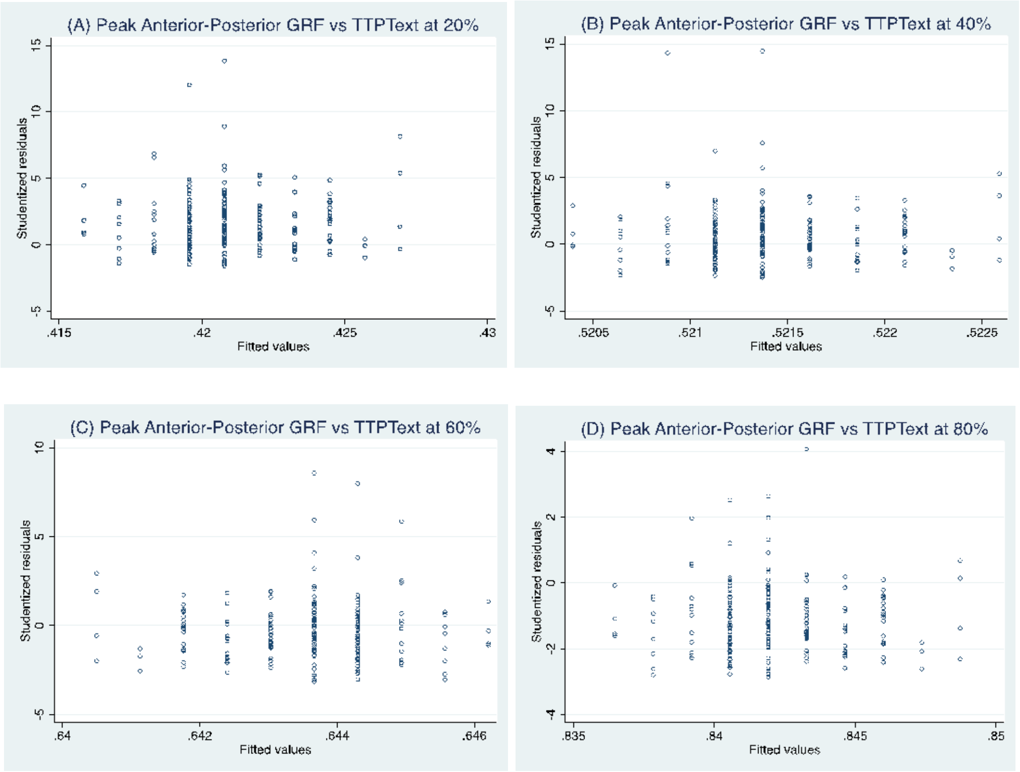
Peak Anterior-Posterior Ground Reaction Force vs. TTDPMext



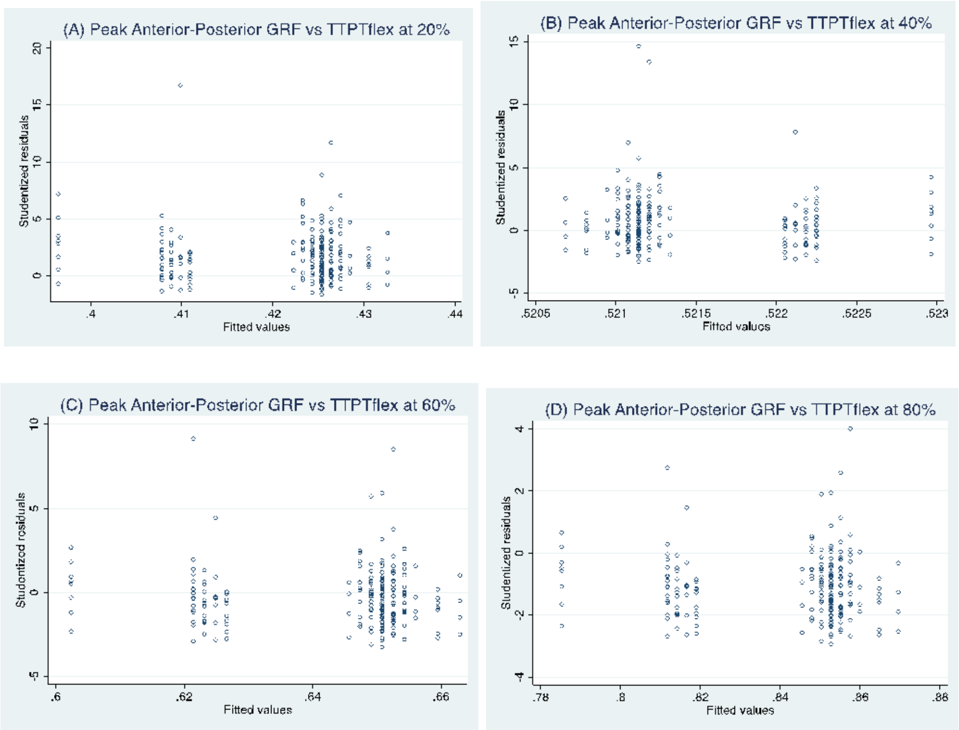
Peak Anterior-Posterior Ground Reaction Force vs. TTDPMflex



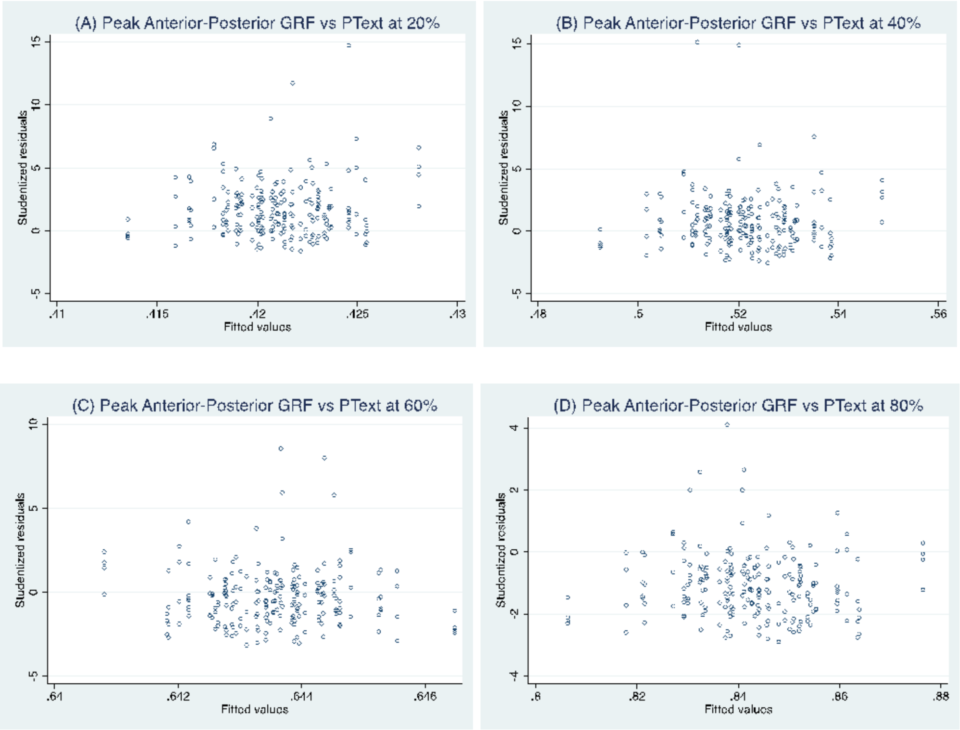
Peak Anterior-Posterior Ground Reaction Force vs. TTPText



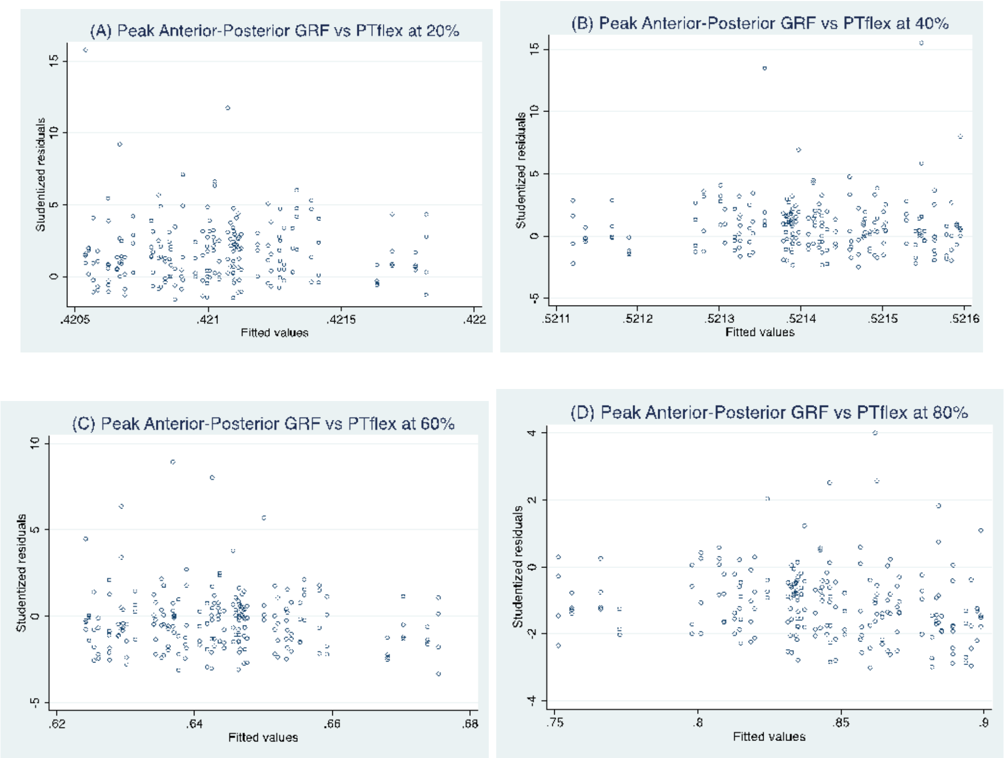
Peak Anterior-Posterior Ground Reaction Force vs. TTPTflex



Peak Anterior-Posterior Ground Reaction Force vs. PText

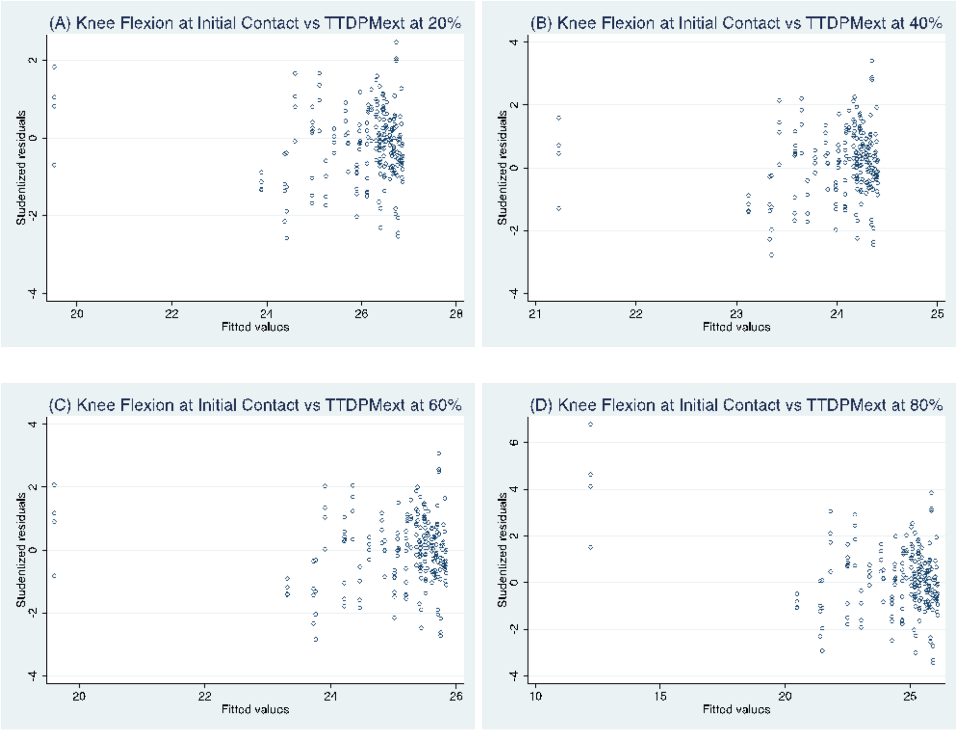


Peak Anterior-Posterior Ground Reaction Force vs. PTflex

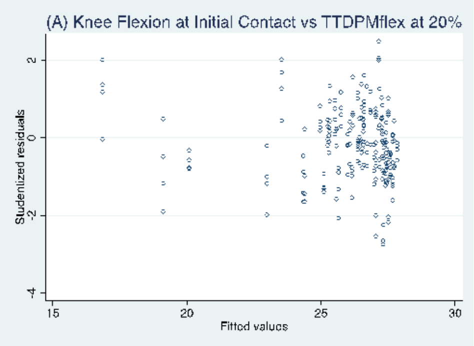


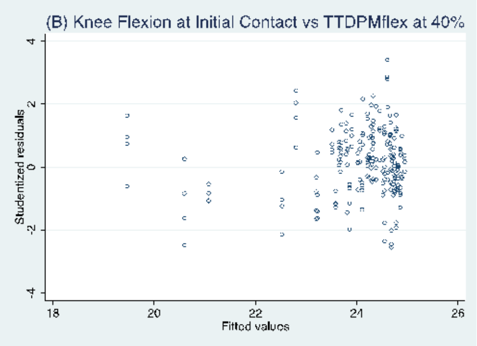
F.3 KNEE FLEXION AT INITIAL CONTACT

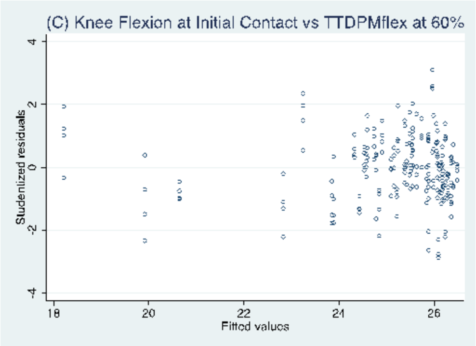
Knee Flexion at Initial Contact vs. TTDPMext

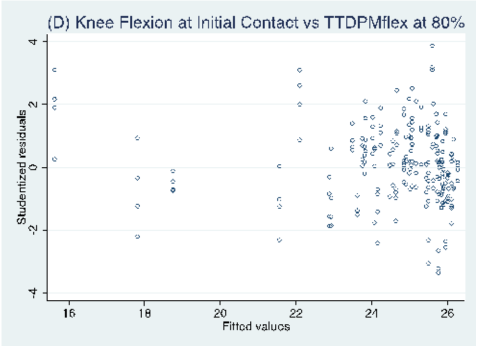


Knee Flexion at Initial Contact vs. TTDPMflex

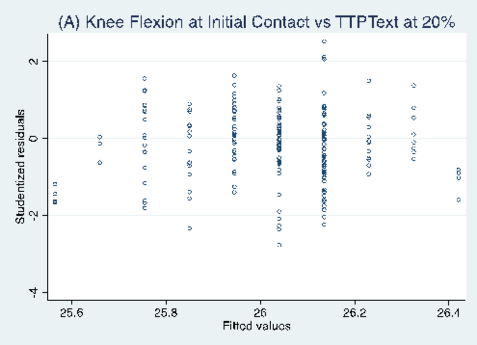


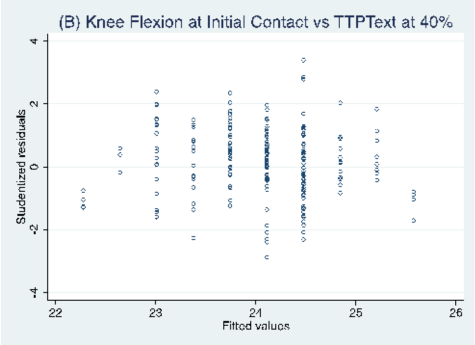


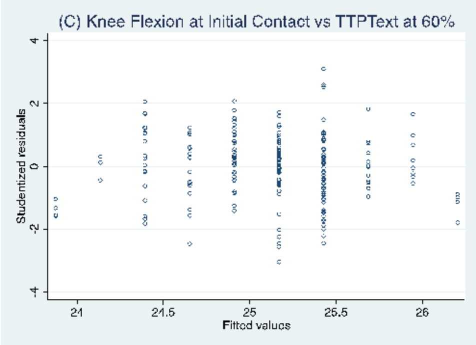


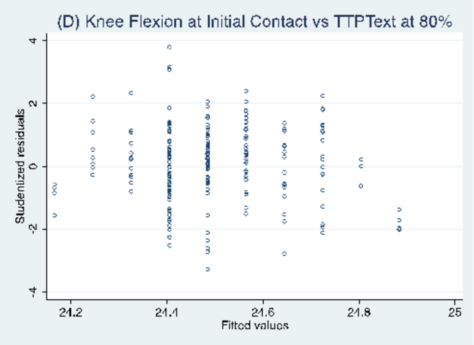


Knee Flexion at Initial Contact vs. TTPText

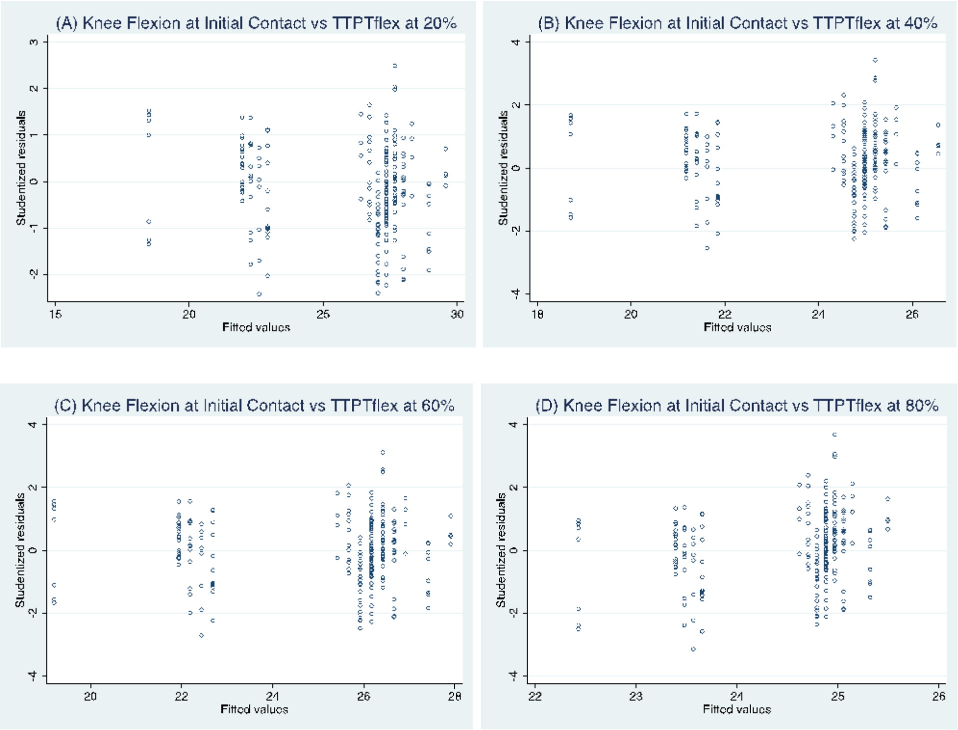




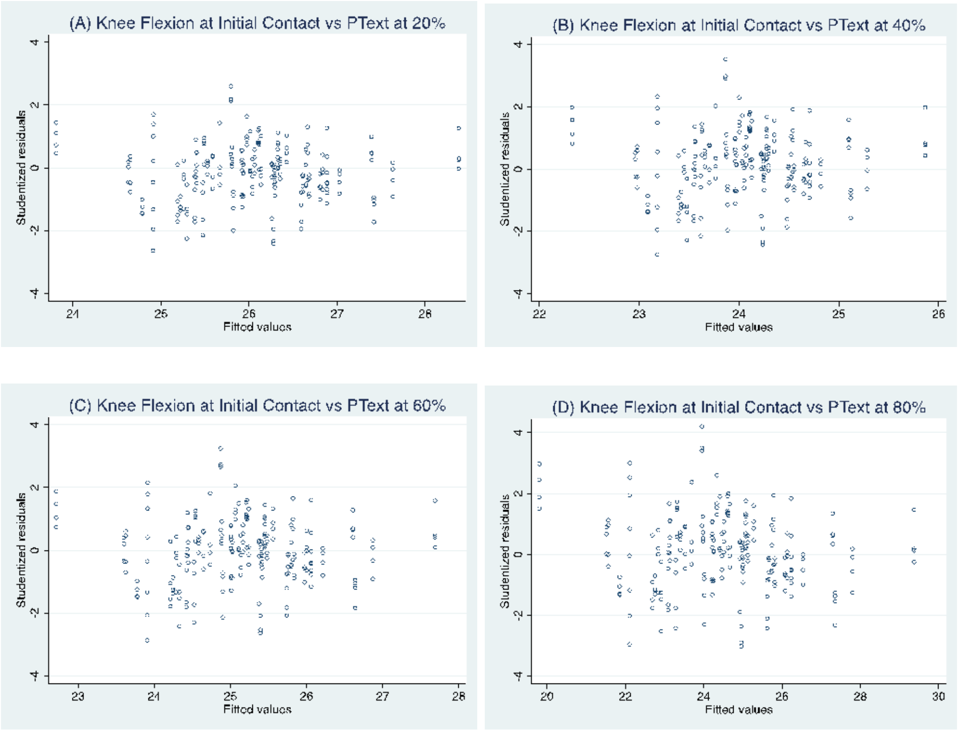




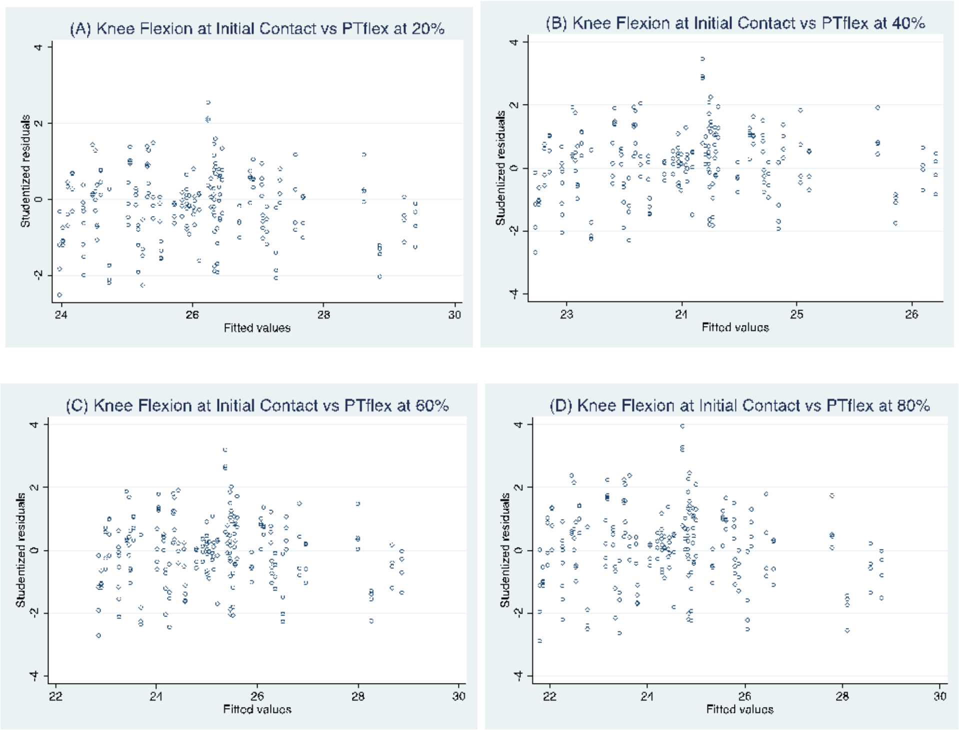
Knee Flexion at Initial Contact vs. TTPTflex



Knee Flexion at Initial Contact vs. PText

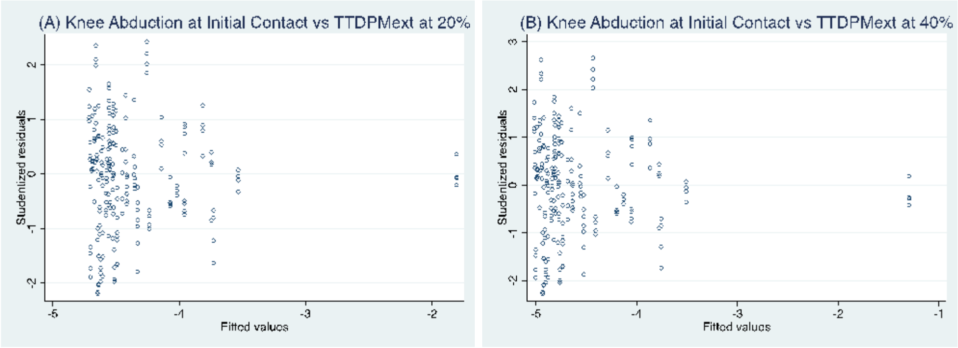


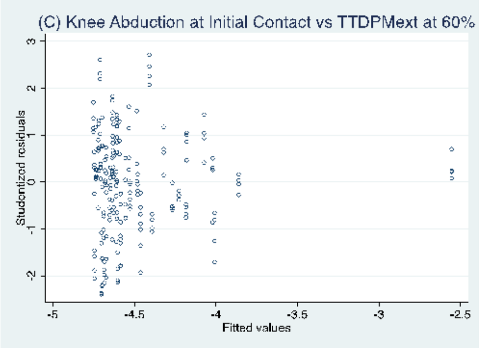
Knee Flexion at Initial Contact vs. PTflex

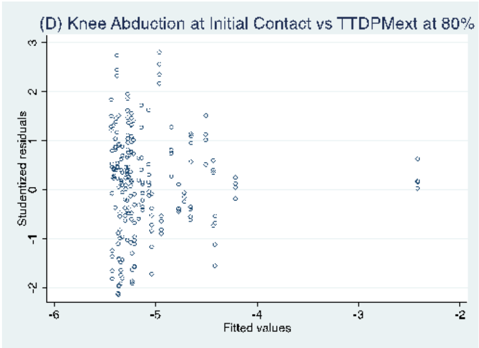


F.4 KNEE ABDUCTION AT INITIAL CONTACT

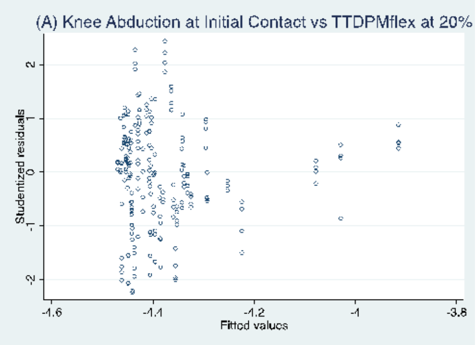
Knee Abduction at Initial Contact vs. TTDPMext

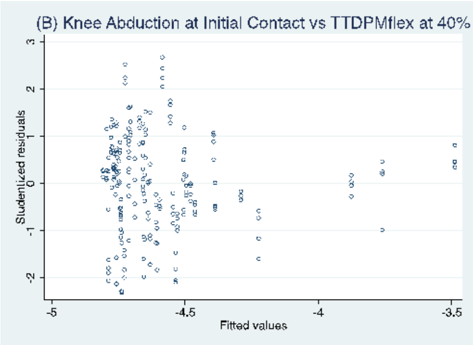


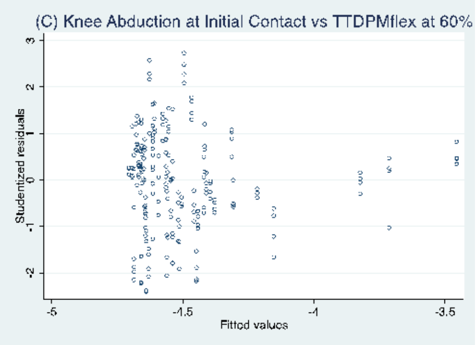


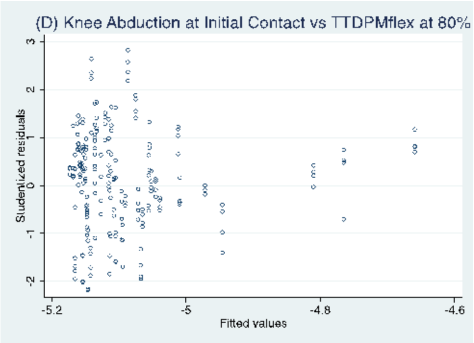


Knee Abduction at Initial Contact vs. TTDPMflex

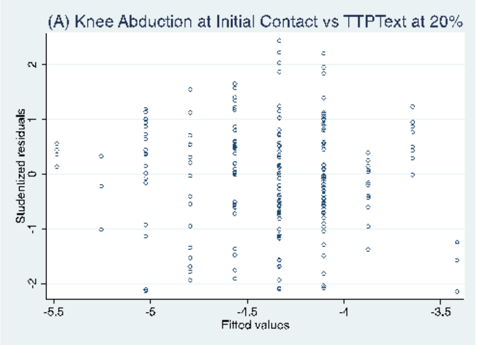


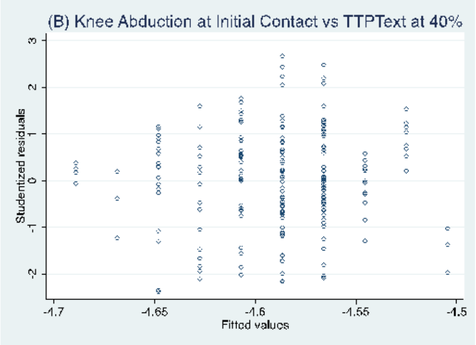


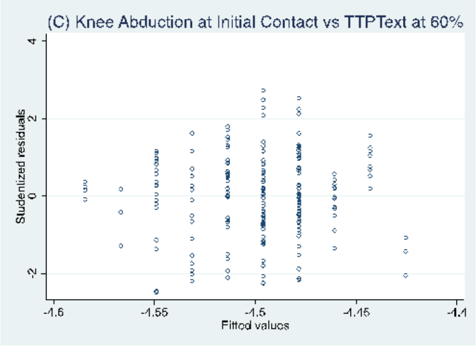


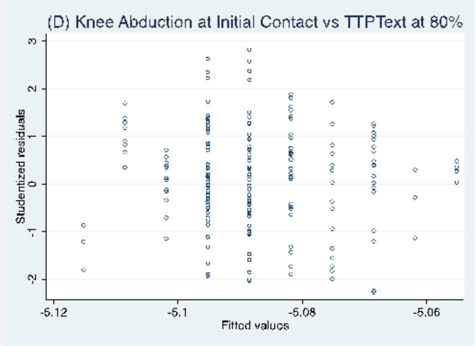


Knee Abduction at Initial Contact vs. TTPText

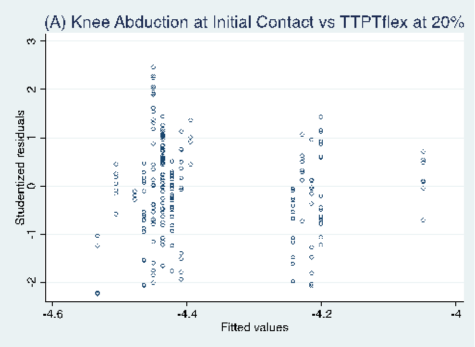


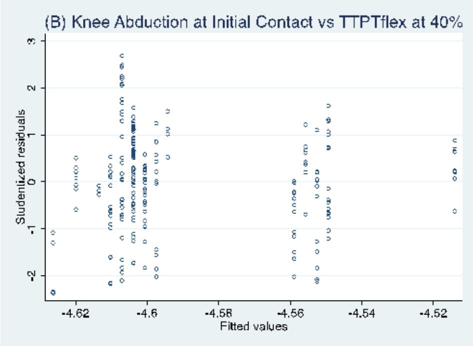


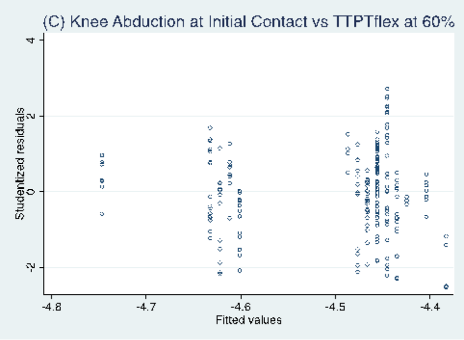

(C) Knee Abduction at Initial Contact vs TTPText at 60%

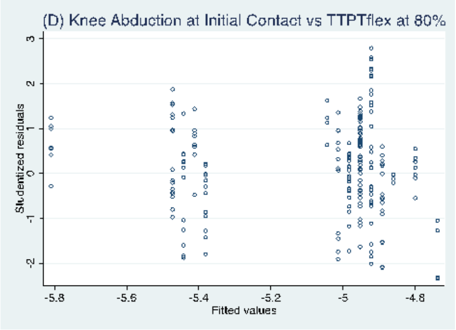

(D) Knee Abduction at Initial Contact vs TTPText at 80%

Knee Abduction at Initial Contact vs. TTPTflex

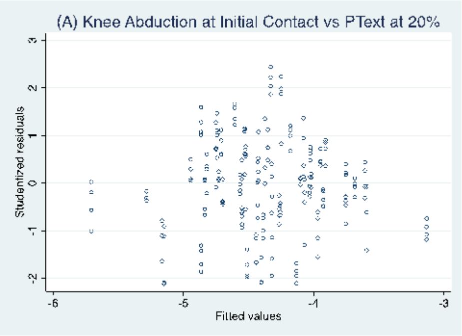

(A) Knee Abduction at Initial Contact vs TTPTflex at 20%

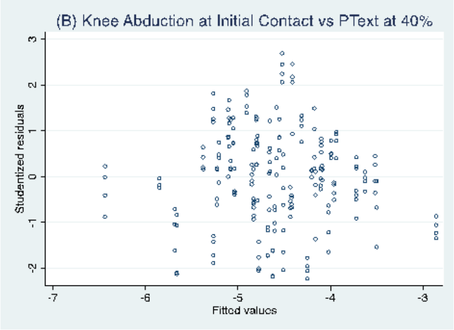

(B) Knee Abduction at Initial Contact vs TTPTflex at 40%

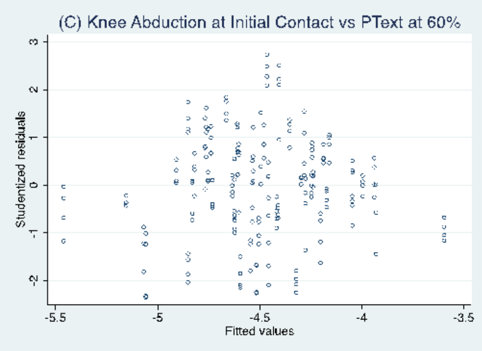

(C) Knee Abduction at Initial Contact vs TTPTflex at 60%

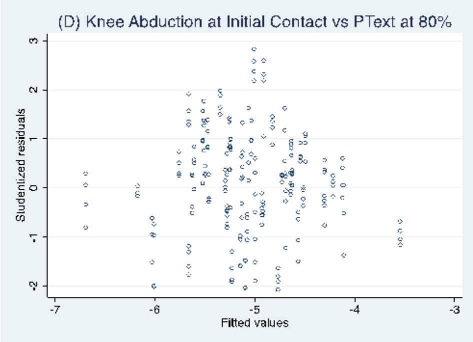

(D) Knee Abduction at Initial Contact vs TTPTflex at 80%

Knee Abduction at Initial Contact vs. PText

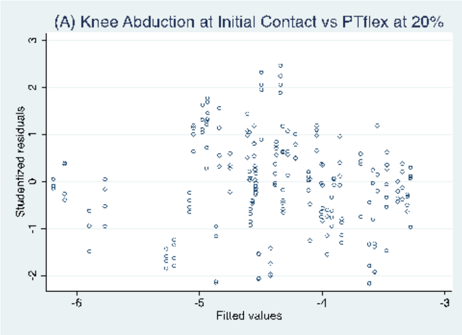

(A) Knee Abduction at Initial Contact vs PText at 20%

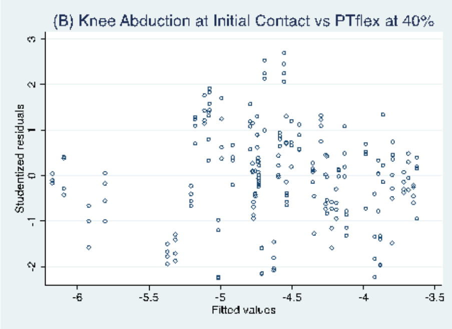

(B) Knee Abduction at Initial Contact vs PText at 40%

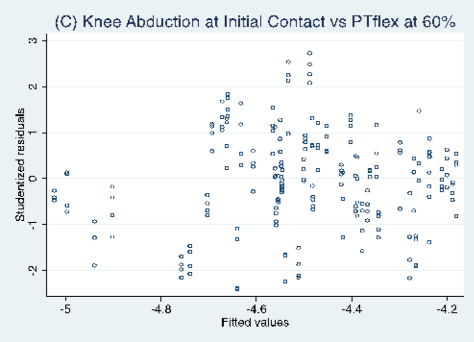


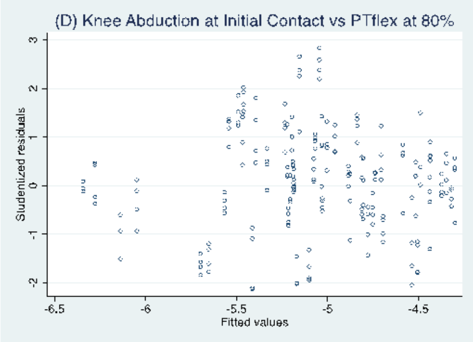


Knee Abduction at Initial Contact vs. PTflex



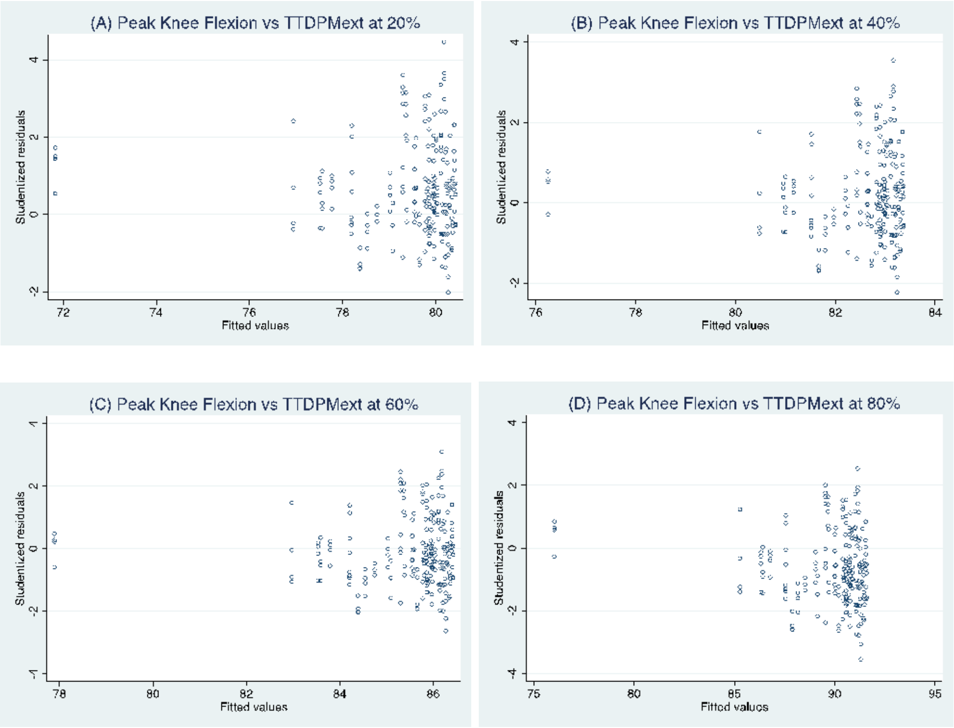




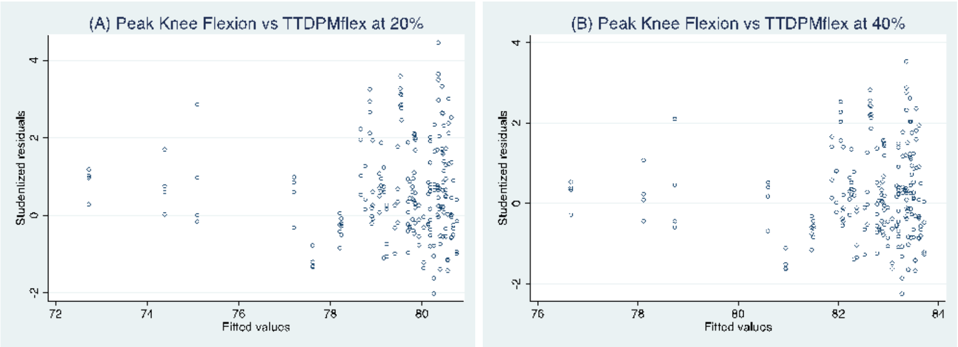


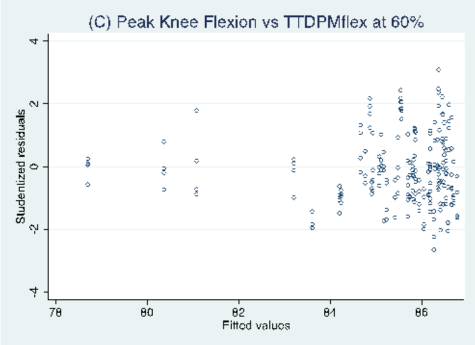
F.5 PEAK KNEE FLEXION

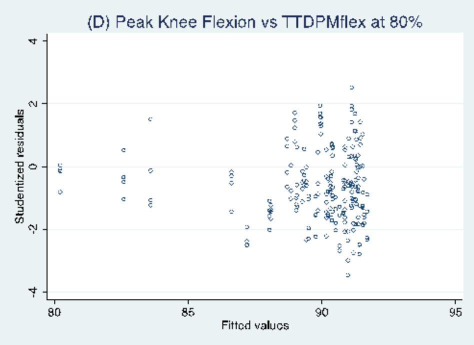
Peak Knee Flexion vs. TTDPMext



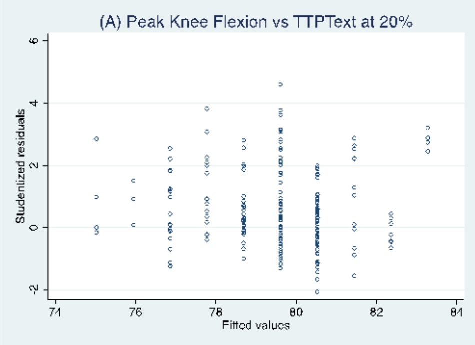
Peak Knee Flexion vs. TTDPMflex

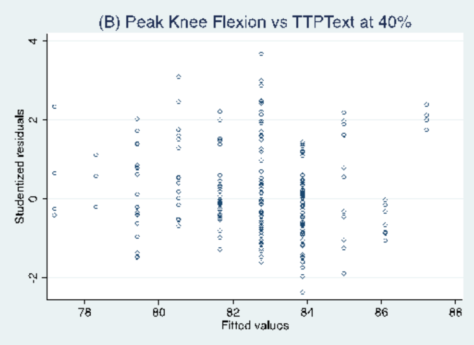


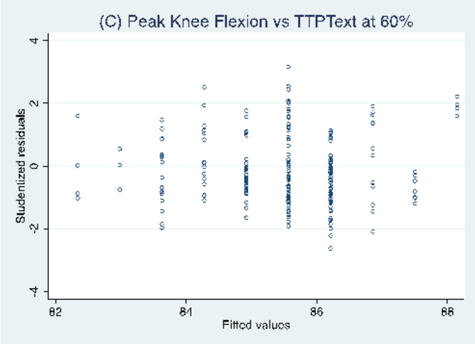


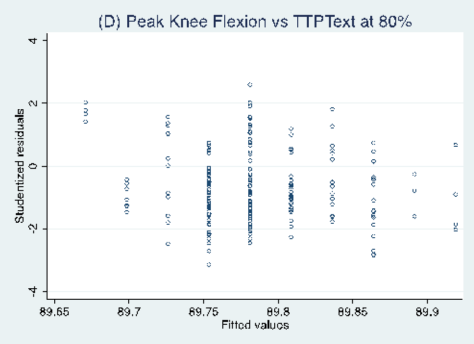


Peak Knee Flexion vs. TTPText

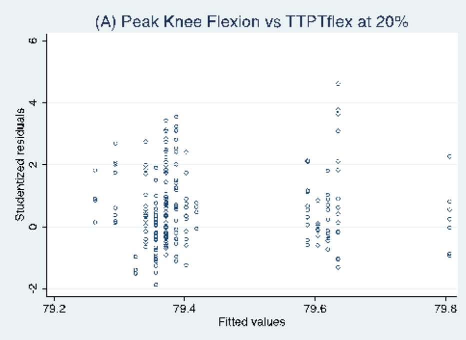


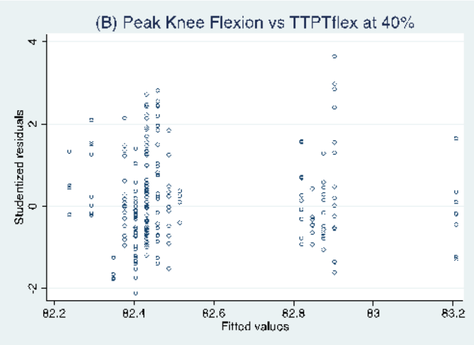


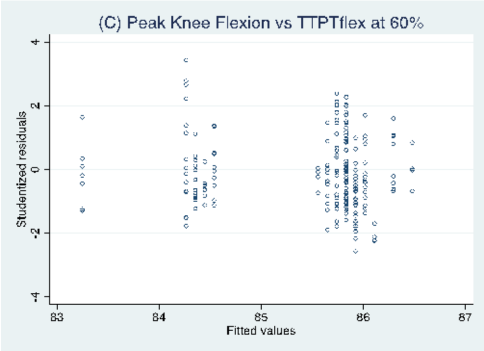


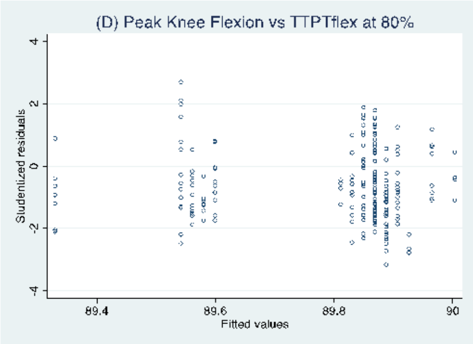


Peak Knee Flexion vs. TTPTflex

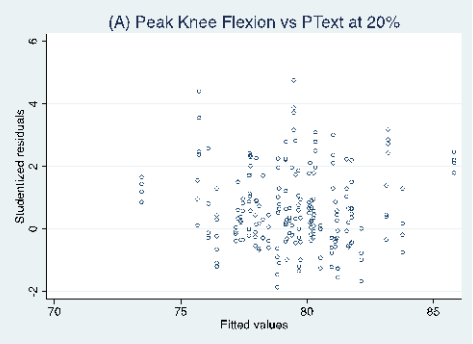


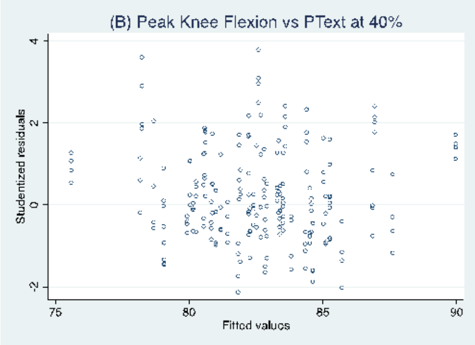


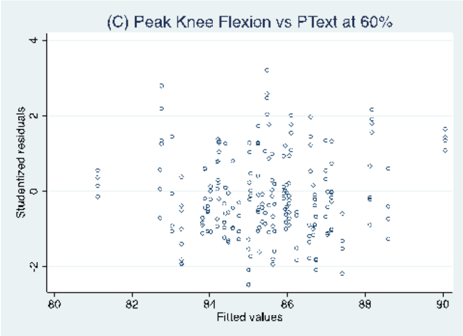


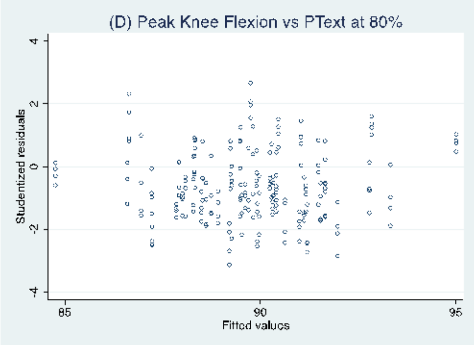


Peak Knee Flexion vs. PText

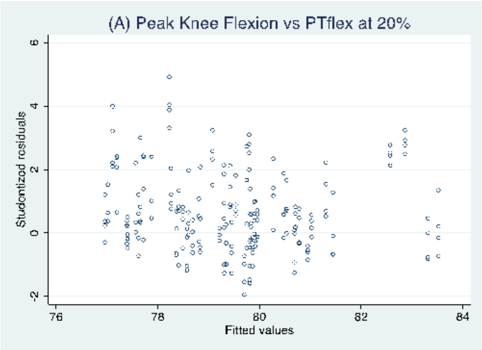


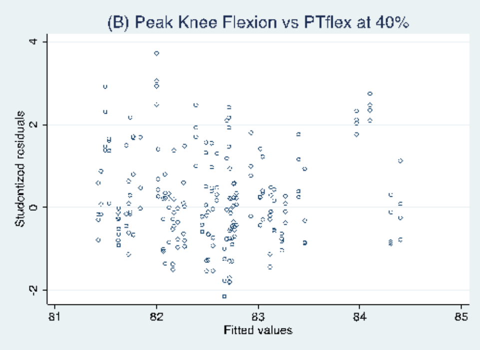






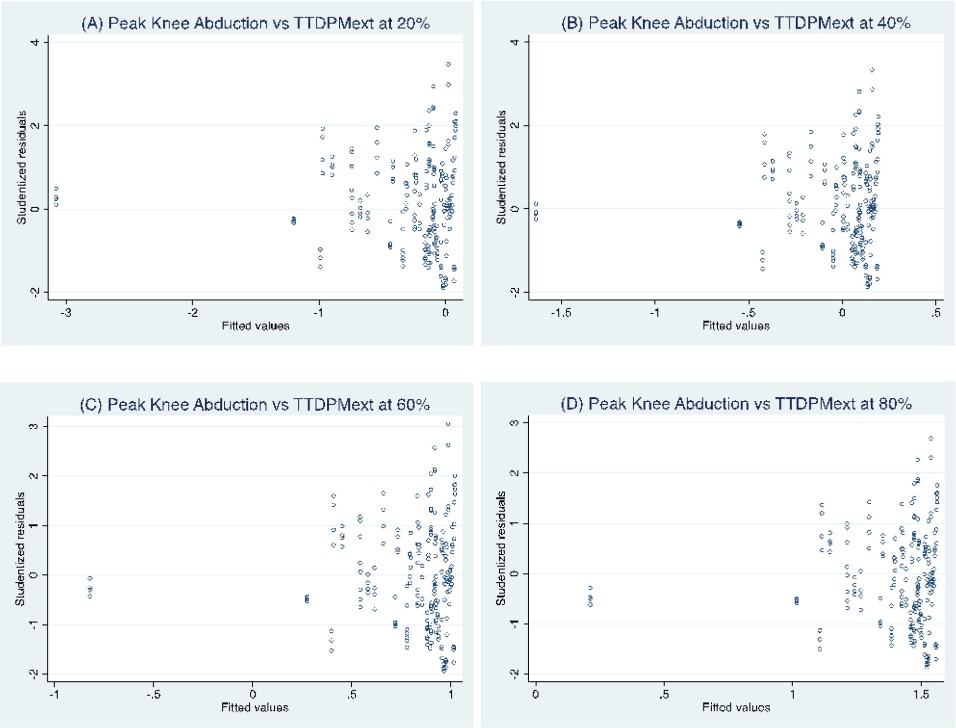
Peak Knee Flexion vs. PTflex



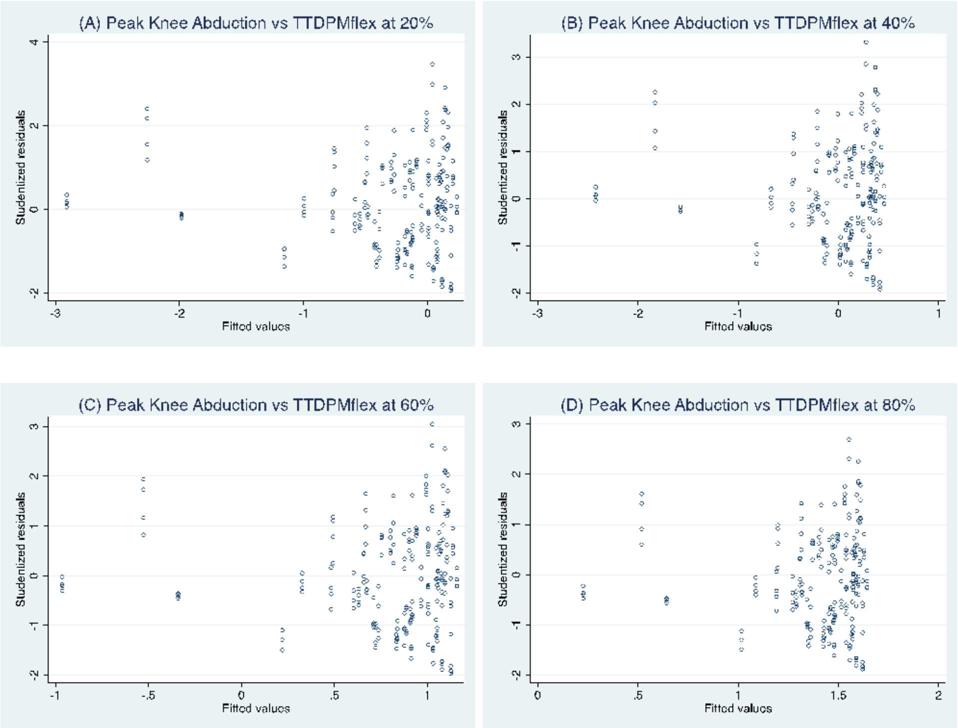


F.6 PEAK KNEE ABDUCTION

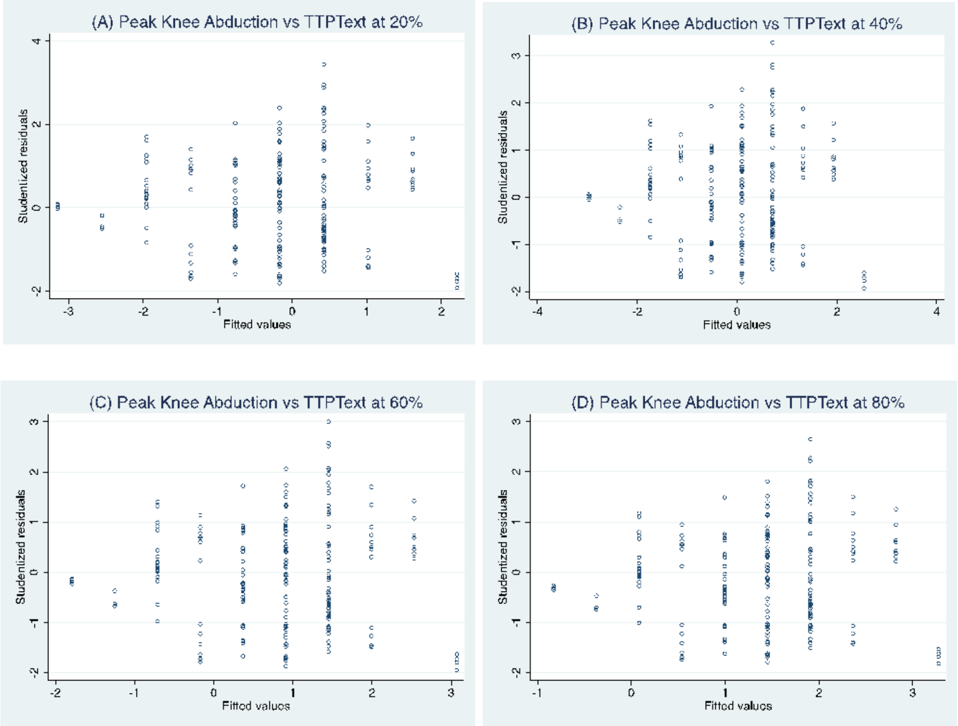
Peak Knee Abduction vs. TTDPMext



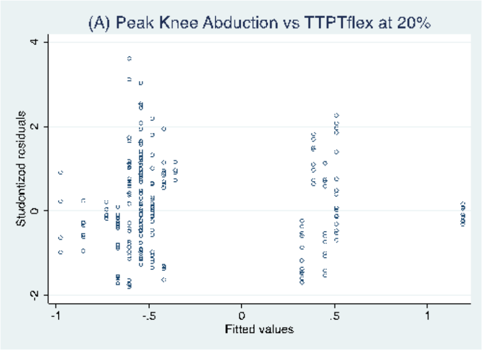
Peak Knee Abduction vs. TTDPMflex

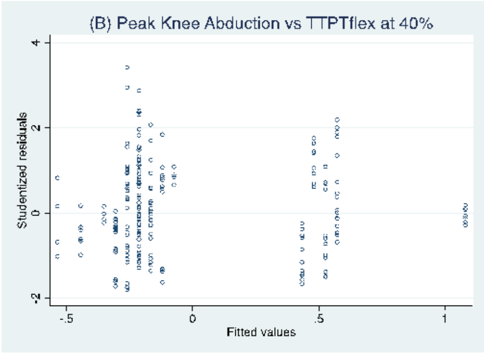


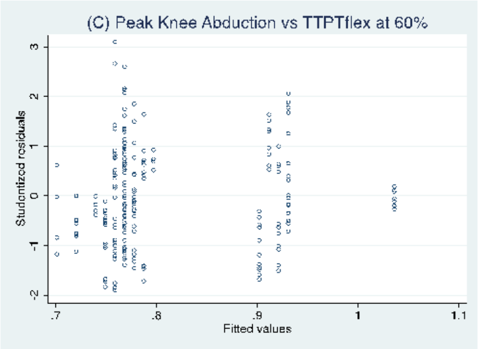
Peak Knee Abduction vs. TTPText

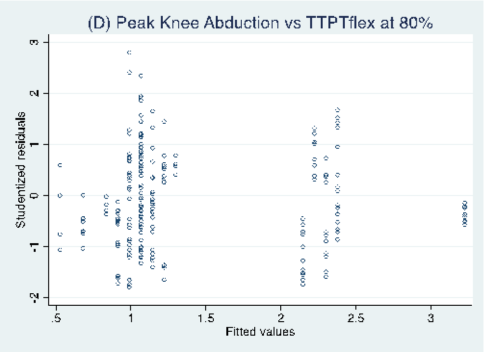


Peak Knee Abduction vs. TTPTflex

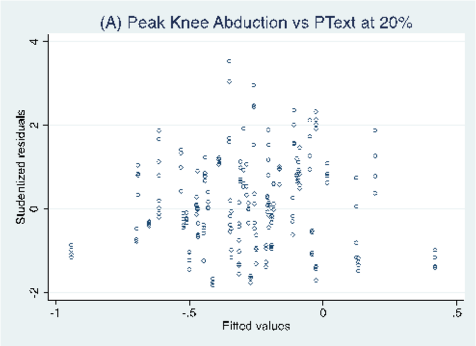


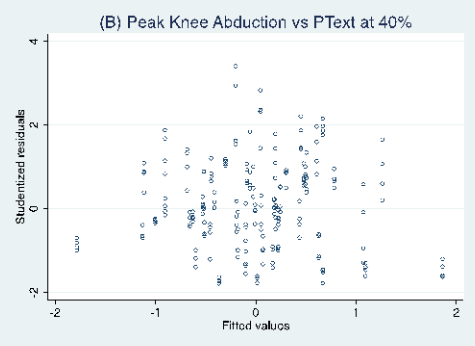


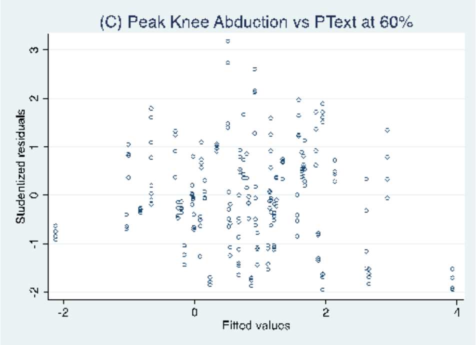


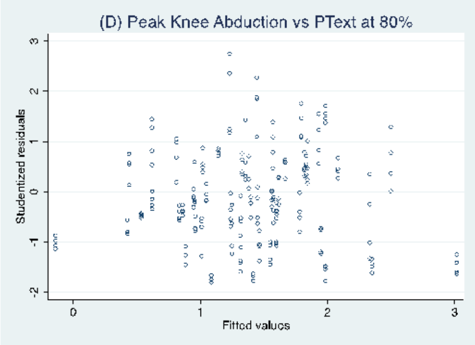


Peak Knee Abduction vs. PText

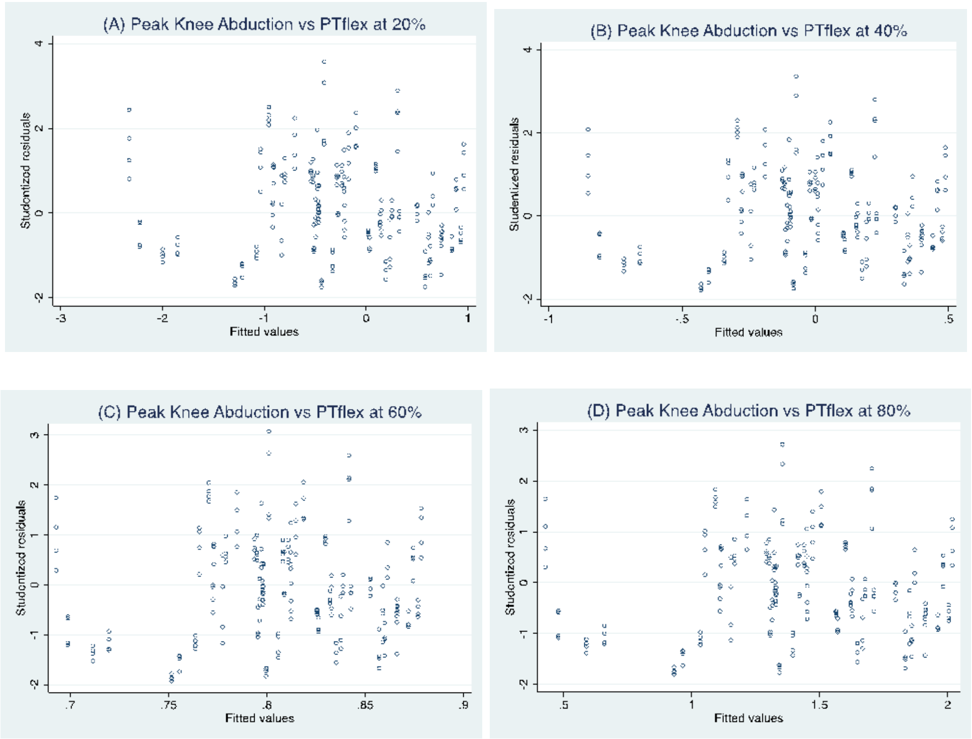






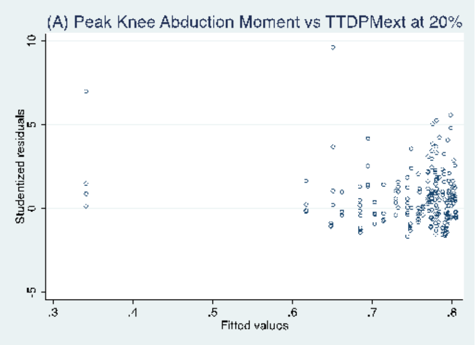


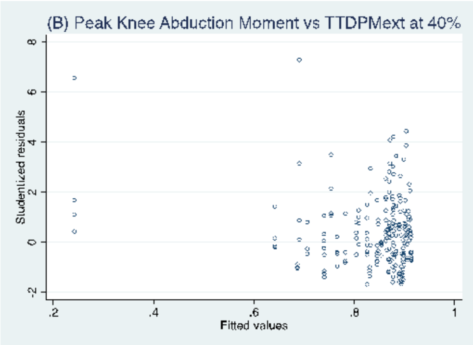
Peak Knee Abduction vs. PTflex

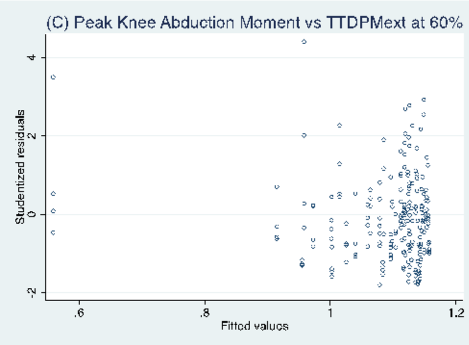


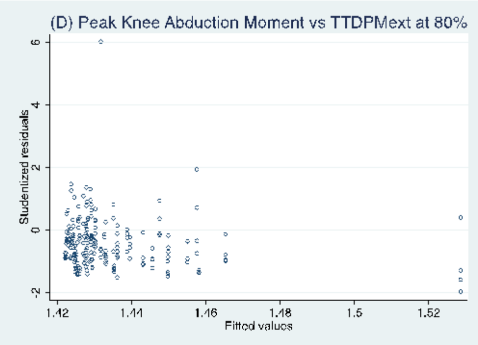
F.7 PEAK KNEE ABDUCTION MOMENT

Peak Knee Abduction Moment vs. TTDPMext

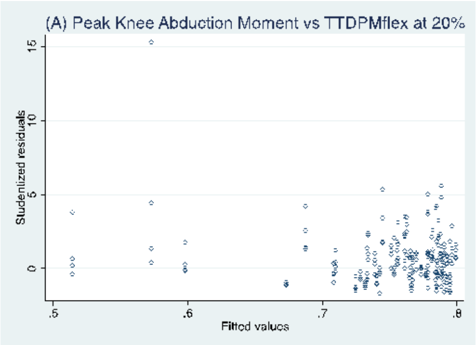


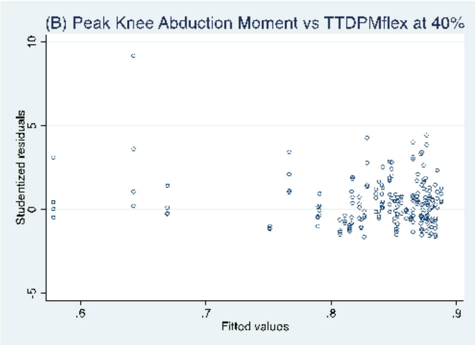


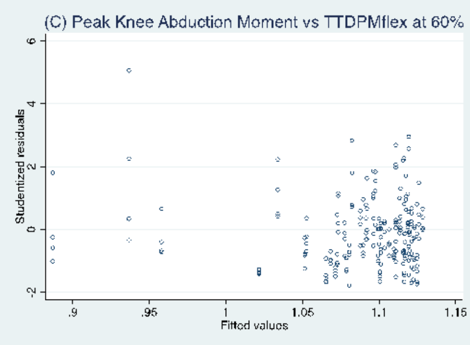


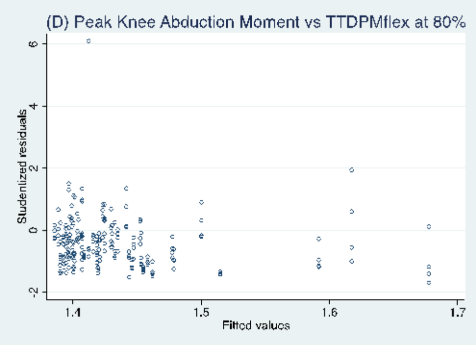


Peak Knee Abduction Moment vs. TTDPMflex

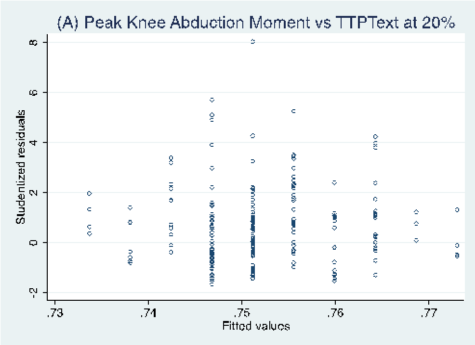


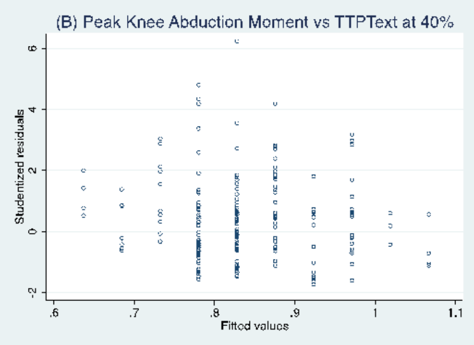


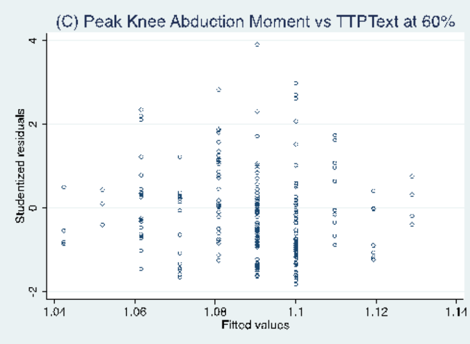


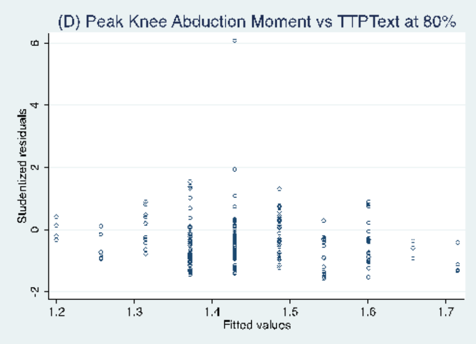


Peak Knee Abduction Moment vs. TTPText

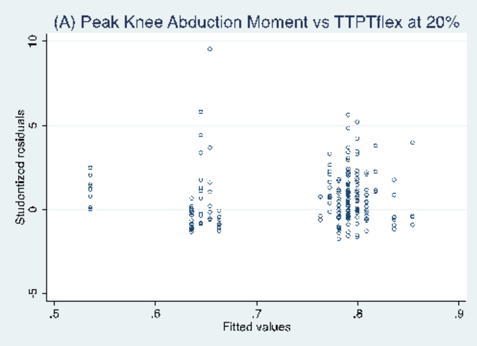


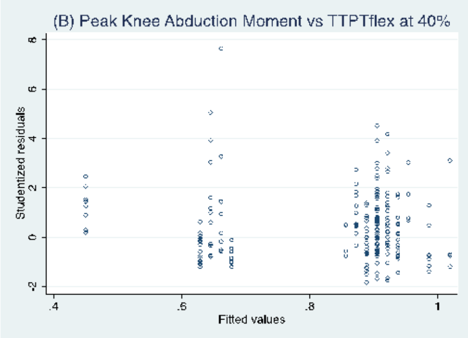


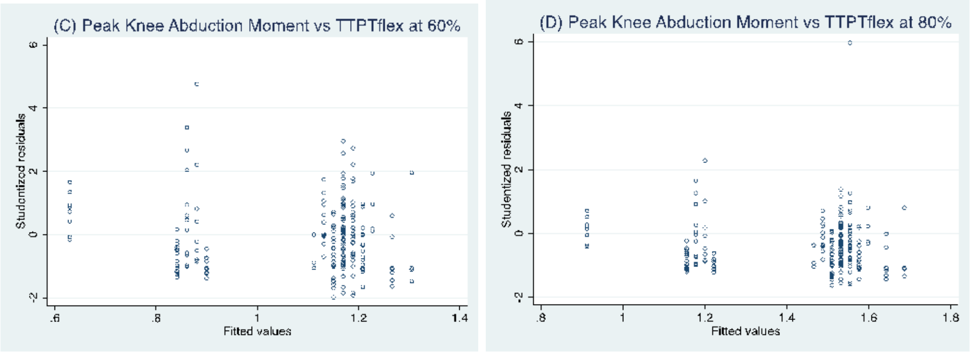




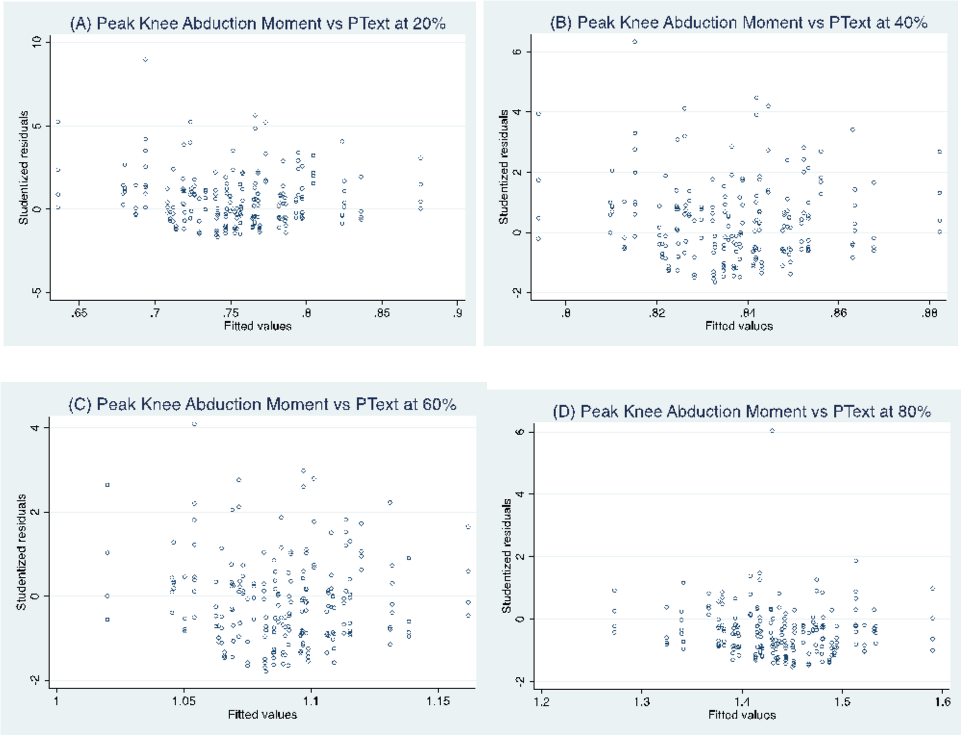
Peak Knee Abduction Moment vs. TTPTflex



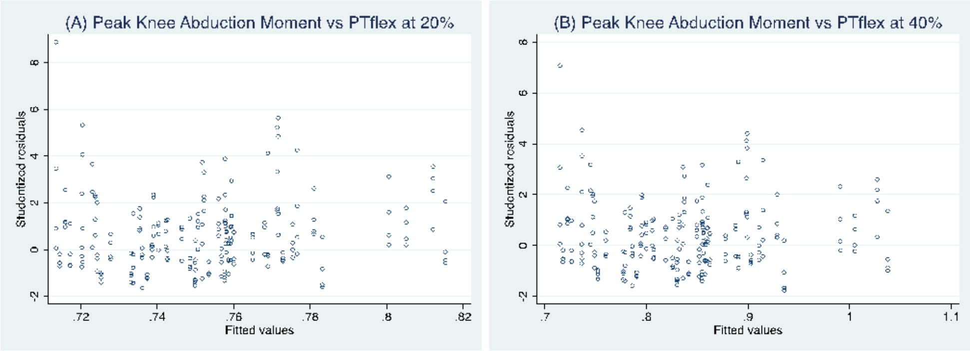


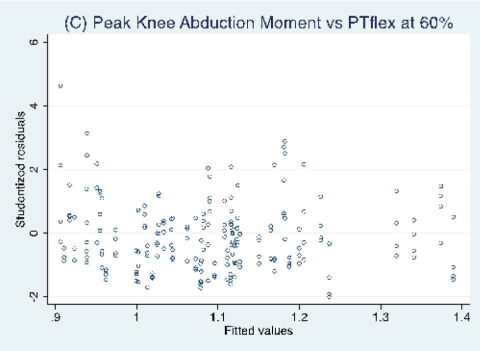


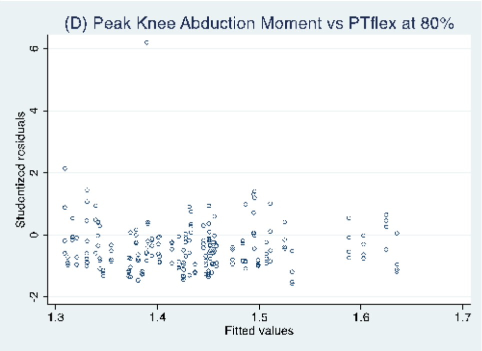
Peak Knee Abduction Moment vs. PText



Peak Knee Abduction Moment vs. PTflex

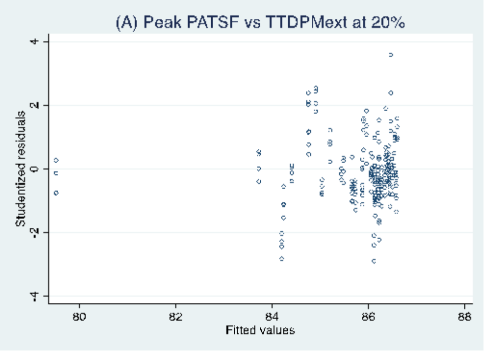


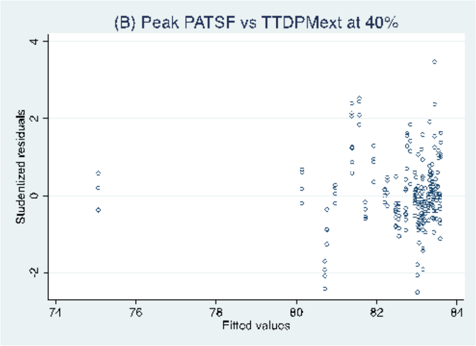

(C) Peak Knee Abduction Moment vs PTflex at 60%
Studentized residuals
-2
0
2
4
6
.9
1
1.1
1.2
1.3
1.4
Fitted values

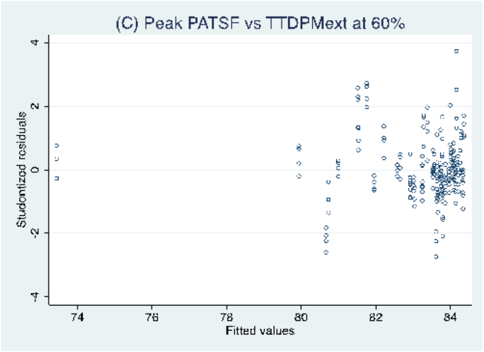

(D) Peak Knee Abduction Moment vs PTflex at 80%
Studentized residuals
-2
0
2
4
6
1.3
1.4
1.5
1.6
1.7
Fitted values

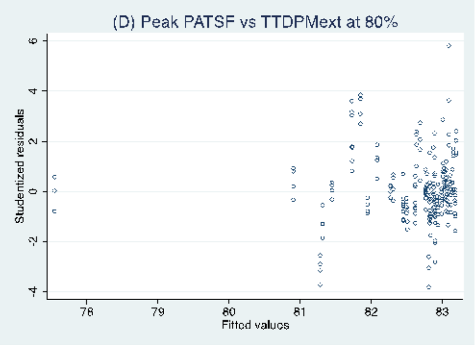
F.8 PEAK PROXIMAL ANTERIOR TIBIAL SHEAR FORCE

Peak Proximal Anterior Tibial Shear Force vs. TTDPMext

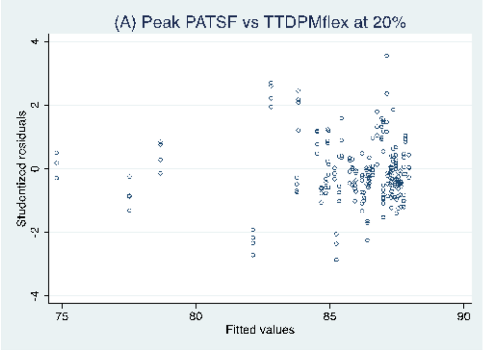


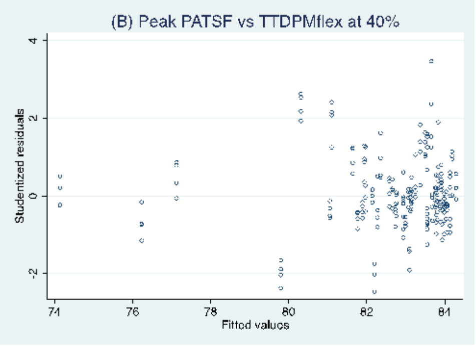




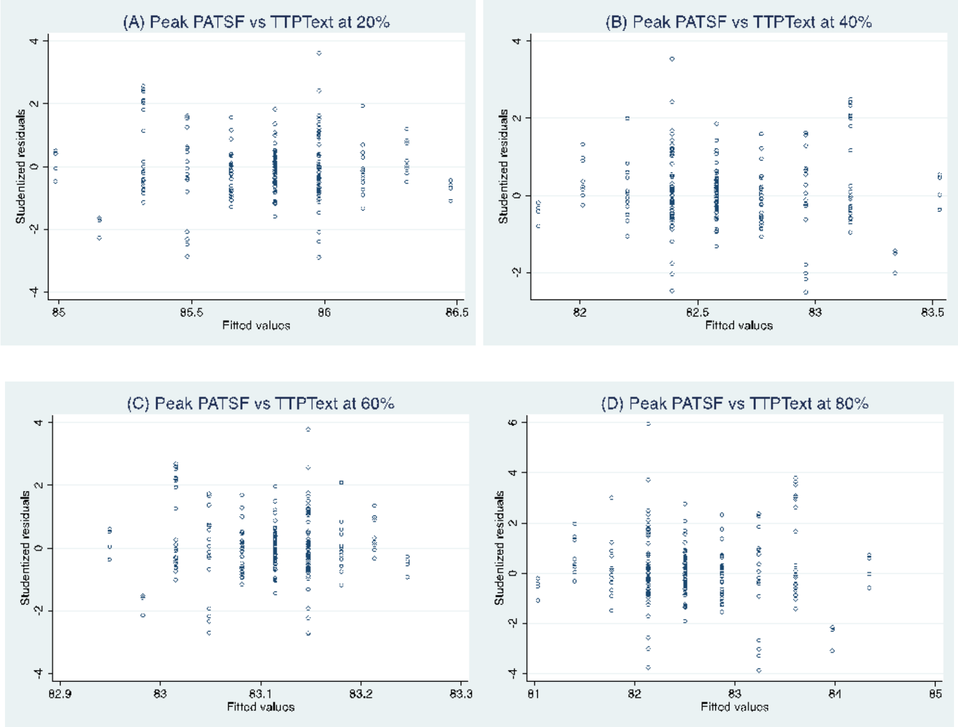


Peak Proximal Anterior Tibial Shear Force vs. TTDPMflex

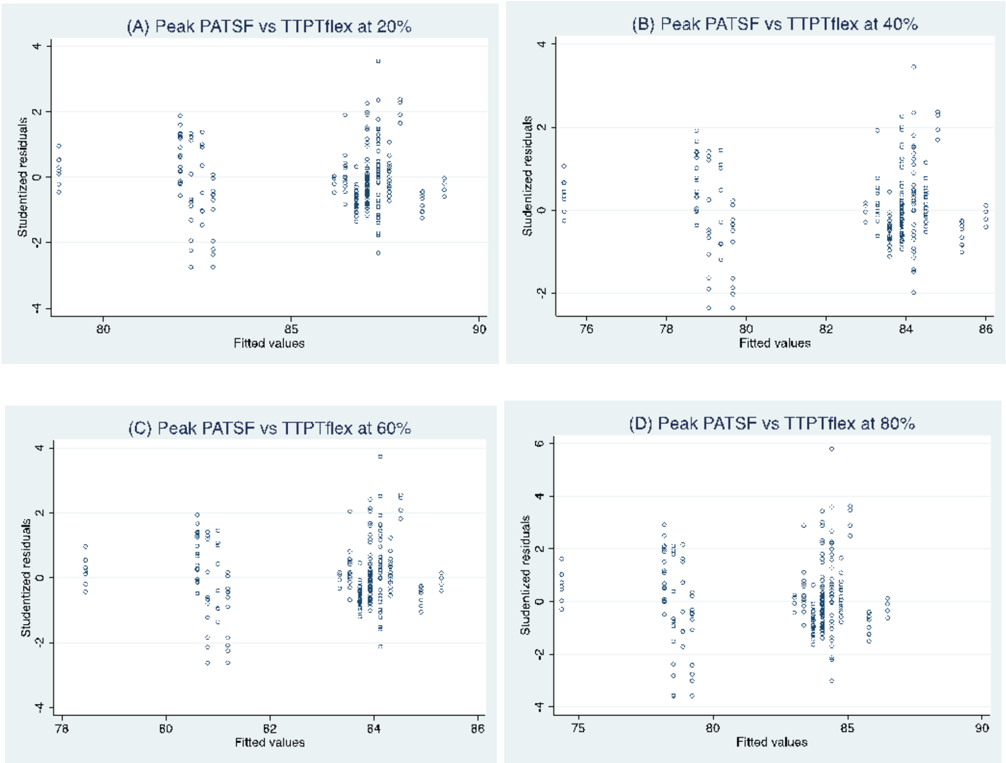




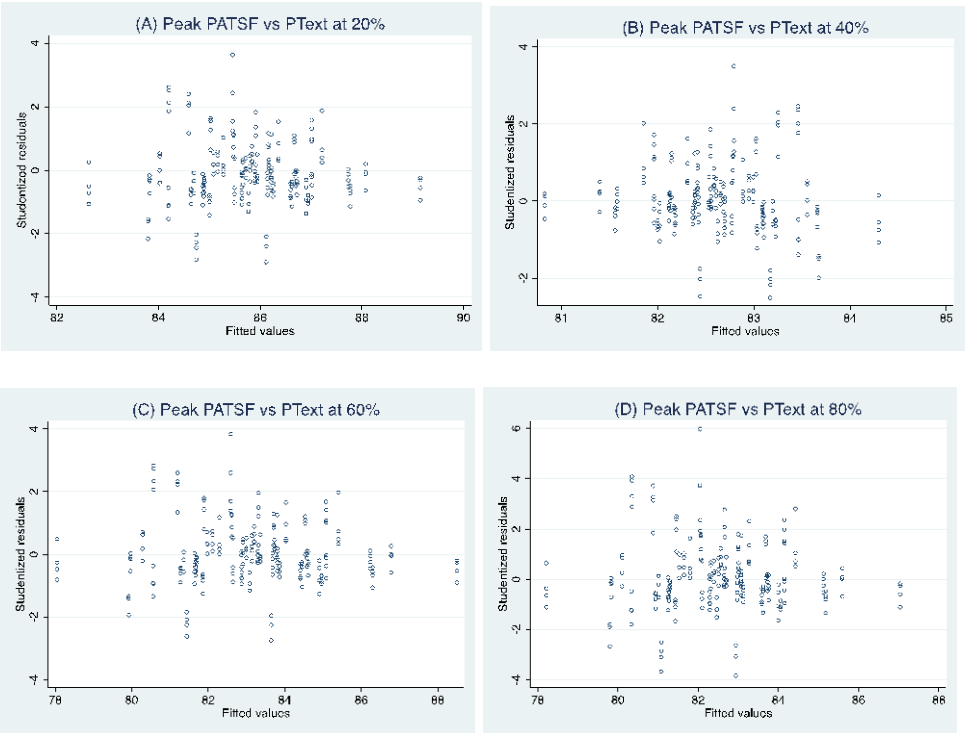
Peak Proximal Anterior Tibial Shear Force vs. TTPText



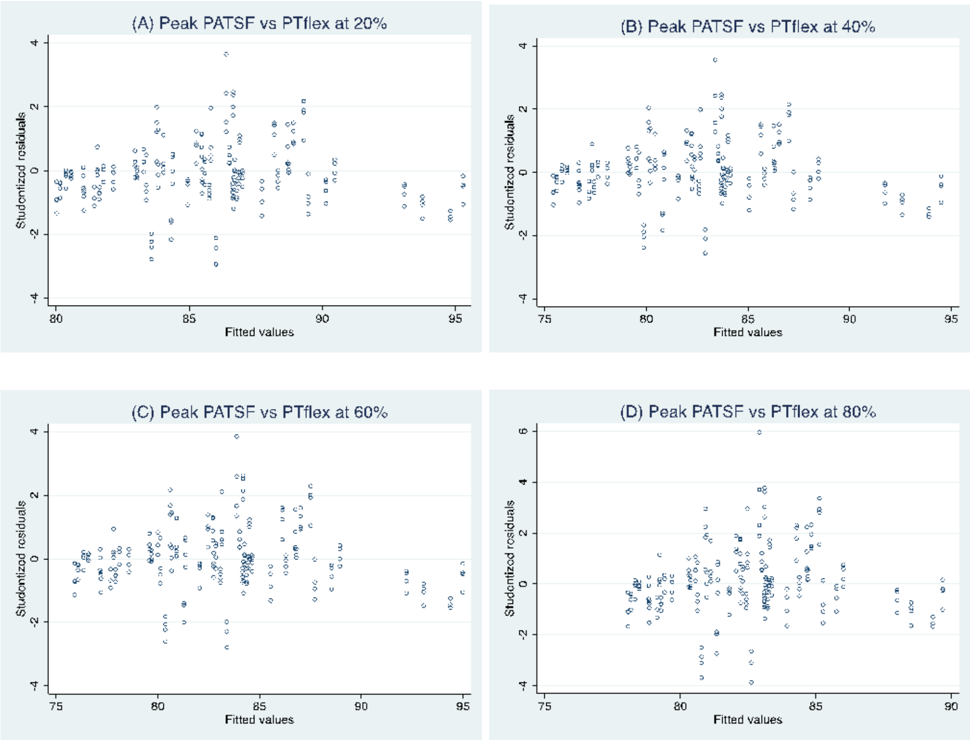
Peak Proximal Anterior Tibial Shear Force vs. TTPTflex



Peak Proximal Anterior Tibial Shear Force vs. PText



Peak Proximal Anterior Tibial Shear Force vs. PTflex


(A) Peak PATSF vs PTflex at 20%
(B) Peak PATSF vs PTflex at 40%
(C) Peak PATSF vs PTflex at 60%
(D) Peak PATSF vs PTflex at 80%
Studentized residuals
Fitted values
-4
-2
0
2
4
6
75
80
85
90
95

APPENDIX G

ROBUST SIMPLE REGRESSION RESULTS

G.1 PEAK VERTICAL GROUND REACTION FORCE BY JUMP DISTANCE

	n	R^2	Adj R^2	F	(df1, df2)	p-value
20% Body Height						
TTDPMext	53	0.0102	-0.0092	0.53	(1 , 51)	0.4720
TTDPMflex	53	0.0084	-0.0110	0.43	(1 , 51)	0.5140
TTPText	53	0.0024	-0.0171	0.12	(1 , 51)	0.7263
TTPTflex	53	0.0998	-0.1310	0.43	(1 , 51)	0.9216
TTPTflex[a]	53	-	-	-	-	-
PText	53	0.0011	-0.0185	0.05	(1 , 51)	0.8171
PTflex	53	0.0038	-0.0157	0.20	(1 , 51)	0.6602
40% Body Height						
TTDPMext	53	0.0030	-0.0166	0.15	(1 , 51)	0.6986
TTDPMflex	53	0.0000	-0.0196	0.00	(1 , 51)	0.9947
TTPText	53	0.0009	-0.0187	0.04	(1 , 51)	0.8349
TTPTflex	53	0.1051	-0.1243	0.46	(1 , 51)	0.9066
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
60% Body Height						
TTDPMext	53	0.0103	-0.0091	0.53	(1 , 51)	0.4687
TTDPMflex	53	0.0195	0.0003	1.01	(1 , 51)	0.3185
TTPText	53	0.0049	-0.0146	0.25	(1 , 51)	0.6167
TTPTflex	53	0.2036	-0.0006	1.00	(1 , 51)	0.4625
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
80% Body Height						
TTDPMext	50	0.0109	-0.0097	0.53	(1 , 49)	0.4696
TTDPMflex	50	0.0072	-0.0130	0.36	(1 , 49)	0.5531
TTPText	50	0.0004	0.0200	0.02	(1 , 49)	0.8918
TTPTflex	50	0.2498	0.047	1.23	(1 , 49)	0.3038
TTPTflex[a]	50	-	-	-	-	-
PText	50	-	-	-	-	-
PTflex	50	-	-	-	-	-

[a] Grouped variable (three groups)

Blank cells did not require robust regression analysis

G.2 PEAK ANTERIOR-POSTERIOR GROUND REACTION FORCE BY JUMP DISTANCE

	n	R^2	Adj R^2	F	(df1, df2)	p-value
20% Body Height						
TTDPMext	53	0.0095	-0.0099	0.49	(1 , 51)	0.4876
TTDPMflex	53	0.0003	-0.0193	0.02	(1 , 51)	0.9011
TTPText	53	0.0004	-0.0192	0.02	(1 , 51)	0.8925
TTPTflex	53	0.0049	-0.0146	0.25	(1 , 51)	0.6187
TTPTflex[a]	53	-	-	-	-	-
PText	53	0.0006	-0.0190	0.03	(1 , 51)	0.8635
PTflex	53	0.0000	-0.0196	0.00	(1 , 51)	0.9814
40% Body Height						
TTDPMext	53	0.0121	-0.0073	0.62	(1 , 51)	0.4333
TTDPMflex	53	0.0079	-0.0116	0.40	(1 , 51)	0.5281
TTPText	53	0.0000	-0.0196	0.00	(1 , 51)	0.9831
TTPTflex	53	0.0024	-0.0171	0.12	(1 , 51)	0.7266
TTPTflex[a]	53	-	-	-	-	-
PText	53	0.0006	-0.0190	0.03	(1 , 51)	0.8578
PTflex	53	0.0046	-0.0150	0.23	(1 , 51)	0.6307
60% Body Height						
TTDPMext	53	0.0314	0.0124	1.65	(1 , 51)	0.2043
TTDPMflex	53	0.0293	0.0103	1.54	(1 , 51)	0.2204
TTPText	53	-	-	-	-	-
TTPTflex	53	0.0086	-0.0108	0.44	(1 , 51)	0.5087
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
80% Body Height						
TTDPMext	50	0.0230	0.0031	1.16	(1 , 49)	0.2876
TTDPMflex	50	0.0647	0.0456	3.39	(1 , 49)	0.0716
TTPText	50	-	-	-	-	-
TTPTflex	50	-	-	-	-	-
TTPTflex[a]	50	-	-	-	-	-
PText	50	-	-	-	-	-
PTflex	50	-	-	-	-	-

[a] Grouped variable (three groups)

Blank cells did not require robust regression analysis

G.3 KNEE FLEXION AT INITIAL CONTACT BY JUMP DISTANCE

	n	R^2	Adj R^2	F	(df1, df2)	p-value
20% Body Height						
TTDPMext	53	0.0794	0.0610	4.31	(1 , 51)	0.0430
TTDPMflex	53	0.1735	0.1569	10.49	(1 , 51)	0.0021
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
40% Body Height						
TTDPMext	53	0.0085	-0.0109	0.44	(1 , 51)	0.5102
TTDPMflex	53	-	-	-	-	-
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
60% Body Height						
TTDPMext	53	0.0546	0.0357	2.89	(1 , 51)	0.0955
TTDPMflex	53	0.0727	0.0546	4.00	(1 , 51)	0.0508
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
80% Body Height						
TTDPMext	50	0.1539	0.1363	8.73	(1 , 49)	0.0048
TTDPMflex	50	0.1460	0.1286	8.38	(1 , 49)	0.0056
TTPText	50	-	-	-	-	-
TTPTflex	50	-	-	-	-	-
TTPTflex[a]	50	-	-	-	-	-
PText	50	-	-	-	-	-
PTflex	50	-	-	-	-	-

[a] Grouped variable (three groups)

Blank cells did not require robust regression analysis

G.4 PEAK KNEE ABDUCTION MOMENT BY JUMP DISTANCE

	n	R^2	Adj R^2	F	(df1, df2)	p-value
20% Body Height						
TTDPMext	53	0.0207	0.0015	1.08	(1 , 51)	0.3035
TTDPMflex	53	0.0128	-0.0066	0.66	(1 , 51)	0.4206
TTPText	53	0.0022	-0.0174	0.11	(1 , 51)	0.7403
TTPTflex	53	0.0236	0.0044	1.23	(1 , 51)	0.2724
TTPTflex[a]	53	-	-	-	-	-
PText	53	0.0050	-0.0145	0.26	(1 , 51)	0.6142
PTflex	53	0.0098	-0.0096	0.50	(1 , 51)	0.4808
40% Body Height						
TTDPMext	53	0.0371	0.0183	1.97	(1 , 51)	0.1668
TTDPMflex	53	0.0124	-0.0070	0.64	(1 , 51)	0.4274
TTPText	53	-	-	-	-	-
TTPTflex	53	0.0610	0.0426	3.32	(1 , 51)	0.0745
TTPTflex[a]	53	-	-	-	-	-
PText	53	0.0000	-0.0196	0.00	(1 , 51)	0.9847
PTflex	53	0.0252	0.0061	1.32	(1 , 51)	0.2563
60% Body Height						
TTDPMext	53	0.0182	-0.0011	0.94	(1 , 51)	0.3358
TTDPMflex	53	0.0069	-0.0126	0.35	(1 , 51)	0.5541
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
80% Body Height						
TTDPMext	50	0.0009	-0.0195	0.04	(1 , 49)	0.8387
TTDPMflex	50	-	-	-	-	-
TTPText	50	-	-	-	-	-
TTPTflex	50	-	-	-	-	-
TTPTflex[a]	50	-	-	-	-	-
PText	50	-	-	-	-	-
PTflex	50	-	-	-	-	-

[a]Grouped variable (three groups)

Blank cells did not require robust regression analysis

G.5 PEAK KNEE ABDUCTION MOMENT (SQUARE ROOT) BY JUMP DISTANCE

	n	R^2	Adj R^2	F	(df1, df2)	p-value
20% Body Height						
TTDPMext	53	0.0182	-0.0010	0.95	(1 , 51)	0.3349
TTDPMflex	53	0.0111	-0.0083	0.57	(1 , 51)	0.4526
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
40% Body Height						
TTDPMext	53	0.0349	0.0156	1.81	(1 , 51)	0.1851
TTDPMflex	53	0.0111	-0.0087	0.56	(1 , 51)	0.457
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
60% Body Height						
TTDPMext	53	0.0182	-0.0010	0.95	(1 , 51)	0.335
TTDPMflex	53	0.0089	-0.0105	0.46	(1 , 51)	0.5008
TTPText	53	-	-	-	-	-
TTPTflex	53	-	-	-	-	-
TTPTflex[a]	53	-	-	-	-	-
PText	53	-	-	-	-	-
PTflex	53	-	-	-	-	-
80% Body Height						
TTDPMext	50	0.0011	-0.0193	0.05	(1 , 49)	0.8201
TTDPMflex	50	0.0060	-0.0143	0.29	(1 , 49)	0.5904
TTPText	50	-	-	-	-	-
TTPTflex	50	-	-	-	-	-
TTPTflex[a]	50	-	-	-	-	-
PText	50	-	-	-	-	-
PTflex	50	-	-	-	-	-

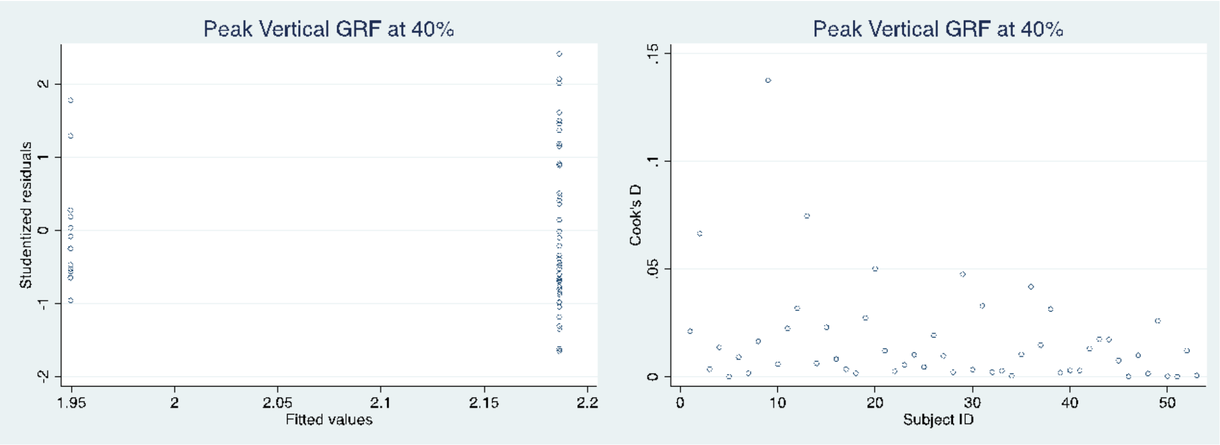
[a]Grouped variable (three groups)

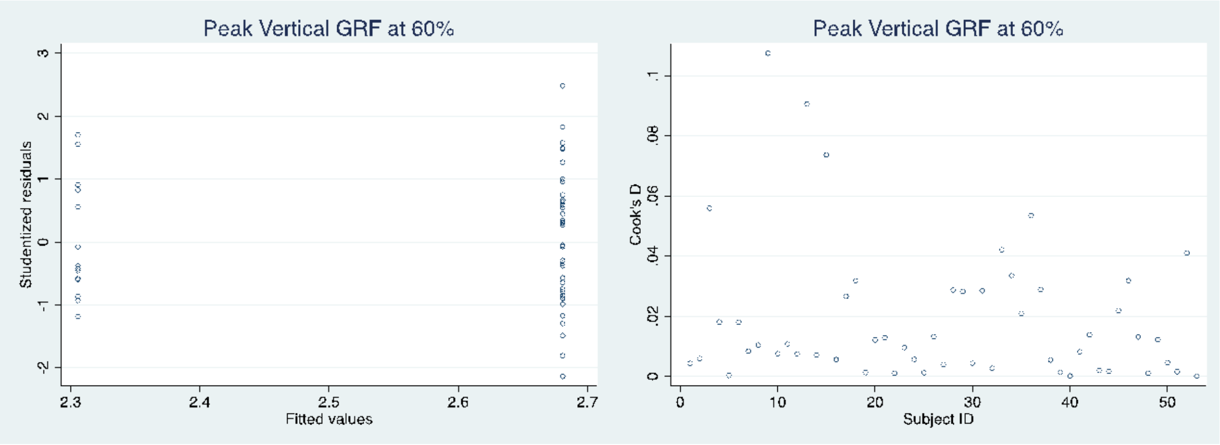
Blank cells did not require robust regression analysis

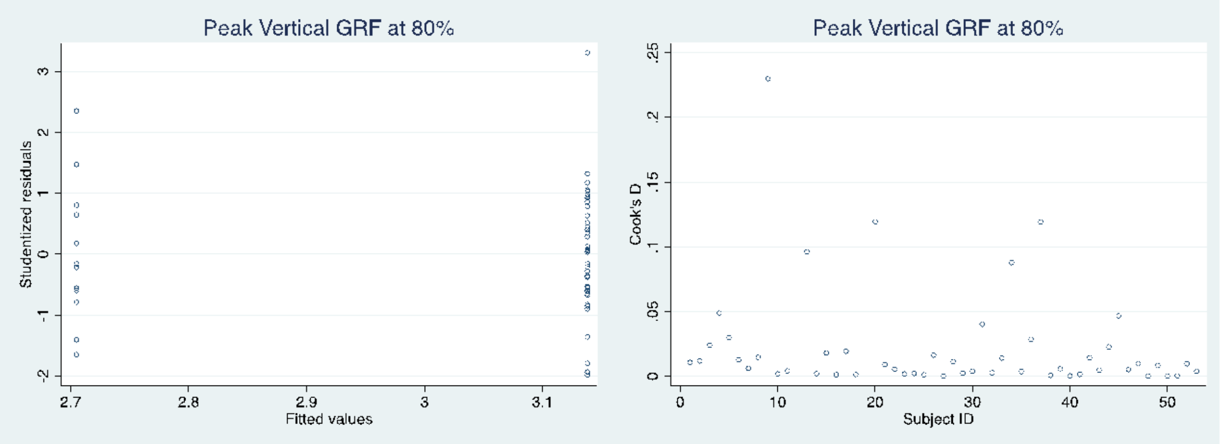
APPENDIX H

MULTIPLE LINEAR REGRESSION: JACKKNIFE RESIDUALS VS. FITTED VALUES

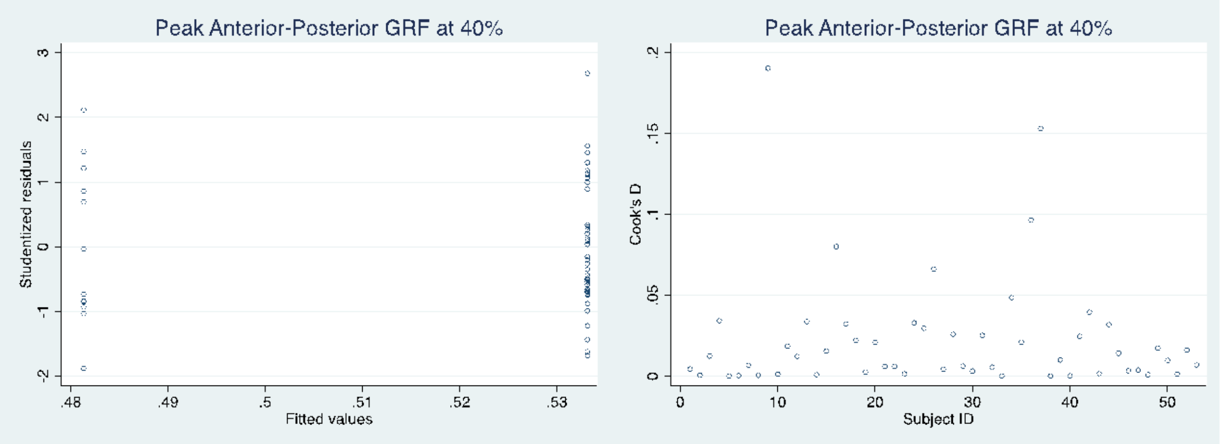
H.1 VERTICAL GROUND REACTION FORCE

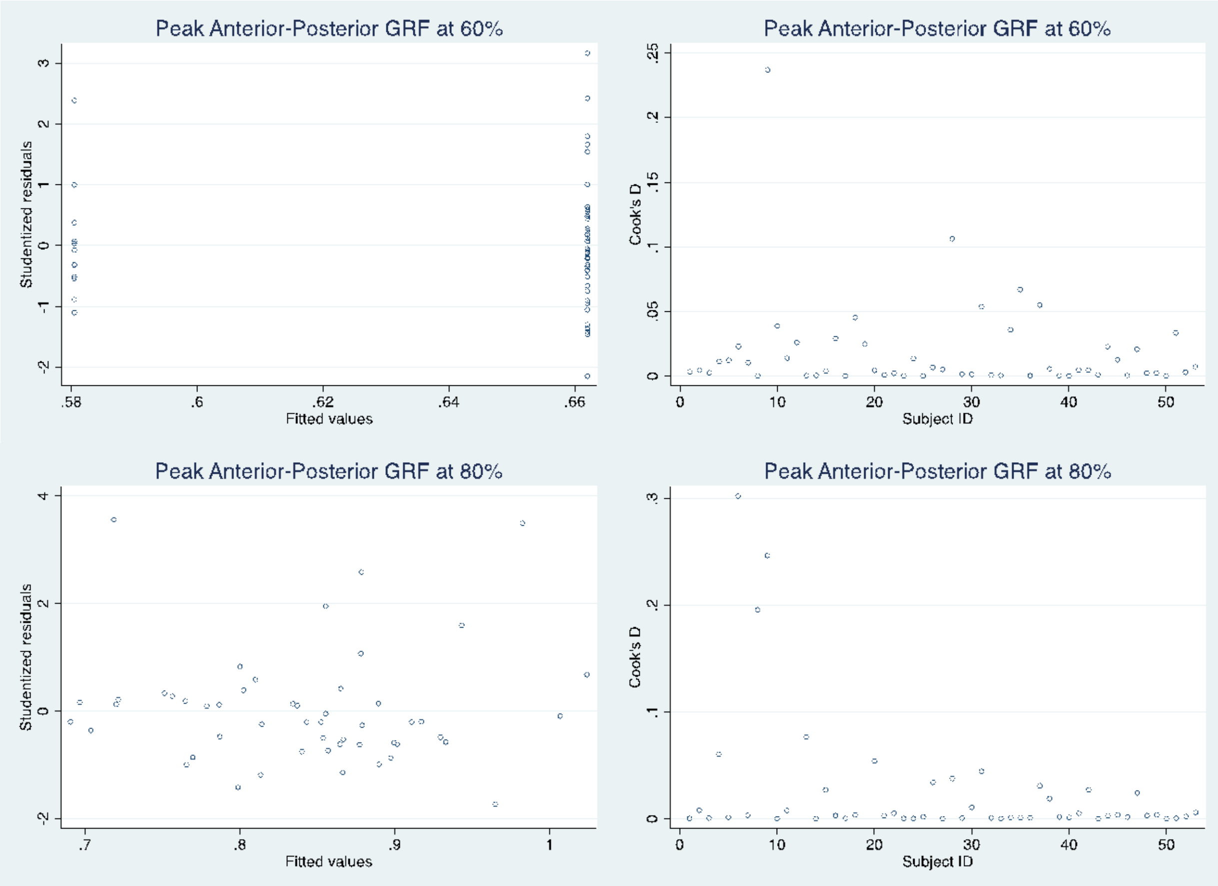




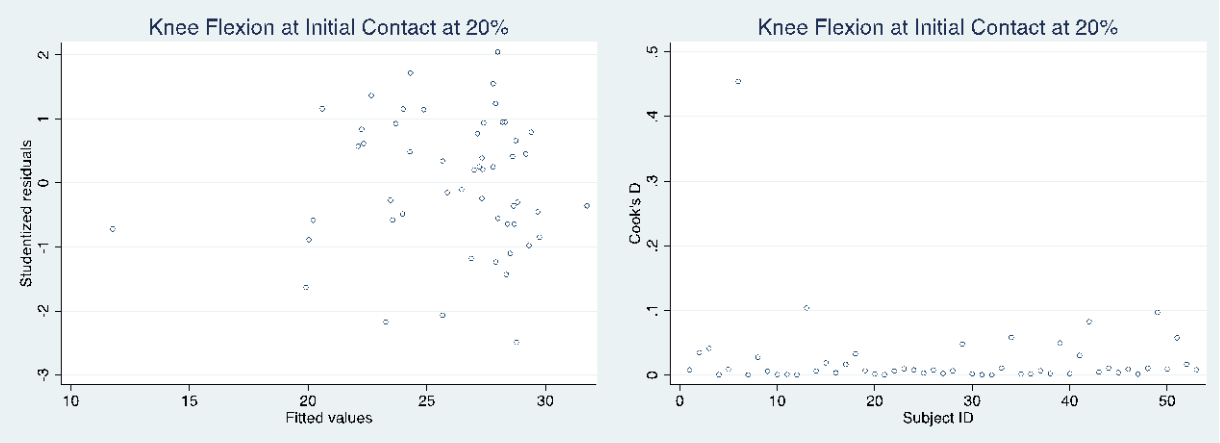


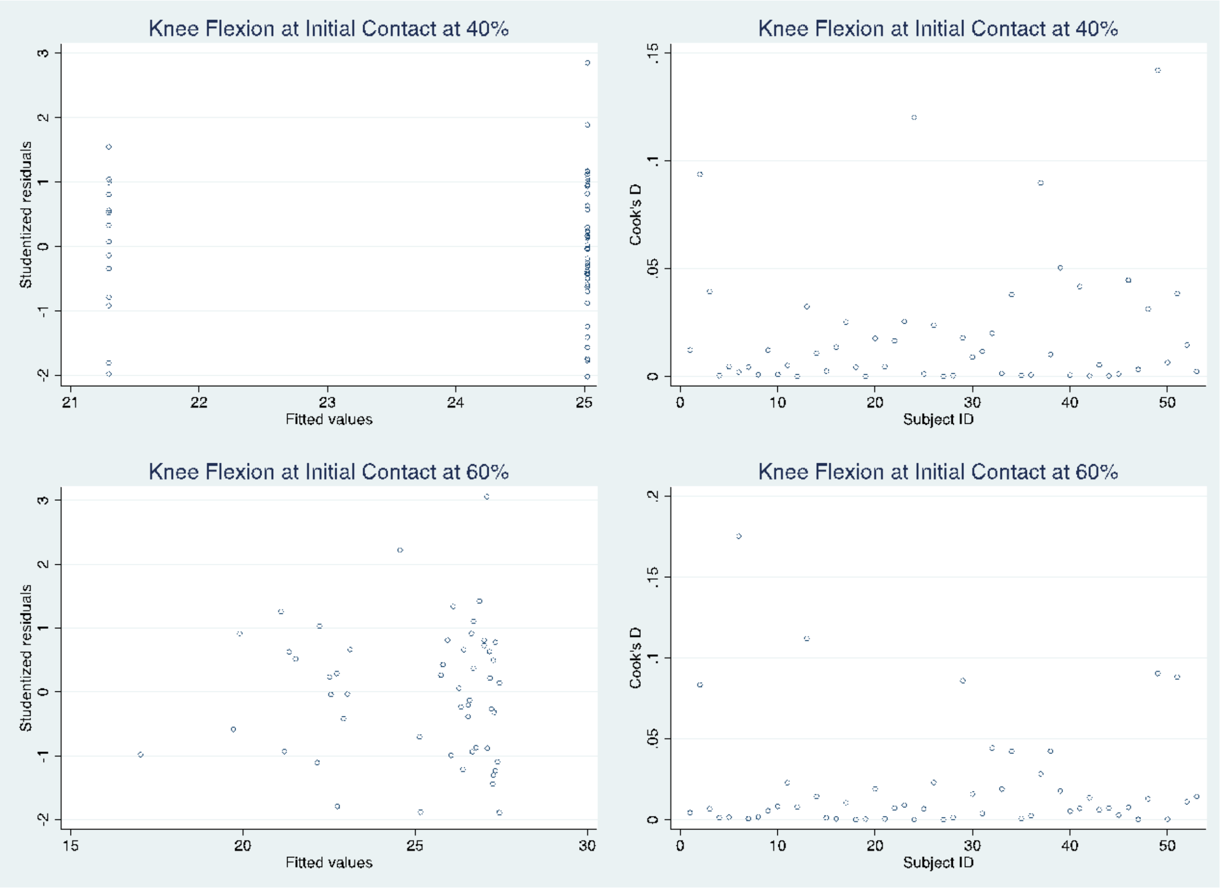
H.2 ANTERIOR-POSTERIOR GROUND REACTION FORCE

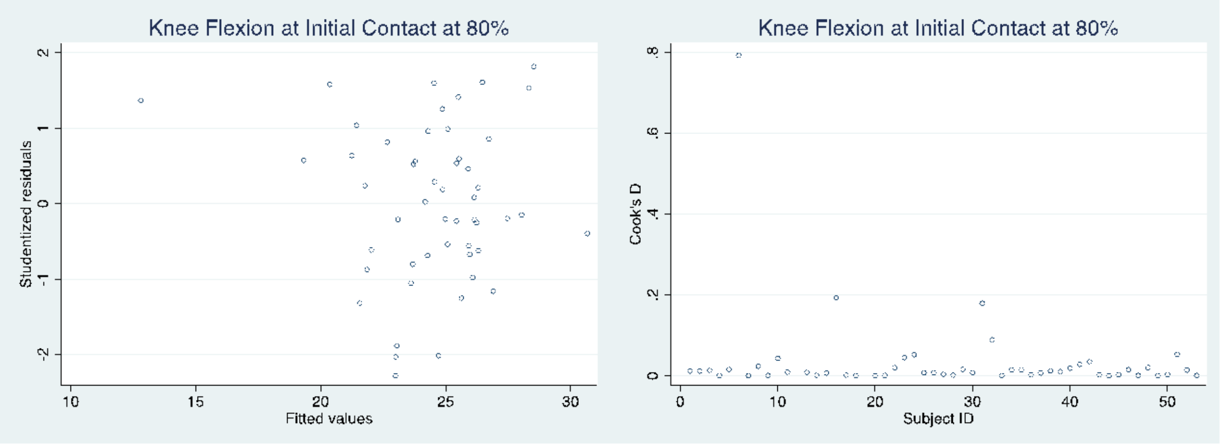




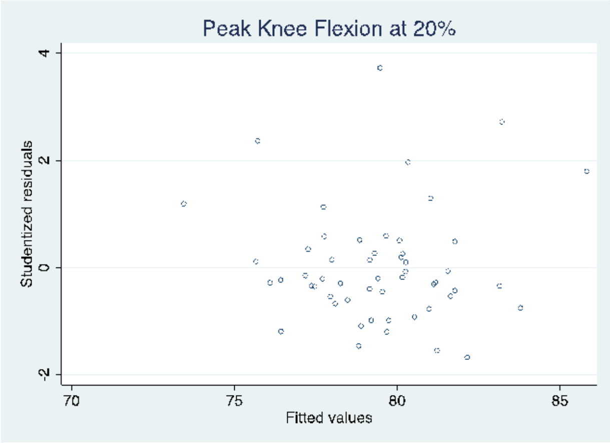
H.3 KNEE FLEXION AT INITIAL CONTACT

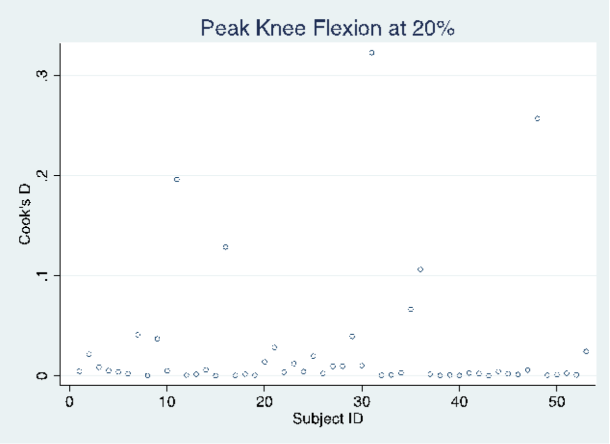


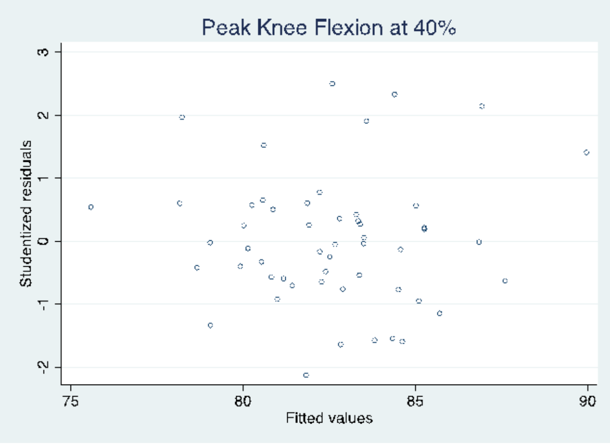

Knee Flexion at Initial Contact at 40%
Studentized residuals
-2
-1
0
1
2
3
Fitted values
21
22
23
24
25
Knee Flexion at Initial Contact at 40%
Cook's D
0
.05
.1
.15
Subject ID
0
10
20
30
40
50
Knee Flexion at Initial Contact at 60%
Studentized residuals
-2
-1
0
1
2
3
Fitted values
15
20
25
30
Knee Flexion at Initial Contact at 60%
Cook's D
0
.05
.1
.15
.2
Subject ID
0
10
20
30
40
50

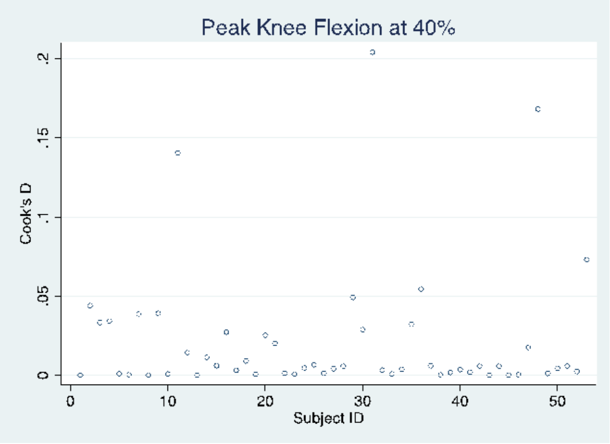

Knee Flexion at Initial Contact at 80%
Studentized residuals
-2
-1
0
1
2
Fitted values
10
15
20
25
30
Knee Flexion at Initial Contact at 80%
Cook's D
0
.2
.4
.6
.8
Subject ID
0
10
20
30
40
50

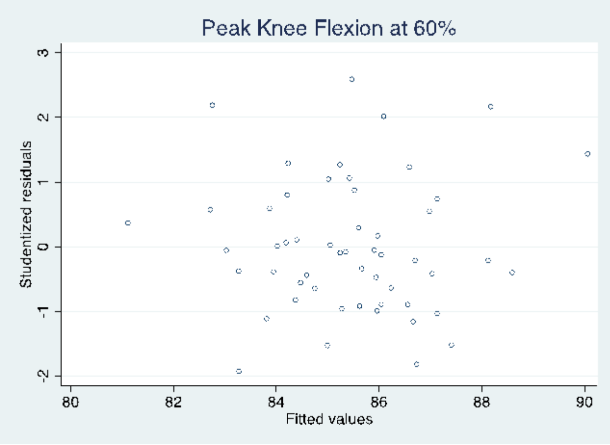
H.4 PEAK KNEE FLEXION

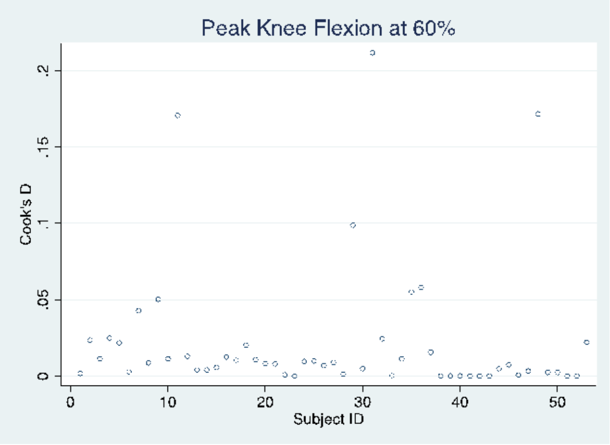


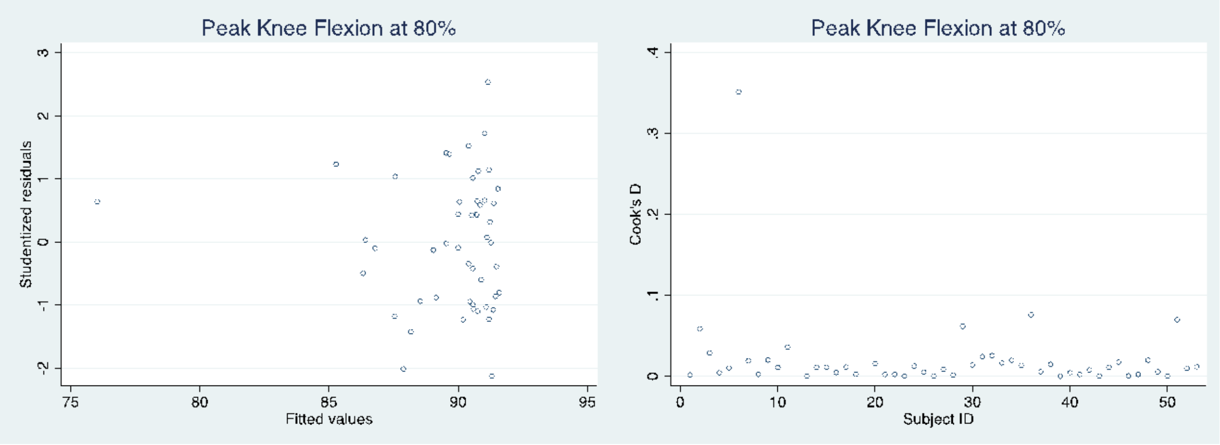




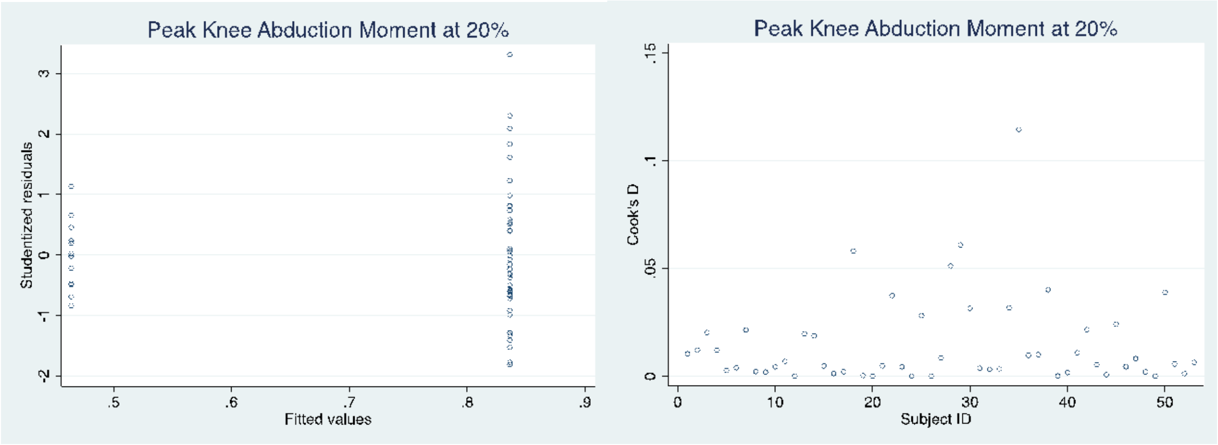


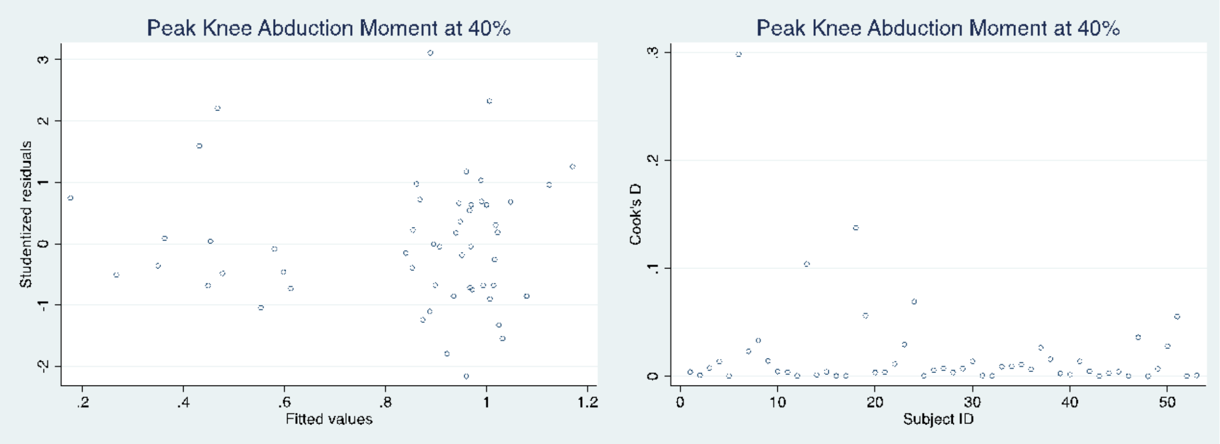




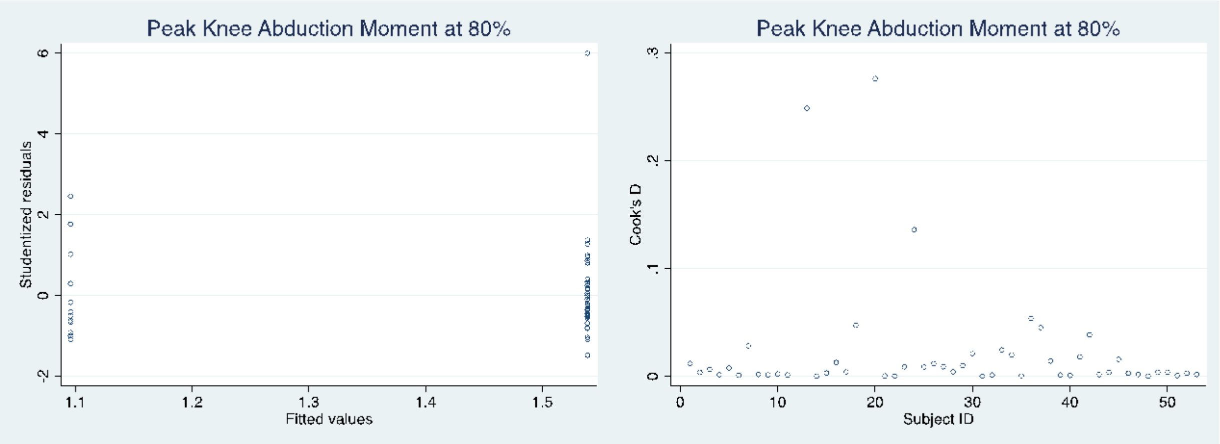


H.5 PEAK KNEE ABDUCTION MOMENT

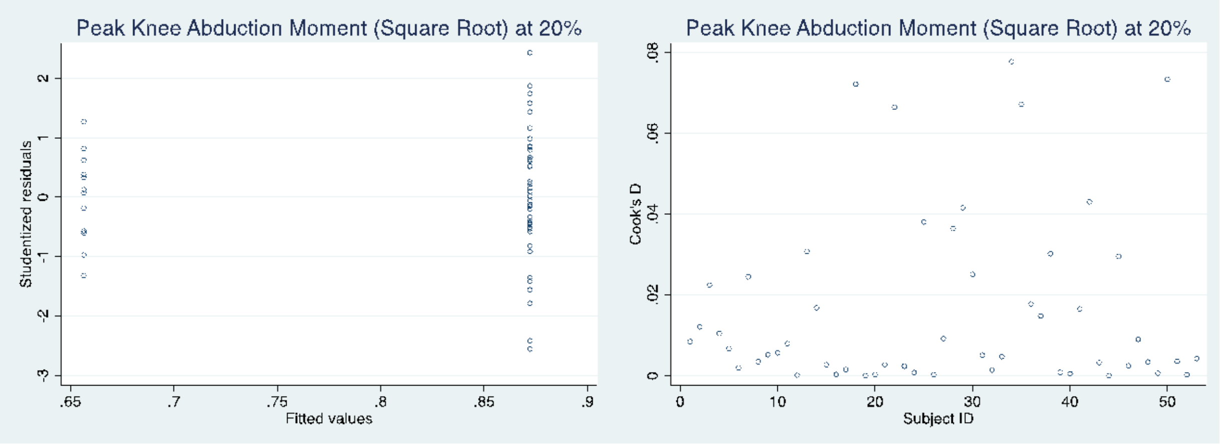


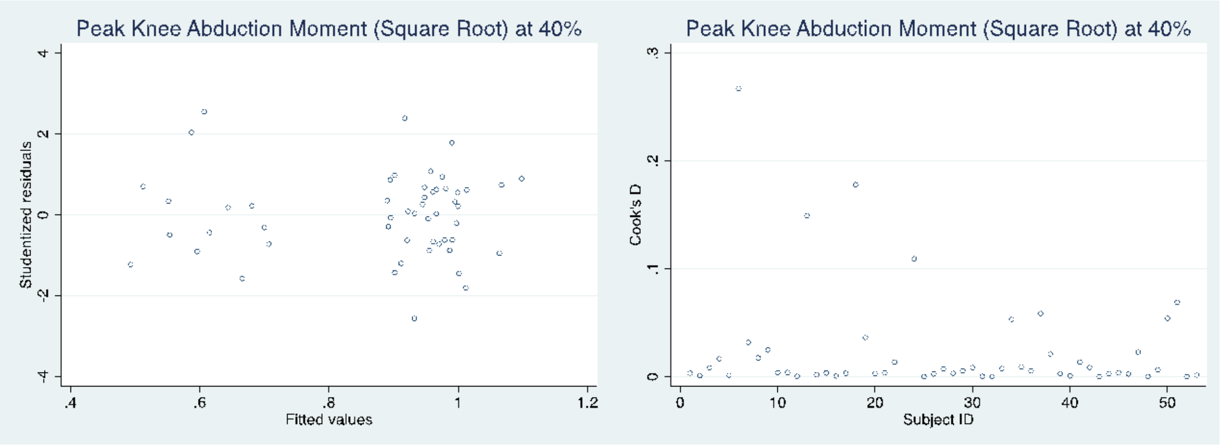

Peak Knee Abduction Moment at 40%
Studentized residuals
-2
-1
0
1
2
3
.2
.4
.6
.8
1
1.2
Fitted values
Peak Knee Abduction Moment at 40%
Cook's D
0
.1
.2
.3
0
10
20
30
40
50
Subject ID

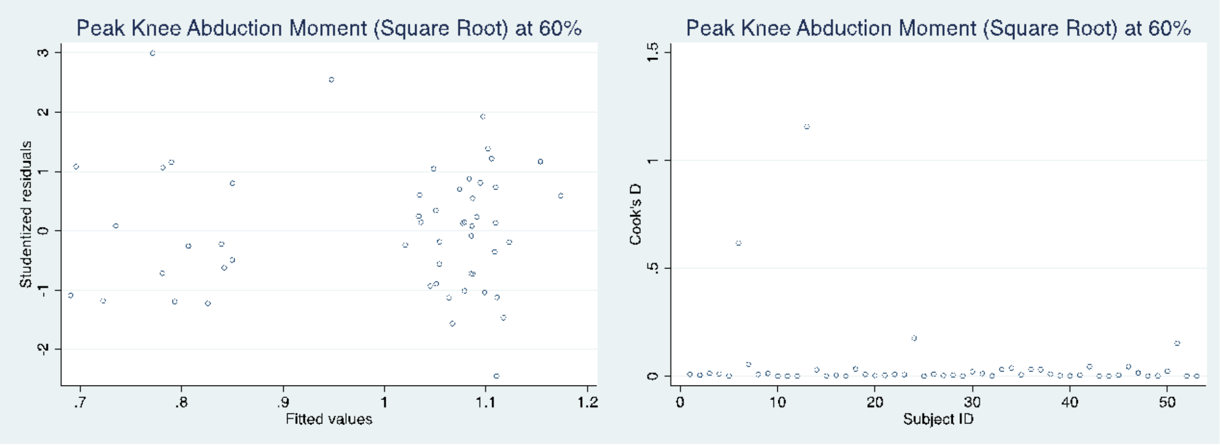

Peak Knee Abduction Moment at 60%
Studentized residuals
-2
-1
0
1
2
3
.4
.6
.8
1
1.2
1.4
Fitted values
Peak Knee Abduction Moment at 60%
Cook's D
0
.2
.4
.6
.8
1
0
10
20
30
40
50
Subject ID

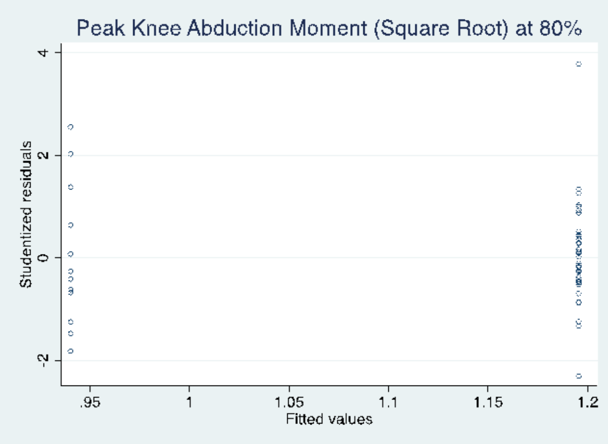

Peak Knee Abduction Moment at 80%
Studentized residuals
-2
0
2
4
6
1.1
1.2
1.3
1.4
1.5
Fitted values
Peak Knee Abduction Moment at 80%
Cook's D
0
.1
.2
.3
0
10
20
30
40
50
Subject ID

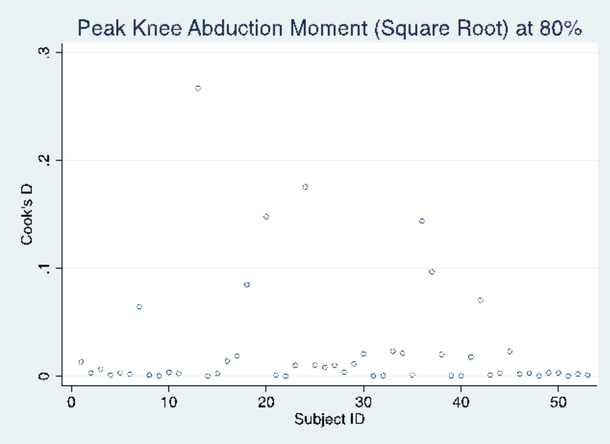
H.6 PEAK KNEE ABDUCTION MOMENT (SQUARE ROOT)



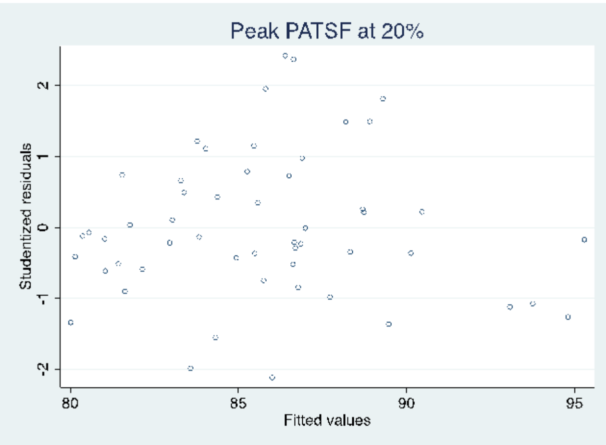


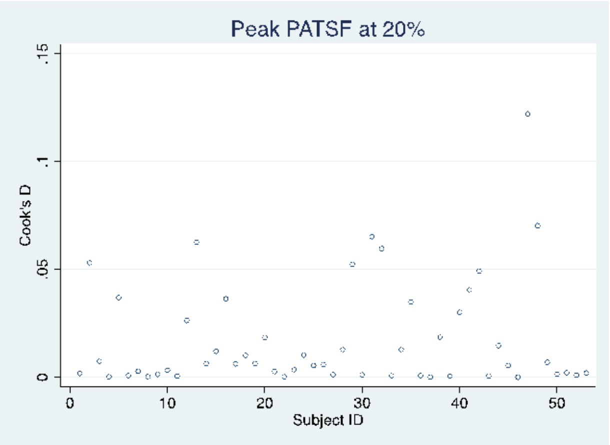


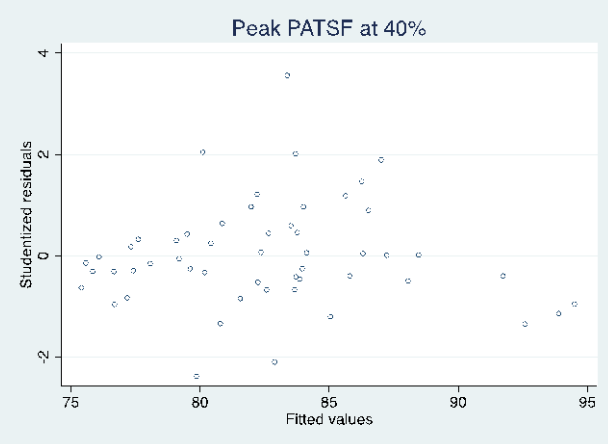


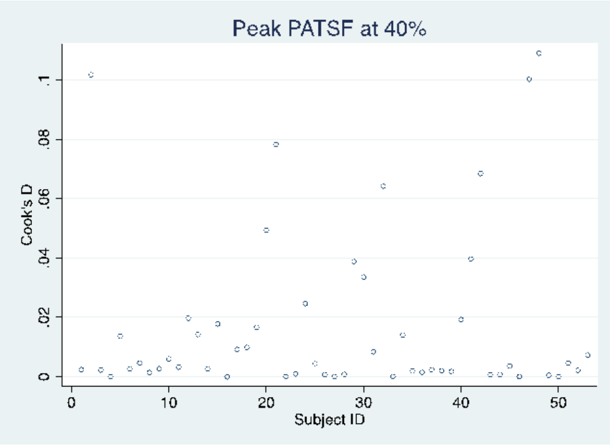


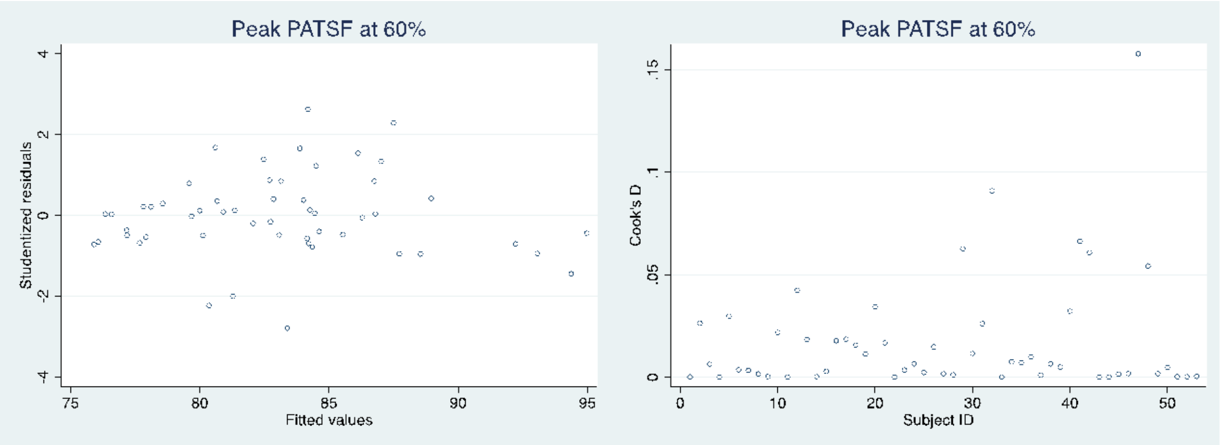
H.7 PEAK PROXIMAL ANTERIOR TIBIAL SHEAR FORCE










Peak PATSF at 60%
Studentized residuals
4
2
0
-2
-4
75
80
85
90
95
Fitted values
Peak PATSF at 60%
Cook's D
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.05
0
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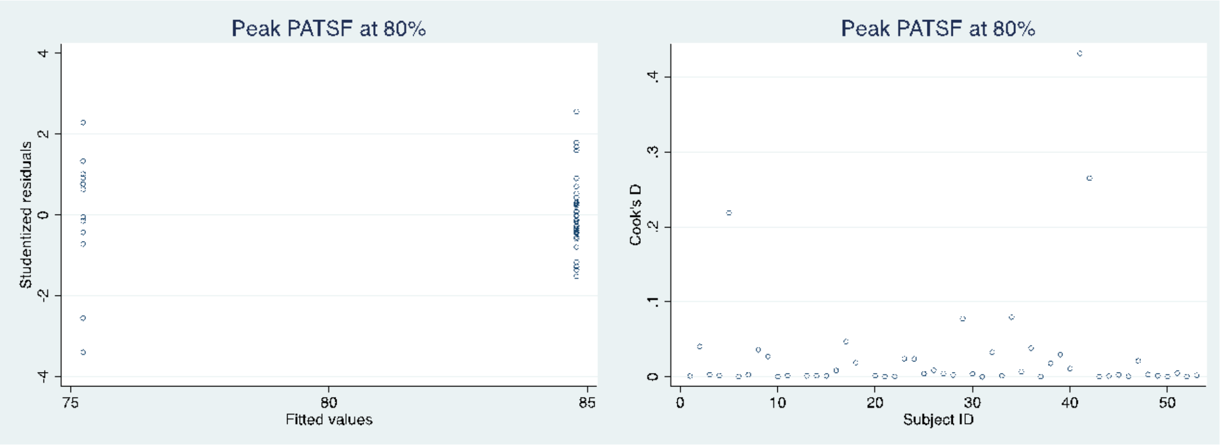
Subject ID


Peak PATSF at 80%
Studentized residuals
4
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Fitted values
Peak PATSF at 80%
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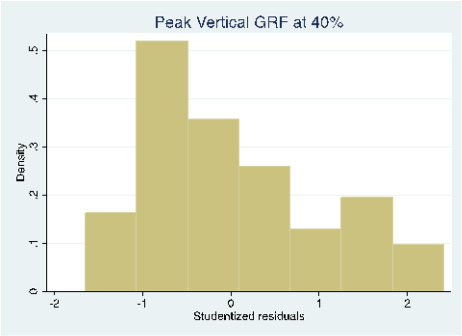
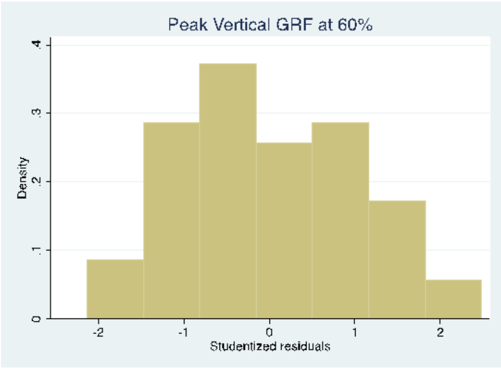
Subject ID

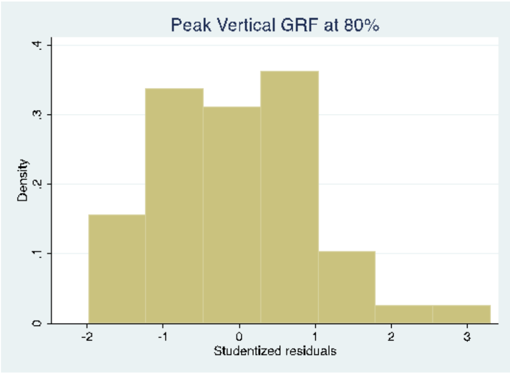
APPENDIX I

MULTIPLE LINEAR REGRESSION: JACKKNIFE RESIDUAL NORMALITY

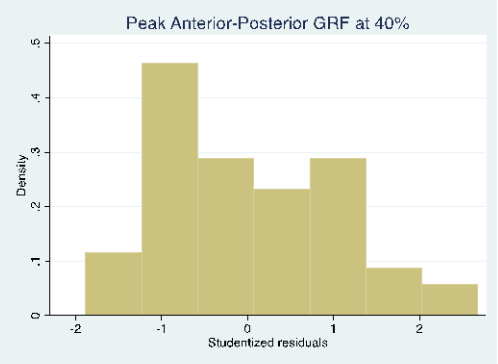
I.1 PEAK VERTICAL GROUND REACTION FORCE

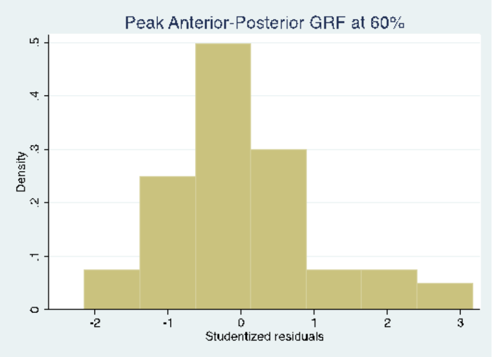


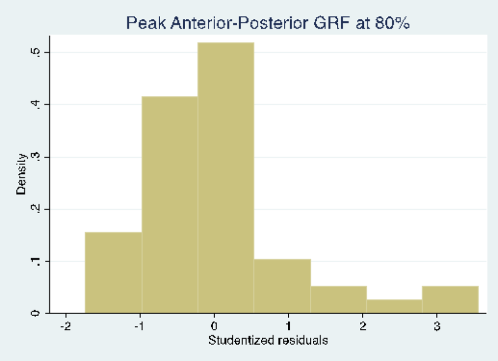




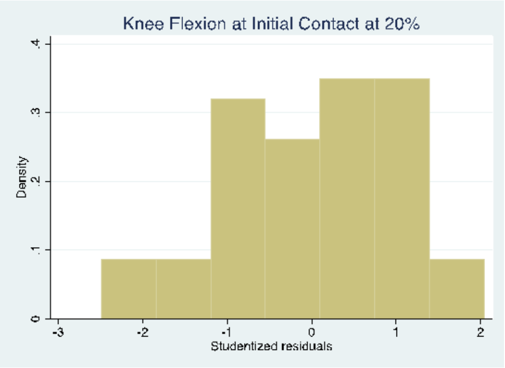
I.2 PEAK ANTERIOR-POSTERIOR GROUND REACTION FORCE

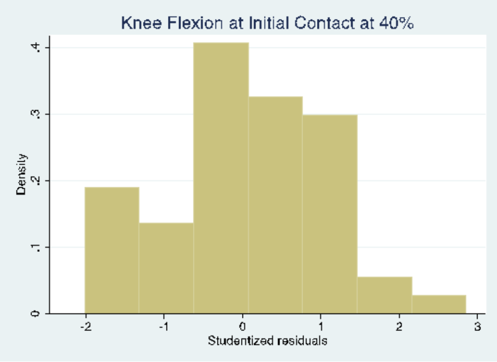


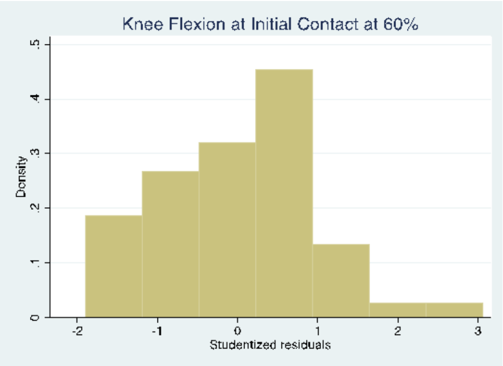


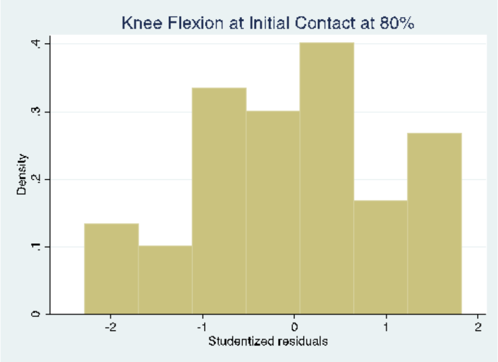


I.3 KNEE FLEXION AT INITIAL CONTACT





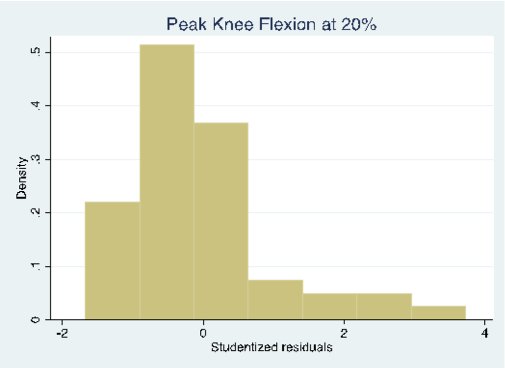


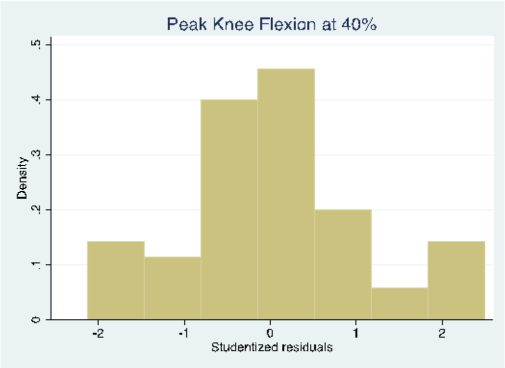


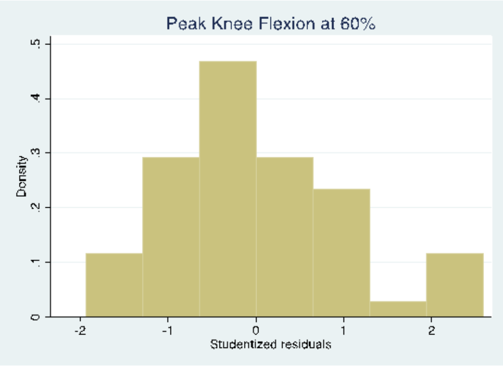
I.4 KNEE ABDUCTION AT INITIAL CONTACT

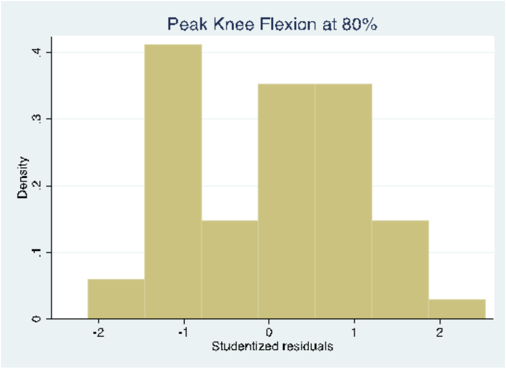
No independent variables were retained in any of the linear regression models.

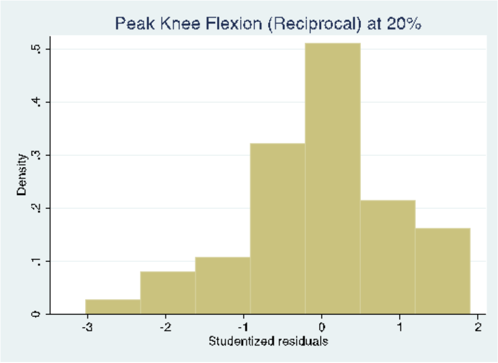
I.5 PEAK KNEE FLEXION

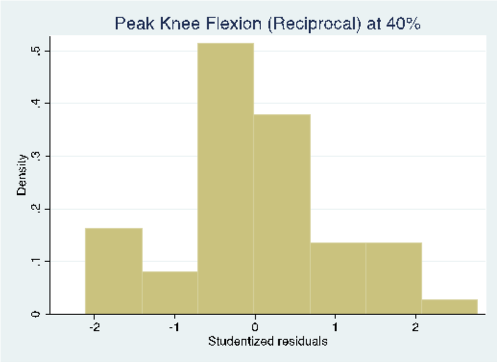


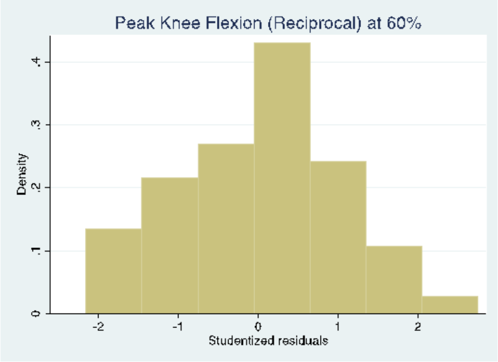


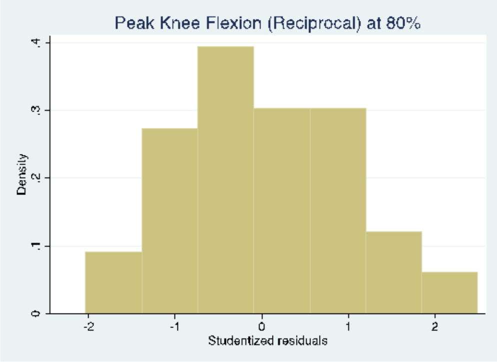

Peak Knee Flexion at 60%


Peak Knee Flexion at 80%


Peak Knee Flexion (Reciprocal) at 20%


Peak Knee Flexion (Reciprocal) at 40%

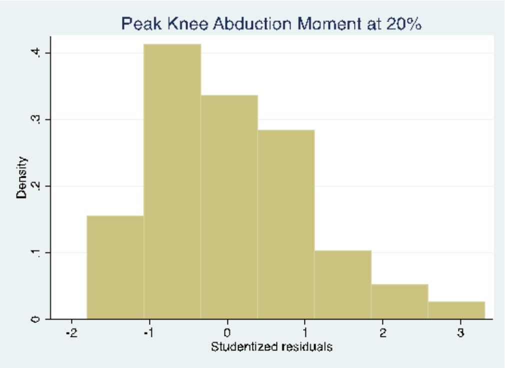

Peak Knee Flexion (Reciprocal) at 60%

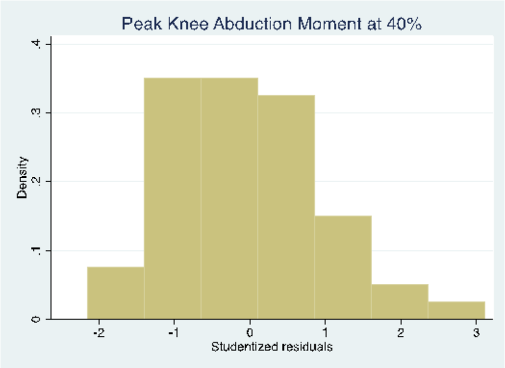

Peak Knee Flexion (Reciprocal) at 80%

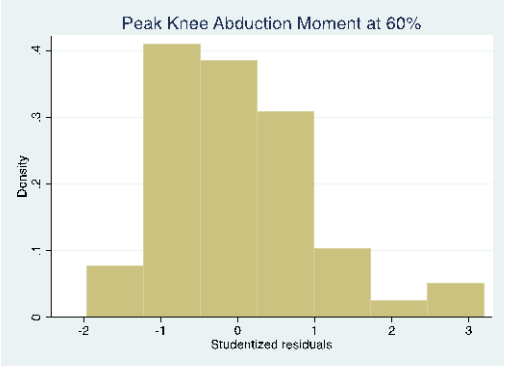
I.6 PEAK KNEE ABDUCTION

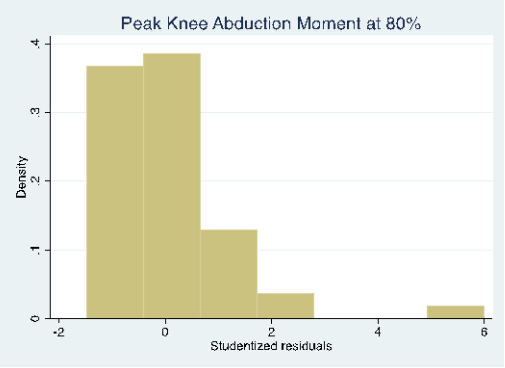
No independent variables were retained in any of the linear regression models.

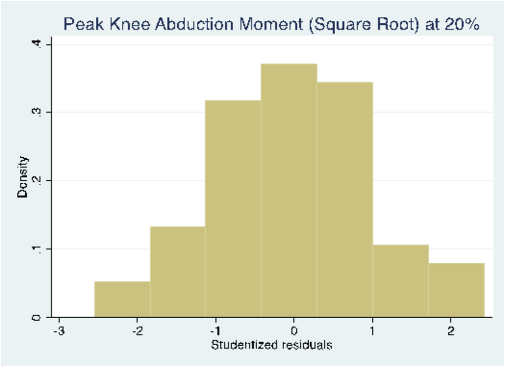
I.7 PEAK KNEE ABDUCTION MOMENT

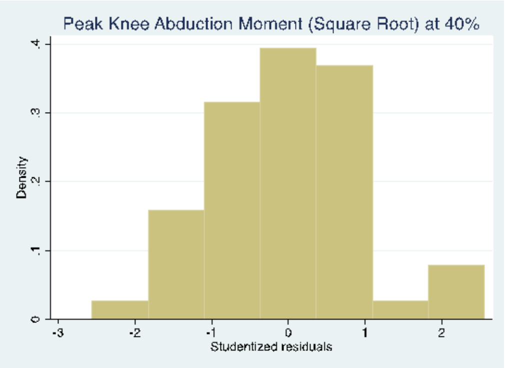


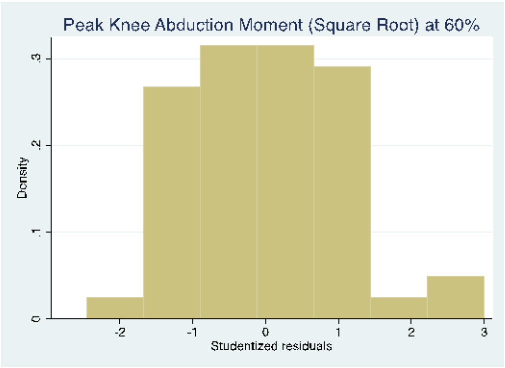


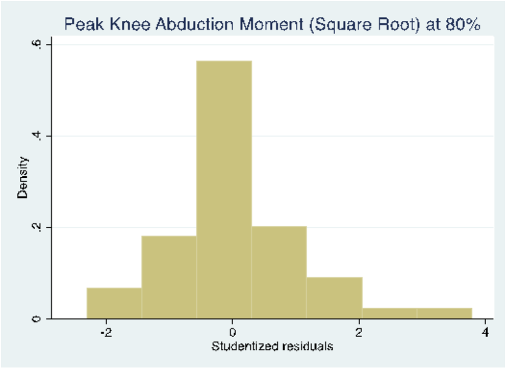









Peak Knee Abduction Moment (Square Root) at 60%


Peak Knee Abduction Moment (Square Root) at 80%

I.8 PEAK PROXIMAL ANTERIOR TIBIAL SHEAR FORCE

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