INERTIAL MEASUREMENT UNITS AS PROXY MEASURES FOR KNEE KINETIC VARIABLES ASSOCIATED WITH ACL INJURY

A Thesis presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

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August 2024

ABSTRACT

Anterior cruciate ligament injuries are among the most severe in sports, with significant long-term consequences for athletes, including a high risk of re-injury and early-onset osteoarthritis. Current rehabilitation methods and return-to-sport assessments often fall short in preventing re-injury, primarily due to a reliance on performance symmetry and strength profiles that do not adequately account for movement strategies. This study aims to bridge the gap between advanced biomechanical assessments and practical rehabilitation methods by investigating the potential of inertial measurement units to serve as reliable surrogates for traditional force plate measurements in assessing kinetic variables during ACLR rehabilitation.

The study involved a healthy population to establish baseline measures unaffected by existing health conditions. Multilevel linear regression models were used to evaluate the effectiveness of predicting GRF outcomes from accelerometer and gyroscopic metrics. It was hypothesized that IMU-derived metrics would strongly correlate with kinetic variables.

Contrary to the hypothesis that IMU-derived metrics would strongly correlate with kinetic variables, the results revealed only one strong correlation across all three movements, with the rest being either moderate or weak. Despite this, most of the regression models demonstrated high explanatory power, accounting for a significant percentage of the variance in the force plate metrics.

These findings suggest that IMU-derived metrics may not directly correlate strongly with force plate measurements, but they can still be valuable predictors when used in comprehensive modeling approaches. This demonstrates the potential for integrating IMUs into practical

rehabilitation and return-to-sport assessments, providing a more accessible and versatile method for monitoring kinetic variables and enhancing current ACLR rehabilitation protocols.

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NOMENCLATURE

ACL	Anterior cruciate ligament
ACLR	Anterior cruciate ligament reconstruction
CMJ	Countermovement Jump
GRF	Ground reaction force
IMU	Inertial measurement unit
RTS	Return to sport
AM	Anteromedial bundle of the Anterior Cruciate Ligament
PL	Posterolateral bundle of the Anterior Cruciate Ligament
IP	Incidence proportion
IR	Incidence rate
NC-ACI	Noncontact Anterior Cruciate Ligament Injury
BTB	Bone-patellar tendon-bone autograft
ST-G	Four-strand hamstring tendon autograft
HQ ratio	Hamstring-quadriceps ratio
LSI	Limb Symmetry Index
MEM	Micro-electromechanical system
ROM	Range of Motion

Chapter 1: Development of the Problem

Background and Rationale

Anterior cruciate ligament (ACL) injuries are among the most devastating in sports, often leading to significant long-term consequences. The annual estimate for ACL injuries in the USA exceeds 120,000, with the majority occurring in high school and college-aged athletes.[1] These injuries usually result in surgery intervention followed by many months of intensive rehabilitation, which in total can cost thousands of dollars per injury.[2] Even after rehabilitation, long-term clinical repercussions are expected for people who suffered ACL injuries. This can include the development of meniscal tears and a heightened vulnerability to experiencing early-onset osteoarthritis.[3, 4] Surgical intervention also leaves the athlete vulnerable to tearing their ACL again, as 20-30% re-injury rates have been reported in youth athletes returning to sports.[5-8] The economic burden associated with the eventual development of osteoarthritis following an ACL injury has been estimated to amount to $2.8 billion annually.[9] An ACL injury's aftermath can transcend physical pain, leading to psychological challenges for athletes due to sports and occupational restrictions.[10]

These injuries are highly prevalent among athletes, primarily due to their occurrence during movements that involve rapid changes in direction, deceleration, cuts, jump landings, and planted foot pivots.[11-13] It has been previously observed that approximately 70% of ACL injuries are non-contact in nature and typically arise from a dynamic malalignment of the lower extremity [14]. A few key potential characteristics of this alignment include increased hip adduction and internal rotation, knee (tibial) abduction, tibial external rotation and anterior translation, and ankle eversion that shift the center of mass outside of the athlete’s base of foot support.[12, 15, 16] ACL reconstruction (ACLR) is often required after an injury to the ligament to regain stability in the knee. Most ACLRs are performed in the adolescent population, and those athletes typically aim to return to competitive sport participation.[8, 17]

Even with improvements in surgical techniques and rehabilitation processes, decreased motor control and function of the injured knee is typical in athletes years after injury.[18-20] The injury-induced changes in mechanics, such as reduced mobility in the knee, lower limb muscle weakness, and impairments in proprioception[21-23], can give rise to changes in stress and tension in some regions of cartilage, resulting in varying tibiofemoral contact forces.[19, 24] This alteration in mechanics can cause difficulties for athletes who want to return to playing sports, which is evident from the statistic that merely 60% of nonelite athletes manage to return to their previous level of sport.[8] For those who return, re-injury rates to the ipsilateral, as well as injuries to the contralateral limbs, are high.[25] Reinjury rates are especially prevalent in athletes newly re-introduced to sport, with about 74% of second ACL injuries occurring within the first two years after surgery.[5, 26] Therefore, gaining a more comprehensive understanding of the factors that might contribute to an elevated risk of reinjury among athletes is crucial to gain a more comprehensive understanding of the factors that might contribute to an elevated risk of reinjury among athletes.

The time between ACLR surgery and the return to sport (RTS) is generally about seven months, but the duration can vary considerably for individual patients.[27] Return to sport clearances frequently rely on the time from surgery, yet this can be an imperfect metric for gauging readiness. Studies have shown no connection between the time elapsed after surgery and factors like force absorption/production and other functional deficits.[28] Another metric for determining readiness to return to sport is limb symmetry between the injured and injured limbs.[29] Isokinetic strength profiling of the thigh muscles and a variety of hopping tasks are commonly used to quantify functional symmetry between injured and non-injured limbs in a way that is accessible and can be administered easily.[30] However, these functional tests have shown limited ability to predict a successful RTS one year after an ACL injury since they only indirectly measure knee function and may not reveal any mechanical asymmetries.[31-35] Thus, it is critical to have a comprehensive assessment of athletes' movement and performance symmetry to guide appropriate RTS decision-making.

Examining kinematic and kinetic data to assess symmetry in the lower limbs has shown to be a more dependable method for predicting subsequent injuries since these metrics give a more in-depth analysis of an athlete's movement strategy.[36, 37] Several crucial parameters that have enhanced our understanding of ACL rehabilitation include hip and knee angles, knee moments, and vertical ground reaction forces (GRF) observed during jump-landing and cutting tasks.[38] Stiff landing patterns with decreased knee and hip flexion angles and increased GRF and knee extension moments during these tasks are associated with increased ACL loading.[37, 39, 40] This kind of analysis traditionally requires a motion capture system and force plates, which are considered the gold standard for measurement. Unfortunately, these tools are inaccessible to most and confine testing to a laboratory.[41]

In recent times, inertial measurement units (IMUs) have emerged as valuable tools in clinical biomechanical research as a cost-effective alternative that enables widespread usage and on-field testing. IMUs can provide researchers with data such as linear accelerations and angular velocities, giving insight into movement patterns during activities like jumping, cutting, and landing.[41] This technological advancement facilitates a more accessible comprehensive assessment than traditionally used to rehabilitate an individual's functional capabilities and imbalances. This supports the decision-making process for a secure RTS post-ACL injury and rehabilitation. Also, wearables such as IMUs enhance the ability to monitor biomechanical changes over time, guiding clinicians in refining rehabilitation strategies.[42]

However, research into IMUs' utilization in ACLR RTS scenarios remains limited. A recent study revealed that wearable sensors demonstrated fair-to-excellent concurrent validity when assessing the joint kinematic measurements often used in RTS evaluations after an ACL injury.[43] Others have reported correlations between knee kinetics associated with ACL injury risk (knee extensor and flexion moment, stiffness, and power) following ACLR, but the experimental protocols of those studies were not representative of those performed in injury risk assessments during the late stages of RTS rehabilitation.[44-46] A recent study aimed to find correlations between lab-based knee variables and IMU-derived measurements during typical ACLR rehabilitation tasks.[47] The results found strong positive correlations between tibial angular velocity and knee range of motion curve area under the tibia velocity curve. Since this study focused primarily on kinematic measurements, additional research exploring correlations between force plate-derived kinetic and IMU-based metrics during ACL rehabilitation assessment tasks is warranted.

Statement of the Problem

Despite advances in research and technological improvements, reinjury rates in athletes remain high. This is partially attributed to the disparity between laboratory biomechanical assessments and the current rehabilitation process. Traditional methods for clearing athletes to return to sport—largely based on performance symmetry and strength profiles—do not adequately account for actual movement strategies and are insufficient in preventing re-injury.

On the other hand, while biomechanical laboratory assessments effectively quantify joint load and stiffness, they are not widely accessible and, therefore, not regularly used in everyday rehabilitative settings. Recent studies indicate that even athletes who meet traditional clearance criteria exhibit asymmetries in certain kinetic variables during dynamic tasks such as jumping and cutting. Although some research has examined the correlation between IMU-derived variables and motion capture derived kinetic variables, there is a lack of studies investigating the potential of IMU variables to serve as effective surrogates or predictors for force plate derived kinetic variables. This gap is particularly relevant as these variables have been proven useful in identifying movement asymmetries in ACLR patients.

Further research is needed to deepen our understanding of the relationship between clinical assessment variables and IMU metrics, thereby bridging this gap and enhancing rehabilitation strategies. This study aims to show how IMU data can be integrated into ACLR return-to-sport assessments, potentially making biomechanical evaluations more accessible and practical in real-world settings.

Statement of Purpose

This thesis aims to identify correlations between force plate-derived kinetic variables and IMU-derived metrics during various movements conducted in the advanced stages of ACLR RTS rehabilitation. The secondary purpose of this study is to assess whether multilevel linear regression models can effectively predict GRF outcomes using the same accelerometer and gyroscopic metrics.

To explore these relationships accurately, the study will test a healthy population. This will ensure that baseline measures are unaffected by existing health conditions, providing a standard reference point for further comparative analyses. If these IMU variables prove to be effective surrogates and/or predictors for the force plate variables, future research could investigate whether this relationship holds in injured populations with the ultimate goal of being used as a replacement for force plates to assess kinetic lower limb asymmetries.

Research Hypotheses

Previous research has shown that tibial acceleration correlates strongly with kinetic variables such as GRF, knee moment, power, and stiffness during walking and running. Correlations have also been identified between both knee stiffness and knee extension moment and the area under the linear tibial acceleration curve during ACL rehabilitation-related tasks. Therefore, it is hypothesized that the IMU-derived measurements will strongly correlate with kinetic variables of interest (GRF, eccentric deceleration impulse, concentric impulse, force at zero velocity, peak landing force, loading rate, contact time, and jump height). Studies have demonstrated that predictive models for both whole body and knee-specific kinetics, created using IMU metrics through simple linear regression, have achieved high accuracy. Therefore, it is hypothesized that this study will produce highly accurate predictive models for whole-body kinetics, similarly using IMU metrics through multilevel linear regression models.

Significance of the Study

The significance of this study lies in its potential to provide valuable insights into incorporating IMUs within the ACLR rehabilitation protocol. The anticipated results may demonstrate a strong correlation between IMU measurements and those obtained from the force plate, potentially establishing IMU measurements as a reliable proxy. The development of IMU-derived proxy measures for knee variables could then be integrated into established protocols widely accepted by practitioners for assessing ACL injury risk. This can help bridge the gap between advanced technology and practical clinical applications, offering a more accessible and efficient method for evaluating ACL injury risk.

Limitations of the Study

- Fatigue could have been a factor that affected movement strategies.
- Participants in the study came from various sports backgrounds, so participants will likely have varying movement strategies.
- Results may have varied from those collected in more realistic field-based settings.
- Using the force plate data to derive initial contact/movement onset for the IMU metrics would not be possible in the field.

Delimitations of the Study

- Participants were between the ages of 18 and 35 years old.
- Participants who experienced a lower extremity surgery within the past year were excluded.
- Participants were recreationally active, exercising for at least 30 minutes for 3 days of the week.
- Participants must have been pain-free on the day of testing.
- Participants wore wear standardized lab shoes (Nike Air Zoom Pegasus).
- The Vicon Blue Trident was the inertial measurement unit used in this study.

Assumptions of the Study

- The force plates and IMUs were calibrated correctly throughout the testing sessions.
- Participants put in the maximum effort during the testing protocol.
- Participants were truthful when filling out their Musculoskeletal History and Fitness Activity Questionnaires.

Operational Definition of Terms

- Noncontact ACL injury: An ACL injury that happens without direct physical contact to the knee from another player. Typically, these injuries occur during unexpected athletic maneuvers, like cutting, landing on one leg, or sudden deceleration.
- ACL loading: The internal stress experienced within the ACL due to increased tensile force applied to the ligament.
- Initial contact: The moment when the feet make contact the ground.
- Concentric impulse: The force exerted during the concentric phase of a jump, which is the phase where the muscles are actively shortening to lift the body off the ground.
- Eccentric deceleration impulse: The force exerted during the eccentric lowering phase of a jump. The eccentric phase occurs when a muscle lengthens while under tension, acting as a brake to control a movement's speed and absorb forces.
- Field testing: The evaluation or assessment of an individual in real-world conditions outside a controlled laboratory environment.

Return-to-sport rehabilitation: a structured recovery program for athletes aiming to resume their previous level of sports participation after an injury.

Chapter 2: Review of the Literature

Introduction

Despite improvements in surgery and rehabilitation, athletes often experience persistent motor control and knee function issues long after ACL injuries.[19, 20] These problems, including limited knee mobility, muscle weakness, and proprioceptive deficits, can lead to variable tibiofemoral contact forces.[24] Traditional functional tests have limitations in predicting a successful RTS as they indirectly assess knee function and may miss mechanical imbalances.[28, 32, 33] Therefore, a comprehensive assessment of movement symmetry is crucial. Analyzing lower limb kinematic and kinetic data offers a more reliable predictor of future injuries. IMUs have emerged as cost-effective tools in recent years, providing data on linear accelerations and angular velocities during activities like jumping and cutting.[41] They offer a more accessible and continuous evaluation of functional capabilities and imbalances in ACL rehabilitation, aiding in secure RTS decisions.[41] However, research on IMU use to measure kinetic metrics in ACL RTS scenarios is limited.

This thesis seeks to identify associations between force plate-derived kinetic parameters and IMU-derived metrics observed in advanced ACLR RTS rehabilitation movements. This chapter provides an extensive literature review, including an introduction to the ACL, ACL injuries and re-injuries, the current state of care, the rehabilitation protocol, an overview of IMUs, and their application in ACL rehabilitation.

ACL Overview

Anatomy

The ACL is anatomically positioned within the knee joint, connecting the tibial intercondylar eminence at its distal attachment point to the posteromedial aspect of the lateral femoral condyle at its proximal attachment point.[48, 49] While it is commonly perceived as a single ligament, research has unveiled that only about 26% of knees possess a single-bundle ACL. A third intermediate bundle may often be present; however, the ACL is typically considered to comprise two primary bundles: the anteromedial bundle (AM) and the posterolateral bundle (PL).[48, 50] The AM fibers originate on the anteroproximal aspect of the femoral condyle and insert onto the anteromedial aspect of the tibia, whereas the PL fibers have their origin from the posterodistal aspect of the femoral condyle and insert onto the posterolateral part of the tibia.[51, 52]

The ACL varies in dimensions, typically 31 to 38 mm long, with a width spanning 10 to 12 mm. Of these two bundles, the AM is slightly larger, measuring approximately 6 to 7 mm in width, whereas the PL has a width of 5 to 6 mm.[48, 51] The tissue is characterized by an organized collagen matrix comprising approximately 90% type I collagen fibers, with about 10% type III collagen fibers.[49] This structure makes the ligament strong and flexible, an important combination for its role in knee stability.

The ACL is classified as an intracapsular ligament and has its own synovial membrane. The middle genicular artery facilitates the ligament's primary blood supply, supplemented by diffusion through its synovial sheath.[53] Innervation of the ACL is provided by the tibial nerve, which houses mechanoreceptors responsible for contributing to the ACL's proprioceptive function.[54]

Function and Loading

The ACL serves as a primary restraint against anterior translation of the tibia. It is also a secondary stabilizer against internal tibial rotation and valgus knee positioning. The ligament plays an important role in resisting anterior shear forces on the proximal tibia, internal tibial torque, and knee abduction moments.[55, 56] It is essential for maintaining knee joint stability, especially during weight-bearing activities. It provides 70–87% of the resistance to an anterior tibial shear force applied to the knee at small flexion angles (0–30°) and somewhat less of this resistance (62–85%) at larger flexion angles (60–90°).[57, 58]

The AM and PL fibers each play unique roles in preserving knee stability. The AM fibers are primarily responsible for countering anterior tibial translation, whereas the PL fibers play a pivotal role in resisting internal tibial rotation.[59, 60] These fiber bundles function as antagonists during standard knee flexion and extension. However, these fibers work synergistically when the knee's motion becomes multiplanar in scenarios involving anterior translation and internal tibial rotation.[61] This is essential for resisting tibial displacement and maintaining knee stability during complex movements. The performance of these fiber bundles is closely tied to the degree of knee flexion. AM fibers have increased activity within the range of moderate knee flexion angles (30-60°), while PL fibers become more dominant as the knee nears full extension (0-15°).[62, 63]

The ACL also plays a critical role in knee proprioception. Receptors within the ligament are activated when it undergoes stretching, and in response, they transmit afferent impulses to the central nervous system. This activation triggers muscle contractions, essential for dynamically adjusting stability and muscular stiffness to prevent ligament injuries.[64] When these mechanoreceptors are damaged, they can compromise the protective mechanisms of the surrounding muscles, potentially reducing overall knee joint stability and functionality.[64]

Injury Overview

Prevalence/Incidence

The prevalence of ACL injuries is a growing concern, with a substantial impact on public health and the financial burden associated with treatment and rehabilitation. The incidence proportion (IP) and incidence rate (IR) of ACL injury in female athletes was 3.5% and 1.5/10,000 athlete-exposures over 25 years. The IP and IR of ACL injury in male athletes were 2.0% and 0.9/10 000 athlete exposures over the same time. This means there was a 1.7-fold increase in the incidence of ACL injuries in females compared with males.[65] However, this pooled data from studies of different types of sports, with differing risks for ACL injury to provide general estimates of ACL injury rather than sport-specific estimates. Studies examining the risk of athletes in pivoting and jumping sports show that the gender gap in prevalence is large, reporting that female athletes who participate in these sports are at a 4- to 6-fold greater risk of suffering ACL injuries compared to their male counterparts.[61, 66]

A meta-analysis examining the annual growth rates of ACL injury incidence in Australia throughout a 20-year period (1998-2018) revealed a 10.4% annual growth rate for ACL injuries among females aged 5 to 14 years and a 7.3% among males within the same age group.[67] Similar trends were found in a previous investigation of ACLR rates in Australia from 2000 to 2015: annual growth rates of 8.8% and 7.7% per year for 5 to 14-year-old females and males.[68] Together, these findings indicate that ACL injuries are becoming increasingly common in the younger population. The authors of the more recent meta-analysis concluded that this trend would continue to rise if no measures are taken to mitigate the incidence rate of ACL injuries, expecting the annual rate to more than double by 2030-2031 compared to 2017-2018 levels, reaching an incidence of 77.2 per 100,000 population.[67]

Similar trends were observed in other countries. In Finland, a study focusing on pediatric ACL injuries concluded that young females aged 13 to 15 experienced the most significant increase in ACL injury incidence over an 18-year period compared to other age and sex groups.[69] Data from Southern Sweden and the United States also exhibited similar patterns. ACL rupture incidence typically increased during childhood, peaked between the ages of 15 to 24 years, and gradually declined thereafter.[70, 71] In the United States, ACL injury data is not systematically recorded, so it is not possible to know the exact number of annual injuries. However, it is estimated that more than 120,000 ACL injuries occur annually.[1]

The escalating occurrence of ACL injuries is becoming a significant issue in public health due to the high costs associated with surgery and the long road to recovery that follows. The prevalence of ACL injuries is not limited to specific age groups or geographical regions, and it affects both male and female athletes differently. Therefore, raising awareness, implementing preventive strategies, and conducting further research are essential steps to combat the rising incidence of ACL injuries and their significant impact on healthcare and society.

Injury Mechanism

The ACL can fail due to a single overload event, where a loading cycle exceeds the ultimate tensile strength of the healthy ligament, or through repetitive submaximal loading events, which cumulatively induce microdamage and weaken the structure over time.[72-75] While the single overload scenario represents an acute failure during an intense load, the repetitive submaximal loading events lead to a gradual accumulation of damage, resulting in the ligament's inability to repair itself and eventually failing under a seemingly routine submaximal load.[57, 73] These high-risk movements typically involve a combination of knee joint compression, anterior tibial shear, internal tibial torque, and knee abduction moment, occurring during jump landings, abrupt changes in direction, or sudden stops.[15, 16, 61, 76]

ACL injuries are typically classified into two main categories: contact and noncontact injuries (NC-ACI). Contact injuries result from direct blows to the knee joint, often due to interactions with other players. In contrast, NC-ACIs are characterized by their occurrence without player-to-player contact.[77] NC-ACIs often happen during activities such as single-leg landings, abrupt changes of direction, or rapid decelerations and makeup nearly 70% of all ACL injuries.[1, 12, 77] Extensive research has been conducted to optimize prevention strategies to comprehend the risk factors and mechanisms leading to ACL injuries.[12]

Most ACL injuries occur within the first 40 to 100 milliseconds after ground contact, with medial knee collapse being a common biomechanical pattern associated with these injuries.[78, 79] This pattern is characterized by increased hip internal rotation, knee abduction, and knee internal rotation, forcefully collapsing the knee inward, increasing the risk of injury.[15, 16, 61, 76] Researchers have identified internal tibial torque and knee abduction moments as crucial elements. However, there is ongoing controversy regarding their respective impacts on ACL loading. One study showed that internal tibial torque significantly influences ACL strain, particularly in scenarios of single-leg pivot landing.[80] Their *in vitro* dynamic knee loading model showed that peak ACL strain was sensitive to axial tibial torque, while the direction of the frontal-plane moment showed negligible effects. This finding suggested that tibial internal rotation torque, not knee abduction moment, plays a predominant role in ACL loading.[80] In contrast, other research has suggested that the role of internal tibial rotation in ACL loading is secondary, with knee abduction motion and anterior tibial translation being more significant factors.[81, 82]

Understanding the mechanics of an athlete's landing plays a pivotal role in the risk of ACL injuries and is essential for developing effective injury prevention and rehabilitation strategies.[83] For example, how an athlete lands significantly affects ACL injury risk. Athletes who experience ACL injuries often land flat-footed or with the hindfoot first, with minimal knee flexion.[84] This landing mechanic bypasses the tricep surae's capacity to dissipate ground reaction forces (GRF), transmitting impulsive forces directly to the knee. [12] In contrast, a forefoot-first landing allows the calf muscles to absorb GRF, reducing the risk of injury.[11] The cumulative results of various studies support the notion that impulsive axial compressive force from single-leg landing in a provocative position is a major contributor to NC-ACI.[12]

Abduction moment, though not a primary contributor, can increase the risk of ACL injury. Excessive abduction moment can lower the compressive force threshold required to cause ACL injury, and it can exacerbate the impact of axial compressive loads on ACL disruption.[85] The relationship between abduction alignment and compressive load threshold is especially important for female athletes, who inherently have more knee abduction. The triceps surae plays a protective role during safe landings. When an athlete lands on their forefoot, the triceps surae helps absorb GRF, reducing the strain on the ACL.[12] Studies have shown that the soleus can generate forces comparable to the hamstrings during jump-landing tasks, providing additional support to the ACL.[86]

Cutting maneuvers are pivotal in many high-contact sports and demand attention due to their role in ACL injury risks. Understanding the cutting phases is essential as each stage—from deceleration and redirection to re-acceleration—presents unique challenges to knee stability. [87] During the early deceleration phase, the point that aligns with the highest risk of injury, the knee undergoes flexion, abduction, and internal rotation.[88] These movements can lead to instability and potentially increase the anterior translation of the tibia. Additionally, during the deceleration phase of a cut, the body's center of mass shifts laterally, leading to greater knee abduction angles and potentially lower hamstring-to-quadriceps co-contraction ratios, further elevating injury risk.[89]

Injury Risk Factors

Understanding the risk factors to prevent ACL injuries effectively is important, and these factors are categorized into intrinsic and extrinsic factors. Intrinsic factors are internal to the individual, while extrinsic factors are external elements that can influence the risk of injury. Understanding the interplay of these factors is key to developing comprehensive prevention strategies and interventions tailored to individual athletes and sports.

Intrinsic Factors. Sex is a key determinant in the risk of ACL injuries, with females more vulnerable due to anatomical and biomechanical differences. They typically have larger Q-angles and smaller intercondylar notches, predisposing them to experience movement patterns that stress the ACL at a high level.[11] Females land with higher knee abduction moments and tend to have deficits in core proprioception and knee stiffness, amplifying their injury risk.[11] Additionally, female athletes often have a steeper posterior tibial slope, potentially contributing to their higher incidence of ACL injuries.[11] Hormonal influences, particularly variations in estrogen levels, may also play a role in the integrity and response of the ligament to stress.[11] Understanding these sex-specific factors is crucial for developing targeted prevention strategies.

Age significantly influences the susceptibility to ACL injuries, with those under 20 and over 40 years old most at risk.[90, 91] This is due in younger individuals to the ongoing development of the musculoskeletal system, where the ACL has not yet fully matured, and in older individuals to the decline in knee joint stability. The risk is compounded by a reduced ability to control neuromuscular function, affecting the support and energy absorption necessary for managing the knee's rapid acceleration and deceleration during activities like landing.[61]

Muscle imbalances, especially between the quadriceps and hamstrings, along with deficits in neuromuscular motor control, are prominent contributors to the risk of ACL injuries.[11, 12] Significant differences in quadriceps-hamstring activation are identified as a risk factor, largely due to their influence on anterior tibial translation and knee flexion.[11, 12] Less activation in the vastus medialis and medial hamstrings compared to the vastus lateralis and biceps femoris has been found to correlate with larger knee abduction angles.[92] Additionally, trunk dominance increases risk, particularly in women. This dominance, often influenced by growth and maturation factors, can lead to challenges in controlling and balancing the body during athletic movements.[15]

Fatigue notably affects biomechanics, heightening the risk of ACL injuries by diminishing neuromuscular control, reducing knee proprioception, and altering lower extremity mechanics.[93] Following intense activity, studies have shown increased knee abduction and knee internal rotation angles and decreased knee flexion angles.[93] Muscle activity, particularly in the quadriceps and hamstrings, also tends to shift after athletes have been pushed to their limits. When these changes combine, they put extra strain on the ACL, raising the risk of injury.[93]

Ligament dominance, where muscles fail to absorb ground reaction forces adequately, often leads to ACL injuries by forcing the knee into a valgus position, a situation seen more commonly in women.[17] Coupled with this is quadriceps dominance, where an overreliance on the quadriceps for knee stabilization can cause the knee to overextend and place excessive anterior shear stress on the tibia and ACL.[17] Additionally, leg dominance, marked by an imbalance in muscle strength and flexibility between limbs, and trunk dominance, which is an impaired ability to control the trunk, both disproportionately affect women and contribute to their higher incidence of ACL injuries.[17]

Extrinsic Factors. One of the primary extrinsic factors is the type of sport an athlete engages in. Sports that necessitate rapid changes in direction, frequent jumping, and physical contact with other players, such as soccer, basketball, and football are known to have a higher incidence of ACL injuries.[94] Another important factor is the playing surface. The playing surface also plays a critical role, as hard, slippery, or uneven surfaces can increase the likelihood of injuries, with those on hard or slick surfaces facing higher risks than softer grounds.[94]

Additional extrinsic factors include the type of footwear and environmental conditions. Footwear that lacks adequate support or is worn out can significantly raise the risk of ACL injuries due to insufficient stability or traction.[94] Similarly, environmental conditions, particularly adverse weather like rain or ice, can increase the chances of slips and falls, leading to potential ACL injuries.[94] Meteorological conditions significantly affect the foot-playing surface interface, particularly for sports played on natural or artificial turf. Low rainfall and high evaporation have been linked to an increased occurrence of non-contact ACL injuries in Australian football.[95]

The use of equipment and protective gear also plays a role. Properly fitted and appropriate protective gear, such as knee braces, can help mitigate the risk, while inadequate or improper gear might elevate it.[94] Training and conditioning practices are equally influential. Training and conditioning practices and the intensity and structure of practice sessions and games significantly impact injury likelihood. Athletes are at a higher risk during games than practices, suggesting that competition level and the manner of competing might increase the risk of ACL injuries.[95]

Re-injury Overview

Prevalence

Age has been consistently identified as a primary risk factor for ACL re-injury. A seminal study reported a 5-year risk for subsequent injury to either knee at 17% for patients younger than 18 years, compared to 4% for those over 25 years.[96] This clear gradient of risk based on age laid the groundwork for understanding ACL re-injury prevalence. Building on these findings, subsequent studies have consistently demonstrated even higher re-injury rates among younger patients. Webster et al. observed a 29% rate of subsequent ACL injury in a group of patients younger than 20, tracked over a minimum of 3 years.[97] This rate aligns with other studies that reported similar rates in cohorts of younger athletes (under 25 and 18 years, respectively), with follow-ups extending to 15 years in some cases.[20, 98] This trend has also been observed in top-level athletes, as a study on National Collegiate Athletic Association Division I sport athletes reported a 37% subsequent ACL injury rate among athletes who had previously suffered an ACL injury before college.[99]

Risk Factors

The risk factors for ACL re-injury are diverse and encompass patient behavior, genetic predisposition, surgical techniques, and gender-specific tendencies. Patients returning to high-demand sports, especially those involving pivoting or cutting movements, face a significantly amplified risk of both graft rupture and contralateral injury, particularly among younger athletes.[22, 100] A family history of ACL injury also doubles the risk of both types of injuries.[101] Surgical factors, including graft type and positioning, are critical; vertical graft positions, small or lax grafts, and the use of hamstring grafts as opposed to patellar tendon grafts, are associated with increased risks.[26] Gender differences also play a role; studies have shown an increased risk of graft rupture in males and a higher rate of revision surgery in females.[26, 101, 102] Additionally, the risk of contralateral knee injury post-reconstruction is reported double that of graft rupture, with younger or female patients, those returning to strenuous sports, and those with a family history of ACL injury being at higher risk.[103, 104] The choice of graft used in the reconstruction further influences the risk, with Allografts being demonstrated to have an increased risk of failure in younger athletes.[103-105]

Current State of Care

Diagnosis and Classification

Someone believed to have suffered an injury to their ACL should be evaluated immediately by a physician. The first observation should be in the patient’s gait, checking for any asymmetries or discomfort. It is also important to check for signs of peripatellar groove loss during an examination, as it could indicate effusion or hemarthrosis.[106] A previous study showed that 77% of athletes with acute knee injuries accompanied by hemarthrosis had experienced a partial or complete ACL tear.[107] Hemarthrosis creates substantial pain during range of motion exercises due to the increased intraarticular volume.[108] This pain leads to extensive muscle guarding and spasms, especially in the hamstrings, restricting the knee's range of motion. While the patient might not be able to flex the knee fully, a loss of hyperextension tells more of an ACL disruption. The compressed torn ACL stump between the tibia and femur, alongside the joint effusion, hinders full extension.[108]

Beyond the initial qualitative observation, extensive studies have been conducted on the diagnostic validity of various physical examination tests for ACL tears, with the Lachman, pivot shift, and anterior drawer tests being the most studied. All three tests have demonstrated high specificity, allowing them to be used independently for a valid ACL diagnosis.[109] Combining multiple history elements and physical examination tests is recommended to enhance diagnostic accuracy. Elements such as experiencing a pivoting traumatic event, a "popping" sensation at the time of injury, or immediate effusion following trauma can significantly aid clinicians in making a valid ACL diagnosis.[106]

If the previous tests indicate that the patient has injured the knee, the next step would be to complete some radiographic imaging. Magnetic resonance imaging is the primary study to diagnose ACL injury in the United States. It has the added benefit of identifying meniscal injury, collateral ligament tear, and bone contusions.[106] The sensitivity and specificity of MRI for detecting an ACL tear are 86 and 95 percent, respectively, as confirmed by arthroscopy.[110] This can also help determine the severity of the injury, classifying it as either a partial or complete tear of the ligament.

If an injury is found on the initial visit, the next typical step includes physical therapy immediately, regardless of the decision to treat this injury conservatively or surgically. This treatment aims to maintain range of motion and develop quadriceps strength. It is suggested during this time that crutches should be offered only for a limited time if the patient has considerable discomfort with ambulation.[106]

Whether to recommend an orthopedic surgical consultation for an ACL injury heavily relies on the patient's preferences and level of physical activity. Surgical intervention tends to be more common among younger, highly active individuals, especially those who wish to resume high-impact activities that involve rapid movements, sharp turns, and jumps.[106] Surgery is also a strong consideration for patients experiencing recurrent knee instability or those with associated injuries to the meniscus or collateral ligaments. However, a subset of athletes regain their pre-injury activity levels without surgery.[111] Indicators favoring a conservative management approach include infrequent episodes of knee instability, a nearly full range of knee extension, minimal meniscal damage as evidenced by MRI, strong quadriceps muscles, and the ability to perform the crossover hop test successfully.[106, 111]

ACL Operative Technique and Graft Choice

ACLR can be performed using two primary operative techniques: single bundle and double bundle. Single bundle ACLR, the traditional gold standard, involves reconstructing only the ACL's AM bundle.[112] On the other hand, double bundle ACLR reconstructs both the AM and the PL bundles, aiming to mimic the native ACL anatomy better.[112] While double bundle ACLR tends to be more common in athletic populations due to its superior anterior and rotational stability, it also demands a more complex, invasive, and costly surgical procedure.[112] A recent meta-analysis suggests that despite the increased stability offered by double-bundle techniques, clinical outcomes, and failure risks are comparable between the two methods.[112]

Regarding graft choices for ACLR, there are three main categories: autografts, allografts, and synthetic grafts, with autografts being the most preferred option due to their lower rejection rates, better long-term strength, and ready availability.[113] The most common types of autografts are the bone-patellar tendon-bone (BTB) and the four-strand hamstring tendon (ST-G).[113] In a meta-analysis of short- to mid-term follow-up after primary ACLR, ST-G autografts failed at a higher rate than BTB autografts.[113] the failure rates in both categories were minimal, and there was minimal variation in graft laxity across both types. It was concluded that each graft type is a feasible alternative for initial ACLR.[113] The slight variance in failure rate should be considered as a component of a comprehensive discussion with each patient regarding the selection of graft, incorporating considerations of possible variations in morbidity at the donor site, rates of complications, and patient outcomes.

Ultimately, the choice of operative technique and graft type should be tailored to fit the individual patient's goals, considering factors such as desired activity level, associated risks, and potential for functional improvement. It is crucial to acknowledge that even though ACLR can significantly enhance the quality of life and athletic performance, it may not fully restore knee kinematics to their pre-injury state.

Rehabilitation

For athletes progressing through the rehabilitation journey after ACLR, the process is structured into various stages, culminating in the ultimate goal of RTS following ACLR. Given the high re-injury rates observed in sports, rehabilitation aims to minimize re-injury risk instead of solely returning to their pre-injury performance levels. The transition through different rehabilitation phases depends on meeting specific criteria encompassing muscle strength, neuromuscular control, and task proficiency, ensuring that the athlete is prepared for the demands of their sport. This approach, centered on criterion-based progression, contrasts most of the existing literature that predominantly relies on the time elapsed since surgery as the main factor for decision-making.[114] On average, athletes take about seven months to return to sport following an ACLR surgery, though this timeline can vary significantly between individual patients, ranging from 6 to 24 months.[115] In a systematic review, Barber-Westin and Noyes examined 264 studies and identified the following criteria for RTS: 40% lacked any criteria for RTS, 32% solely relied on post-operative time, 15% combined time with subjective criteria, and 13% emphasized objective criteria for RTS. Among the studies focusing on objective criteria, the criteria utilized included: 9% based on muscle strength criteria (requiring 80-90% of quadriceps and hamstrings strength), 6% considered factors such as effusion and range of motion, 4% used a single-leg hop test, 1 study assessed stability, and another relied solely on validated questionnaires.[116]

Time from surgery

The reliance on a time-driven approach for RTS is partly due to accelerated rehabilitation protocols, decreased clinician activity restrictions, and an athlete's regained confidence and reduced pain during sports-related activities. However, this method can also be swayed by pressures from coaches, parents, or teammates who are eager to see the athlete return to competitive activities within specific timelines.[28] These factors, in combination, might lead to an athlete being cleared for full activity based solely on the time from surgery, potentially overlooking crucial aspects of their actual readiness for high-level sports participation. On the downside, using time from injury as the sole metric for RTS clearance presents significant risks and limitations. This method might lead to a gap between the athlete's perceived readiness and physical capabilities, as subjective assessments do not always align with objective function and strength scores.[117] Given the elevated risk of a second ACL injury upon RTS, the sole reliance on time-driven guidelines can be counterintuitive and risky as the athlete begins to expose their lower extremity to high loads and complex motions, potentially leading to re-injury.[118] Significant deficits in muscle strength, motor coordination, and proprioception often remain even at 1-year follow-up, demonstrating the need for a more comprehensive and objective approach to RTS decision-making, incorporating both time and functional performance measures.[28]

Strength Recovery

Objective testing is pivotal in rehabilitation and the collaborative decision-making process associated with RTS following ACLR.[114] Nonetheless, there remains a lack of agreement on the optimal tests and metrics to effectively monitor an athlete's progress, particularly during the critical phase of RTS. One of the pivotal aspects of this journey is restoring and evaluating muscle strength, particularly in the quadriceps and hamstrings.[33, 119] The muscle strength deficit has been directly linked to the potential risk of future knee injuries, enhancing quadriceps and hamstring strength, an important component of the rehabilitation process.[120]

The hamstring muscles are crucial protectors of the ACL, particularly at 15°–30° knee flexion—a common position during ACL injuries.[121] Hamstring strength deficits in ACL-deficient knees often signify poor knee function.[122] Neurophysiological responses like quadriceps inhibition can amplify hamstring strength, contributing to a muscle balance protective of the ACL. [123] However, persistent strength deficits typical after ACL injuries and reconstructions can skew the hamstring-quadriceps ratio (HQ ratio), resulting in dynamic instability and a heightened risk of further injuries. [124, 125]

Post-ACLR strength evaluation tests have become a staple in clinical and research settings, with a battery of tests being administered to gauge the athlete's readiness for RTS.[33] Isokinetic dynamometry is a valuable tool in this process, providing an objective measure of muscle strength.[126] Despite critiques about its lack of functional relevance to actual sporting and training scenarios, it is hailed as the ‘gold standard’ for muscle strength measurement due to its convenience, reproducibility, and reliability.[127] This method assesses key aspects of knee strength following ACLR surgery, including the Limb Symmetry Index (LSI) of the injured and uninjured leg and the HQ ratio of the injured leg. These metrics are important in guiding clinicians in making informed decisions about an athlete’s readiness to re-engage in sports.[128] Clinicians aim to bring patients back to 'normal' strength levels after an ACL injury. They do so by helping their patients achieve benchmark values in strength symmetry, which are then integrated into their RTS criteria.[33] 'Normal' LSI values are typically expected to range above 70-90%, and the 'normal' HQ ratio is reported to fall between 0.5 and 0.8, serving as indicators for a safe and successful RTS.[33]

While assessing and achieving muscular strength benchmarks are crucial steps in the ACL rehabilitation process, it is important to complement these strength tests with functional assessments that closely mimic the patient's real-life activities and sports-specific movements when assessing an athlete's readiness to return to sport.[119] Muscular strength tests obtained through isokinetic dynamometry provide valuable quantitative data on the patient's progress, but the results need to be interpreted within the broader context of the patient's overall functional ability. Integrating functional tests ensures a more comprehensive evaluation of the patient's readiness for RTS, capturing their strength, movement mechanics, and stability under dynamic conditions.[33] This holistic approach ensures that the patient is strong and capable of safely and effectively performing in their specific sport, minimizing the risk of re-injury and enhancing their long-term athletic performance.[114]

Horizontal Hop Tests

Functional hop tests, such as single hop, triple hop, and cross-over hop for distance, have traditionally served as standard assessments to gauge an athlete's readiness for RTS post-ACLR because they are considered clinician-friendly options for evaluating knee function.[119, 129] In many cases, when clinicians do not have access to biomechanical equipment, the athletes are only assessed on hop distance.[129] To pass this functional test, the patient must exceed 90% LSI for safe RTS. However, this practice has been questioned due to the task's potential to mask underlying movement deficits and compensatory strategies.[130, 131]

Recent research has questioned the reliability of horizontal hop distance as a definitive metric for assessing RTS readiness, pointing out that achieving symmetry in hop distance does not necessarily translate to symmetry in crucial aspects of lower limb biomechanics, such as joint angles, muscle forces, and power generation.[34, 35, 132] It has also been observed that during rehabilitation, symmetry in hop distance is typically achieved sooner than symmetry in isokinetic knee strength[133], implying that using the hop distance LSI as a measure may lead to overestimating the rehabilitation progress.[134] Additionally, these tests have been unsuccessful at predicting the success of RTS one year post-injury.[135] Evidence also indicates that even after reaching the traditional discharge criteria time from surgery and strength and hop test performance symmetry), athletes may still experience a decline in their performance levels, and the re-injury rates remain significantly high post-RTS.[36] However, the landing phase of the hop-for-distance task can hold significant potential for providing valuable insights by identifying possible compensatory mechanisms in the hip and ankle joints.[36] The issue with this detailed information is that it requires 3D biomechanical analysis, which is not readily accessible in typical clinical settings.

Vertical Jump Tests

More recent research has been exploring the use of vertical jump tests for assessing limb asymmetry during the end stages of rehabilitation.[36, 39, 136] Achieving symmetry in the single-leg vertical jump height proves more challenging than attaining symmetry in horizontal hop distance.[36, 132] In the context of healthy athletes, vertical jump tests are frequently utilized as a performance assessment tool, primarily because of their straightforward nature and the efficiency with which they can be conducted.[137, 138]

An important factor distinguishing this task from the horizontal hop is the equipment needed to access asymmetries. Horizontal hopping has been shown to be most useful in quantifying lower limb kinematics, but the kinetics of this task are difficult to measure due to the unpredictability of each hop distance. This means this task can only be useful with a 3D motion capture system, which is inaccessible to most.[34] In contrast, the stationary nature of the vertical jump ensures consistent contact with the force plate, allowing for more reliable and comprehensive capture of kinetic data throughout the entirety of the jump. This means that vertical ground reaction forces and asymmetries during double and single-leg jump activities can be achieved through dual-force platforms, a technology that is becoming increasingly common.[136]

A recent study analyzed asymmetry in vertical jump data from patients who met traditional discharge criteria, using solely the information captured by a force plate.[136] Despite passing the traditional discharge criteria, the findings revealed persistent asymmetries in athletes across various aspects of vertical jump tests, including the concentric (push-off) phase, the landing phase of bilateral jumps, and numerous performance metrics like jump height, reactive strength index, and contact times. The study's authors emphasized the need for clinicians to prioritize achieving symmetry in ground reaction forces and key performance metrics during vertical jumps, as a strategy to diminish the risk of injury and enhance overall athletic performance.[136]

In summary, ACL rehabilitation has experienced significant advancements, particularly in developing optimal tests and metrics for monitoring athletic progress. Vertical jump tests have emerged as a promising tool for assessing limb asymmetry during the final phases of rehabilitation, providing a more consistent and reliable means of capturing kinetic data than horizontal hopping. Despite these advancements, a need remains for wider access to essential equipment like dual-force platforms, which facilitate comprehensive analysis and ensure accurate assessments. The recent study shows persistent asymmetries in vertical jump tests, even after patients met traditional discharge criteria, emphasizes the necessity of focusing on symmetry in ground reaction forces and performance metrics. This approach not only aids in mitigating injury risk but also significantly enhances overall athletic performance. The healthcare community must continue advocating for and investing in advanced monitoring technologies, ensuring clinicians are well-equipped to navigate athletes safely and efficiently through their recovery journey.

Inertial Measurement Units

Introduction

In the 1930s, IMU technology emerged, primarily for aiding in aircraft navigation, but its utility was initially constrained by the size, cost, and power requirements, confining it to larger-scale applications.[139] This technology, which could gauge velocity, orientation, and gravitational forces, later became more accessible with the inception of micro-electromechanical system (MEMS) IMUs. Drawing inspiration from the natural world, specifically the human inner ear's mechanism for balance, these devices employed a combination of movable and stationary electrodes, or piezoelectric strain gauges, to create differential capacitance.[140] The evolution of MEMS IMUs introduced devices that were not only more affordable but also compact and energy-efficient, triggering a rapid expansion in their application across diverse fields.

The development of MEMS IMUs catalyzed their popularization, particularly in the last two decades, as they began to provide valuable data for athletes, coaches, and sports researchers.[141] These devices, notable for their portability and affordability, have been valuable in monitoring athletic motion during training and competitions—a task for which traditional marker-based motion capture systems were not always practical. IMUs are valuable in transforming an athlete's performance into quantifiable data, which helps refine training programs, manage exertion levels, optimize athletic development, and increase training effectiveness by providing immediate feedback to coaches and athletes.[142]

The technology's early iteration included two types of sensors: accelerometers, which measure inertial acceleration, and gyroscopes, to gauge angular rotation, each with three degrees of freedom to measure from three axes.[139] Acceleration values obtained from the accelerometer and angular velocity from the gyroscopes are kept separately. Since both sensors can measure angles, calibrating both data sets can yield a more accurate output. Advancements in technology led to the inclusion of a third sensor, the magnetometer, which measures the yaw angle rotation and enhances gyroscope readings. This addition particularly benefits dynamic orientation calculations, providing more reliable measurements with reduced drift errors over time. [139]

Utilization in Sports Performance and Clinical Research

Integrating IMUs in sports like tennis, badminton, baseball, and golf has revolutionized performance analysis and equipment calibration. In racquet sports, these devices provide detailed wrist motion data, which is essential for technique refinement and strategic development.[143] Similarly, embedded IMUs in sports equipment like baseballs and golf clubs offer real-time feedback, enhancing athletes' interaction with the gear.

The expansion of IMU data usage has been useful in sports focusing on lower body dynamics, like basketball and volleyball. It can provide insights into position-specific training, and match demands. This has been reported to improve coaching strategies and play a role in preventing general knee injuries.[143] IMUs are also particularly effective in tracking tibial shock, which is invaluable in running due to its close link with the risk of stress fractures, a common issue in high-impact activities like distance running.[144, 145]

IMUs also play an important role in monitoring athlete workloads in these team sports settings. They track various variables, ranging from accelerometry to the total distance covered.[146] These measurements are then categorized into external loads, which measure the work done by the athlete, and internal loads, representing the biological stress responses. This comprehensive load monitoring is key to optimizing recovery periods and reducing the risk of overuse injuries.[146]

While the application of IMUs in sports performance has impacted load management and injury prevention, their role has recently extended into the realm of clinical research and rehabilitation, where these devices offer a bridge between the precision of lab-based evaluations with the practical needs of assessments and the practical needs of real-world recovery scenarios. In sports medicine, IMUs have been utilized to measure parameters like impact severity, metabolic rate, and kinematic data, which are critical for improving training regimens and managing athlete load. The accuracy of these devices is crucial, as they can serve as a more cost-effective and practical alternative to laboratory gold standards in certain contexts.

There are potential benefits of utilizing IMUs in home health scenarios as they offer an objective method to assess exercise performance, which is crucial in settings where professional supervision is absent. A recent study focused on using IMUs to monitor home-based exercise programs, a topic gaining relevance due to the increased demand for remote healthcare services following the COVID-19 pandemic. It involved healthy volunteers performing specific lower limb and core exercises in both a controlled lab environment and unsupervised home settings, with wearable IMUs collecting kinematic data. The findings revealed that participants executed the exercises similarly in both settings. However, they performed all movements faster than the physiotherapist's demonstrations, highlighting the need for a wearable system that provides user feedback to regulate the pace of movement.[147] This feature is particularly beneficial for monitoring compliance and quality in physical therapy programs, ensuring exercises are performed with the correct speed, intensity, and range of motion. In the growing trend towards telerehabilitation, IMUs allow for remote consultation, providing healthcare professionals with reliable data on patient exercise routines.

IMU Utilization in ACL Rehabilitation

Overview

The utilization of IMUs for collecting kinetic data is well-validated by numerous studies demonstrating their accuracy and applicability across various movements and conditions.[41] IMUs effectively evaluate the mechanical load on joints through their wearability and the capability for widespread measurement.[148] For instance, studies have demonstrated that IMUs can reliably estimate joint kinetics in daily activities, with methodologies such as inverse dynamics and machine learning providing data analysis.[148]

In rehabilitation and sports, IMUs have been particularly valuable. The diversity of activities IMUs can monitor is evidenced by studies analyzing everything from level walking to sports-specific movements.[139] For example, walking has been thoroughly assessed in various studies to ensure the accuracy of IMUs in capturing gait-related kinetic data. Such studies have consistently shown that IMUs can provide reliable estimates of joint kinetics during walking, with certain adaptations allowing for the assessment of other dynamic activities like running turns and sports maneuvers, thus broadening their utility beyond the laboratory setting.[149]

The development of advanced algorithms has greatly enhanced the potential of IMUs in clinical settings. A systematic review examined the use of IMUs in estimating joint kinetics. The reviewed studies' methodologies are divided into Inverse Dynamics-Based Method (IDM) and Machine Learning-Based Method (MLM). IDM involves biomechanical modeling and uses either a bottom-up or top-down approach to estimate joint force and moment. In contrast, MLM estimates joint kinetics by training models based on datasets of IMU data and truth reference values. The study found that both IDM and MLM can be effectively applied in the kinetic analysis of joints across various activities like walking and running, despite their differing characteristics in estimating kinetic variables. The results of 37 studies, validated for estimation accuracy, showed strong correlations with truth references, particularly in activities like walking, material handling tasks, and sit-to-stand movements. These findings indicate the potential of IMUs systems to provide comparable results to laboratory-based systems in estimating joint kinetics, offering a versatile tool for ACL research and rehabilitation.[148]

The growing evidence supporting using IMUs for kinetic data collection in clinical and sports settings is compelling. As research continues to evolve, IMUs are likely to become even more integral to biomechanical assessments, rehabilitation strategies, and sports performance evaluations, thanks to their validated accuracy and versatility. A scoping review out-of-lab on the effectiveness of IMUs in ACL rehabilitation monitoring and assessment demonstrated the use of IMUs in various activities closely related to daily life and sports, such as walking, running, and other dynamic movements. This underscores the versatility of IMUs in capturing a broad spectrum of joint movements, making them invaluable tools for tracking the rehabilitation progress of patients with ACL injuries. The studies employed inverse dynamics-based and machine learning-based methods to estimate joint kinetics, signifying a methodological diversity that caters to a wide range of research and clinical objectives. Additionally, some studies utilized both methods, demonstrating the adaptability of IMUs to various analytical frameworks. Machine learning techniques were also highlighted, showcasing how neural networks, random forest regression, and other algorithms can be trained to predict joint kinetic variables from IMU data, further validating the use of IMUs in dynamic and complex rehabilitation scenarios like ACL recovery.[41]

IMUs as proxy measures in ACL research

In exploring the utilization of IMUs in ACL research, it's clear that traditional methods for collecting kinetic data, which largely depend on inverse dynamics and machine learning-based approaches, face practical clinical constraints. These methodologies utilize the modeling of biomechanical systems that require detailed data specific to each subject, including information on body segment masses, dimensions, and inertial characteristics. Consequently, these approaches are heavily subject-dependent, demanding extensive knowledge for accurate modeling.[41] This complexity not only increases the requirement for specialized expertise but also limits the applicability of these methods in diverse clinical settings.

Adding to these limitations, most ACL research has been performed in a laboratory setting. This often involves sophisticated equipment and is constrained by the need for specialized environments, such as motion capture laboratories.[41] This restricts their accessibility and applicability in typical clinical scenarios, where such resources may not be available. This setting can also impose restrictions on the movement tasks, thus reducing the ecological validity of the findings.[41] Laboratory methods also involve time-consuming protocols, complex analysis, and expensive technical equipment, which may hinder their widespread adoption by clinicians working with ACLR athletes. To address these challenges, field-based functional assessments have been proposed as a more realistic approach, offering a practical alternative that could facilitate more naturalistic and accessible rehabilitation processes.[146]

Per Newton's Second Law, estimating GRFs using measured body kinematics presents an enticing alternative. This approach bypasses the need for direct force sensors at the point of contact. By aggregating inertia forces across multiple body parts, the GRF can be accurately estimated.[150] Significantly, this method serves as a low-cost, reliable proxy measure of kinetics in non-laboratory settings, offering a practical solution to the constraints posed by previous approaches. Adopting proxy measurements through IMU technology marks a substantial leap forward in the field, facilitating broader and more efficient applications in clinical environments.

A study by Havens et al. investigated how accelerations recorded by wearable accelerometers can indicate knee loading asymmetries found in individuals post-ACLR. The research involved participants performing running exercises while a marker-based motion system concurrently recorded their movements with accelerometers attached to their shanks and thighs. The studies showed that between-limb differences in thigh axial acceleration were instrumental in explaining a substantial portion of the variance in knee power absorption and ground reaction forces. However, the shank-mounted IMUs were not found to be effective in explaining the variance in any kinetic variables. This could be attributed to the placement of the IMUs between the lateral malleolus and the lateral epicondyle rather than on a rigid part of the segment, such as on the mediodistal tibia, superior to the medial malleolus. Additionally, this study focused on comparing limb asymmetries between the devices, yet it did not analyze the correlations between the actual values obtained from these devices.[44]

Another study by Sigward aimed to assess gait concurrently using inertial sensors and a marker-based motion system with force platforms to identify knee loading deficits in clinical settings post-ACLR. The research involving individuals post-ACLR demonstrated a strong correlation between peak shank angular velocity and knee extensor moment. Additionally, the study found that the between-limb ratios of angular velocity effectively predicted the ratios of extensor moments, even when spatiotemporal gait parameters showed no differences between limbs.[151]

These studies have established a correlation between IMU metrics and kinetic measurements during walking and gait, a significant step forward in movement analysis. However, extending this research to understand how well these metrics align during more dynamic tasks such as cutting and landing is crucial. These activities are critical in assessing an athlete's readiness to return to play and closely mimic the conditions that could lead to injury, thereby providing a more comprehensive evaluation of rehabilitation progress and injury risk.

Sigward and Pratt aimed to address this by assessing single-leg landing concurrently using inertial sensors and a marker-based motion system with force platforms to identify knee loading deficits in clinical settings post-ACLR. The study, which involved individuals approximately 5.1 months post-ACL reconstruction, found that angular velocities measured with gyroscopes placed on the lateral thighs and shanks could predict knee moments and power (with strong validity). The high intraclass correlation coefficients (greater than 0.947) and the fact that thigh angular velocity accounted for 66% of the variance in knee power absorption indicate that gyroscopes can provide meaningful clinical data.[45]

In a 2019 study by Morgan, the primary focus was comparing tibial accelerations during drop landings with kinematic and kinetic risk factors for ACL injuries, as measured through three-dimensional motion capture. The research also delved into the differences in these measures between soft and stiff landings. Key findings include a moderately positive correlation between peak internal knee extension moment and peak anterior acceleration and a strong positive correlation between peak internal knee extension moment and peak posterior acceleration, both across conditions and within subjects. Interestingly, no significant associations were found between peak internal knee adduction moment and peak medial acceleration, whether across conditions and within subjects or for preferred landings alone. These outcomes indicate that accelerometers may predict knee joint moments in the sagittal plane during bilateral landings, suggesting their greater suitability for predicting knee joint moments in the sagittal plane compared to the frontal plane.[152]

A study by Jones et al. aimed to establish correlations between laboratory-derived knee variables—such as knee range of motion (ROM), changes in knee moments, and knee stiffness—and metrics derived from IMUs. The IMUs, placed on both the tibia and thigh, captured angular velocities and accelerations during movements typically used in ACLR RTS assessment, like bilateral and unilateral drop jumps and a 90° cutting task. The results revealed strong positive correlations between knee ROM and the area under the tibia angular velocity curve across all movements, suggesting that IMU-derived angular velocities and accelerations could serve as proxy measures for knee variables in ACL injury risk monitoring. Additionally, the study found that data from tibial-mounted IMUs showed the strongest correlations with knee variables, suggesting that tibia placement is the most effective location for IMUs in assessing ACL injury risk.[47]

While recent studies have demonstrated encouraging outcomes of IMU accelerometer data as an alternative to traditional kinetic data in ACLR RTS assessments, there remain gaps in the existing literature. Previous studies have focused on using accelerometer data as a stand-in for kinetic data typically derived from motion capture systems. However, in most clinical environments, force plates represent the extent of biomechanical tools available, leading to a gap between the comprehensive data collected in advanced biomechanical assessments and what is practically utilized in these settings. This disconnect highlights the need to evaluate the efficacy of IMU accelerometer data in accurately replicating the kinetic variables that recent ACLR RTS research has deemed sensitive to limb asymmetries. This would not only validate the practicality of IMUs in clinical settings but could also significantly enhance the precision and effectiveness of ACLR rehabilitation protocols, bridging the gap between high-end biomechanical research and day-to-day clinical applications.

Conclusion

Enhancing the RTS clearance process during ACLR rehabilitation is essential for reducing re-injury risks, improving knee functionality, and minimizing the likelihood of knee osteoarthritis development in patients.[153] Recent studies have demonstrated the significance of vertical jump tests in detecting limb kinetic asymmetries among ACLR patients. Unfortunately, these tests often require specialized equipment like force plates, which many clinicians cannot access.[36, 132] In response to this challenge, IMUs offer a more accessible alternative for gaining a deeper insight into patients' mechanics.[111] Therefore, further research must be conducted to evaluate the effectiveness of IMU measures as proxy indicators for the kinetic variables identified by recent ACLR RTS research as sensitive to limb asymmetries. [36, 132] This research can facilitate more widespread implementation of accurate assessment methods and improve ACLR patient outcomes.

Chapter 3: Methods

This study aimed to identify whether correlations exist between force plate-derived kinetic variables and IMU-derived metrics during various movements conducted in the advanced stages of ACLR return-to-sport rehabilitation. The second aim of this study was to develop and evaluate predictive models for the force plate-derived kinetic variables with the IMU-derived metrics using multilevel linear regressions.

Participants

Participants were recruited through flyers, emails, or word of mouth. For participants recruited via word of mouth or flyer, an email containing a pre-approved email script was sent to ensure they were still interested and qualified to participate in the study. An a priori power analysis (G*Power, v3.1.9.1) was performed to determine the appropriate sample size for this study. Using data from existing literature[47], 29 participants were determined to be adequate to achieve 80% power at a statistical significance criterion of 0.05 with a large effect size (f=0.40) using Cohen's f. Therefore, 43 healthy, recreationally active participants (thirteen females and 30 males) between 18 and 35 years (age:23.6±3.2 yrs., height:69.5±3.6 in.) were recruited from the university's student body and surrounding community. Recreationally active was defined as being physically active at least three days per week for at least 30 minutes each session. Inclusion criteria included no history of lower extremity surgical repair, no lower extremity injuries within the past six months, and no lower extremity pain on the day of testing. The University Institutional Review Board approved all procedures before the start of the study, and all participants were provided informed consent. Participants provided their age, gender, and leg dominance, which were noted on a data collection sheet. Later, their height was measured, and their weight was recorded during a static trial.

Experimental Protocol

The experimental protocol was conducted within a single testing session lasting approximately 45 minutes, which took place in the Biomechanics/Sports Medicine Laboratory at the University of Tennessee, Knoxville. Before starting any testing procedures, participants were instructed to wear appropriate athletic clothing for comfort and mobility during the study. Upon arrival at the laboratory, each participant provided written informed consent and completed two questionnaires: a musculoskeletal health history questionnaire to assess any existing musculoskeletal conditions and a fitness history questionnaire to gather information about their current physical activity experiences.

After completing the necessary paperwork, participants started a standardized warm-up routine to prepare for the experimental procedures.[134, 154] The warm-up consisted of a 2-minute jog on a treadmill at a self-selected speed and five bodyweight squats to engage lower body muscles further and ensure readiness for the subsequent testing protocols. Then, using molded straps, participants were fitted by the same researcher with one IMU on each mediodistal tibia, superior to the medial malleolus. Previous research has demonstrated this location to be effective in assessing tibial shock compared to other sites due to the proximity to the bone, which offers a more accurate representation of segmental acceleration.[47, 155]

After completing the warm-up, participants familiarized themselves with the experimental task on the force plates. Before the collection of any data for each task, participants were required to undergo a familiarization phase consisting of 5 trials: 2 submaximal attempts and 3 maximal attempts. The experimental tasks included the Countermovement Jump, Drop Cut, and Drop Jump. To mitigate order effects, the order of the tasks was randomized for each participant.

A 31-centimeter platform was placed half the distance of the participant's height and used to jump off for the drop jump and drop cut. For the drop jump, participants were instructed to roll from the platform with both feet together and, on hitting the force place, immediately jump as high as possible while spending as little time as possible on the force plate. For the drop cut, participants were guided to roll off the platform with one foot in the air and use that foot to land on the force plate. Upon their foot contacting the force plate, they were instructed to execute a 45-degree cut in the designated direction immediately. For the countermovement jump, participants were instructed to stand fully upright and remain motionless for a minimum of 1 second before the initiation of the test. Participants were instructed to countermove quickly and jump as high as possible.

For the Countermovement and Drop Jump tasks, a trial was deemed successful when the participant maintained both feet on the force plate throughout the take-off and landing phases while maintaining balance for a minimum of 2 seconds upon landing. For the Drop Cut task, success in a trial was determined by the participant's ability to execute a landing with their entire foot placed on the force plate and execute the cut within predefined boundaries.

Participants were required to succeed in 5 trials for each task, with a 30-second intertrial break and a 3-minute interval between tasks. Participants were also be asked to provide their subjective exertion assessment by reporting their Rate of Perceived Exertion (RPE) before and after completing each task.

Instrumentation

Two 60x60 cm AMTI Force Platforms (2000 Hz, BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to measure ground reaction force data. Both force plates were zeroed and calibrated to ensure no excess noise and a proper baseline was present during data collection. During testing, an IMU with a high-g accelerometer (1600 Hz; Vicon Blue Trident, Vicon Motion Systems Ltd, Oxford, UK) was used to measure 3D linear accelerations and gyroscopic velocities at the mediodistal tibia. This data was then imported into MATLAB R2022b (MATLAB, MathWorks, Natick, MA, USA) computing software for data processing and analysis using Matlab and JMP Pro 17(JMP®, Version 17. SAS Institute Inc., Cary, NC, USA).

Data Analysis

GRF data was imported into MATLAB R2022b, where data conditioning was performed. The following variables were extracted for each jumping task: eccentric deceleration impulse, concentric impulse, force at zero (vertical) velocity, peak landing force, loading rate, contact time, and jump height. For the cutting task, the following variables were extracted: impulse, peak landing force, contact time, and loading rate. [136, 156]

First, the static trial was used to calculate the average weight and the standard deviation of that weight. This was used in each of the experimental trials, where the vertical ground reaction force for both legs was combined to give total force to find universal events such as movement onset, flight start, minimum velocity (start of the braking phase), the first instance of zero velocity (start of the concentric phase), and landing index (first instance that vertical GRF exceeded 10 N after take-off). The velocity-time curve was found by integrating the vertical GRF time curve.

For countermovement jumps, movement onset was defined as the instant the vertical GRF was less than body weight, while drop jumps and drop cuts were defined as the initial contact with the force plate when the vertical GRF exceeded 10 N. The contact time was calculated on each side as the time between the combined onset of the movement and take off time for that specific limb. The start of the braking phase was defined as the point of minimum vertical velocity between the movement's combined onset and take-off. The start of the concentric phase was defined as the point of zero velocity between the same time intervals. The eccentric impulse was calculated using the trapezoidal rule to integrate vertical GRF from the start of the braking phase to the start of the concentric phase (zero velocity). The concentric impulse was calculated using the same method, but the time interval was between the combined start of the concentric phase and take off. For the cutting tasks, an overall impulse was calculated using the same method from the start of the braking phase to take off.

The peak landing force was identified as the maximum vertical ground reaction force on each side after the landing index. With this information, the loading rate was calculated on each side by dividing the peak landing force by the time elapsed from landing to the peak landing force occurrence. Finally, jump height was calculated by dividing the combined take off velocity squared by two times gravity.

Raw data was imported from the IMU sensors into MATLAB, where initial data conditioning was performed. The first step involved the application of a 4th order Butterworth low-pass filter, set to a cutoff frequency of 50 Hz, to both the linear acceleration and gyroscopic velocity data. For the linear acceleration data, there are two types of accelerometers to consider: the 'low-g' accelerometer, which has a saturation limit of 16g (units of gravity), and the 'high-g' accelerometer, which can measure up to 200g without saturating. In instances where the low-g accelerometer's data reaches its saturation point at 16g, these specific time points were replaced with the corresponding readings from the high-g accelerometer. Following this, the resultant linear acceleration was calculated for each leg.

Since the force plate and IMU data were synchronized and captured at the same frame rate, the starting point of the movement calculated in MATLAB from the force plate data was also used as the starting point for the IMU data. From there, the following metrics were identified for each trial: peak tibial acceleration, area under the linear acceleration curve, peak gyroscopic velocity, and area under the gyroscopic velocity curve. [47] The area under the acceleration and gyroscopic velocity curves was measured between the first landing and subsequent take-off, while the peak linear accelerations and gyroscopic velocities were identified as the highest values observed throughout the entire trial.

Statistical Analysis

The data collected in this study included a range of metrics from the IMU and force plate measurements during each experimental task. IMU metrics comprised peak linear acceleration, area under the linear acceleration curve, peak gyroscopic velocity, and area under the gyroscopic velocity curve. Force plate metrics included peak vertical GRF, eccentric deceleration impulse, concentric impulse, force at zero velocity, peak landing force, loading rate, contact time, and jump height.

The Pearson product-moment correlation examined the linear relationship between the accelerometer and GRF metrics. This analysis provided insights into how changes in accelerometer readings are associated with changes in GRF measurements. For each pair of accelerometers and GRF metrics, a Pearson correlation coefficient (r) was calculated, along with a p-value, to assess the statistical significance of the correlation. Correlations were interpreted as negligible ($-0.3 > r < 0.3$), weak ($0.3 \leq r < 0.5$ or $-0.3 \geq r > -0.5$), moderate ($0.5 \leq r < 0.7$ or $-0.5 \geq r > -0.7$), or strong ($r \geq 0.7$or $r \leq -0.7$).[47]

A multilevel linear regression model was employed to investigate the predictive capability of the IMU metrics on force plate metrics, accommodating the hierarchical data structure resulting from multiple trials per participant. Considering the repeated measures on the same subjects, each participant was treated as a random blocking factor to account for intra-subject variability. This approach allowed the model to predict the data more precisely, capturing both the fixed effects of the accelerometer metrics and the random effects associated with individual differences among participants. Separate regression models were developed for each GRF metric in each condition as the dependent variable, with the various accelerometer metrics serving as independent variables. This method ensured that the specific characteristics of each GRF metric and condition were accurately represented.

IMU metrics were incorporated as fixed continuous factors, with all metrics initially included in the model. Non-significant IMU variables were systematically removed through backward selection, retaining only those with p-values less than or equal to 0.05 in the final model. The normality for the residuals was evaluated using the Shapiro-Wilk test and visually inspected through quantile-quantile (QQ) plots to ensure compliance with the fundamental assumptions of linear regression. Variables that were found to be skewed were normalized through square root transformations.

In each model, the overall fit was assessed through R^2 values and analyze the significance of individual predictors using p-values and beta coefficients. The following formula was used to calculate the R^2 value for the models:

$$R^2 = \frac{Residual_{intercept} - Residual_{full}}{Residual_{intercept}}$$

First, an intercept-only model was fit, which includes only the intercept term and no predictors. The residual sum of squares from this model captures the variance in the outcome variable that is not explained by any predictors. Next, the full model was fit, which included the intercept and all the predictors (IMU variables) and possibly random effects (subjects). The residual sum of squares from the full model represents the variance in the outcome variable that remains unexplained after accounting for the predictors.

Chapter 4: Results

Correlations

The countermovement jump had no strong correlations. Two moderate correlations were the loading rate with max linear acceleration ($r= 0.628$, $p<0.01$) and max gyroscopic velocity ($r=0.501$, $p<0.01$). There were five weak positive correlations: peak landing force with max linear acceleration ($r= 0.412$, $p<0.01$) and max gyroscopic velocity ($r=0.307$, $p<0.01$), as well as the area under the linear acceleration curve with eccentric impulse ($r=0.366$, $p<0.01$), force at zero velocity ($r=0.387$, $p<0.01$), and contact time ($r=0.342$, $p<0.01$). The rest of the correlations were negligible.

The drop jump had one strong positive correlation: contact time with the area under the linear acceleration curve ($r=0.723$, $p<0.01$). One moderately strong correlation was the loading rate with max linear acceleration ($r= 0.552$, $p<0.01$). There were four weak positive correlations: max jump height with max linear acceleration ($r=0.332$, $p<0.01$) and max gyroscopic velocity ($r=0.351$, $p<0.01$), loading rate with max gyroscopic velocity ($r=0.425$, $p<0.01$), and contract time with the area under the gyroscopic velocity curve ($r=0.487$, $p<0.01$). There were also two weak negative correlations: force at zero velocity with the area under the linear acceleration curve ($r=-0.366$, $p<0.01$) and area under the gyroscopic velocity curve ($r=-0.38$, $p<0.01$). The rest of the correlations were negligible.

The drop cut had no strong or moderate correlations. There were six instances of weak positive correlations: Impulse with the area under the linear acceleration curve ($r=0.39$, $p<0.01$) and area under the gyroscopic velocity curve ($r=0.31$, $p<0.01$), contact time with the area under the linear acceleration curve ($r=0.48$, $p<0.01$) and area under the gyroscopic velocity curve

($r=0.42$, $p<0.01$), and loading rate with max linear acceleration ($r=0.38$, $p<0.01$) and the area under the linear acceleration curve ($r=0.342$, $p<0.01$). The rest of the correlations were negligible.

Linear Regressions

The variables eccentric impulse and loading rate exhibited skewness for both jumping tasks. To address this, a square root transformation was applied to these variables to normalize their distributions.

Eccentric Impulse

The model predicting eccentric impulse for the countermovement jump included the area under the acceleration curve, maximum gyroscopic velocity, and the area under the gyroscope curve as significant predictors. The intercept was estimated at 6.97 ± 0.66 ($p < 0.001$). The coefficient for the area under the acceleration curve was 0.11 ± 0.09 ($p < 0.001$), the maximum gyroscope was -0.001 ± 0.00043 ($p < 0.05$), and the coefficient for the area under the gyroscope curve was 0.038 ± 0.008 ($p < 0.001$). The R^2 value for this model was 0.656, indicating that the model explained approximately 65.6% of the variance in eccentric impulse (Table 1).

For the drop jump, the model predicting eccentric impulse included the area under the gyroscope curve as the significant predictor. The intercept was estimated at 13.52 ± 0.44 ($p < 0.001$). The coefficient of the area under the gyroscope curve was 0.095 ± 0.023 ($p < 0.001$). The R^2 value for this model was 0.992, indicating that the model explained approximately 99.2% of the variance in eccentric impulse (Table 2).

Concentric Impulse

The model predicting concentric impulse for the countermovement jump included the area under the acceleration curve as the significant predictor. The intercept was 599.09±35.95 ($p < 0.001$), with a coefficient for the area under the acceleration curve of −9.42±1.53 ($p < 0.001$). The R^2 value was 0.676, suggesting that 67.6% of the variance in concentric impulse was explained by the model (Table 1).

For the drop jump, the model predicting concentric impulse included the area under the acceleration curve and the area under the gyroscopic curve as significant predictors. The intercept was 173.49±14.26 ($p < 0.001$), with a coefficient for the area under the acceleration curve of 4.79±1.68 ($p < 0.001$), and the coefficient for the area under the gyroscope curve was -0.45±0.178 ($p<0.05$). The R^2 value was 0.89, suggesting that 89% of the variance in concentric impulse was explained by the model (Table 2).

Impulse

The impulse model for the drop cut included the area under the acceleration curve as the significant predictor. The intercept was 363.15±21.9 ($p < 0.001$), and the coefficient for the area under the acceleration curve was 7.38±1.053 ($p < 0.001$). The R^2 value for this model was 0.958, indicating that the model explained 95.8% of the variance in impulse (Table 3).

Force at Zero Velocity

The force at zero velocity model for the countermovement jump included the area under the acceleration curve and maximum gyroscopic velocity as significant predictors. The intercept was 318.57±34.75 ($p < 0.001$). The coefficient for the area under the acceleration curve was 9.97±1.409 ($p < 0.001$), and for the maximum gyroscope, it was 0.06±0.029 ($p < 0.05$). The R^2

value for this model was 0.523, indicating that the model explained 52.3% of the variance in force at zero velocity (Table 1).

For the drop jump, the force at zero velocity model included the area under the acceleration curve as the significant predictor. The intercept was 932.37±57.91 ($p < 0.001$), and the coefficient for the area under the acceleration curve was -8.01±3.48 ($p < 0.05$). The R^2 value for this model was 0.857, indicating that the model explained 85.7% of the variance in force at zero velocity (Table 2).

Peak Landing Force

The peak landing force model for the countermovement jump included maximum acceleration as a significant predictor. The intercept was 1622.56±117.03 ($p < 0.001$), and the coefficient for maximum acceleration was 1.478±0.28 ($p < 0.001$). The R^2 value was 0.776, showing that 77.6% of the variance in peak landing force was explained by the model (Table 1).

For the drop jump, the peak landing force model included maximum acceleration as a significant predictor. The intercept was 1570.93±126.65 ($p < 0.001$), and the coefficient for maximum acceleration was 1.85±0.35 ($p < 0.001$). The R^2 value was 0.774, showing that 77.4% of the variance in peak landing force was explained by the model (Table 2).

For the drop cut, the peak landing force model included the area under the acceleration curve as a significant predictor. The intercept was 1682.98±150.03 ($p < 0.001$), and the coefficient for the area under the acceleration curve was 44.32±10.96 ($p < 0.001$). The R^2 value was 0.804, showing that 80.4% of the variance in peak landing force was explained by the model (Table 3).

Contact Time

The model predicting contact time for the countermovement jump included the area under the acceleration curve as a significant predictor. The intercept was 1.44±0.08 ($p < 0.001$), with a coefficient for area under the acceleration curve of 0.022±0.0036 ($p < 0.001$). The R^2 value was 0.586, indicating that the model explained 58.6% of the variance in contact time (Table 1).

For the drop jump, the model predicting contact time included maximum acceleration, the area under the acceleration curve, and maximum gyroscopic velocity as significant predictors. The intercept was 0.30±0.029 ($p < 0.001$). The coefficient for the maximum acceleration was 0.00009±0.000043 ($p>0.05$), the area under the acceleration curve had a coefficient of 0.026±0.0018 ($p < 0.001$), and max gyroscopic velocity had a coefficient of -0.000031±0.000011 ($p<0.01$). The R^2 value was 0.893, indicating that the model explained 89.3% of the variance in contact time (Table 2).

For the drop cut, the model predicting contact time included maximum acceleration, the area under the acceleration curve, and maximum gyroscopic velocity as significant predictors. The intercept was 0.3±0.017 ($p < 0.001$). The coefficient for the maximum acceleration was -0.000071±0.000022 ($p>0.01$), the area under the acceleration curve had a coefficient of 0.0052±0.0016 ($p < 0.01$), and max gyroscopic velocity had a coefficient of 0.0053±0.0002 ($p<0.01$). The R^2 value was 0.896, indicating that the model explained 89.6% of the variance in contact time (Table 3).

Loading Rate

The loading rate model for the countermovement jump included maximum acceleration as a significant predictor. The intercept was 153.68±13.20 ($p < 0.001$), and the coefficient for maximum acceleration was 0.166±0.033 ($p < 0.001$). The R^2 value was 0.762, showing that 76.2% of the variance in peak landing force was explained by the model (Table 1).

For the drop jump, the loading rate model included maximum acceleration as a significant predictor. The intercept was 152.9±13.82 ($p < 0.001$), and the coefficient for maximum acceleration was 0.17±0.036 ($p < 0.001$). The R^2 value was 0.811, showing that the model explained 81.1% of the variance in loading rate (Table 2).

For the drop cut, the loading rate model included maximum acceleration as a significant predictor. The intercept was 31320.58±5799.69 ($p < 0.001$), and the coefficient for maximum acceleration was 99.14±19.47 ($p < 0.001$). The R^2 value was 0.811, showing that the model explained 81.1% of the variance in loading rate (Table 3).

Max Jump Height

The model predicting max jump height for the countermovement jump included maximum gyroscopic velocity and the area under the gyroscope curve as significant predictors. The intercept was estimated at 0.297±0.0154 ($p < 0.001$). The coefficient for the maximum gyroscopic velocity was 2.66±7.18 ($p < 0.01$), and the coefficient for the area under the gyroscope curve was 00.00052±0.00012 ($p < 0.001$). The R^2 value for this model was 0.921, indicating that the model explained approximately 92.1% of the variance in the max jump height (Table 1).

For the drop jump, the model predicting max jump height included maximum acceleration as the significant predictor. The intercept was estimated at 0.2404±0.015 ($p < 0.01$). The coefficient for the maximum acceleration was 0.000016±0.0000005 ($p < 0.01$) The R^2 value for this model was 0.944, indicating that the model explained approximately 94.4% of the variance in the max jump height (Table 2).

Chapter 5: Discussion

The first research question aimed to evaluate the effectiveness of IMU variables as surrogates for force plate variables. The analysis revealed a weak correlation between most variables measured by the IMUs and the force plate, meaning that the hypothesis was not supported. This finding indicates that, despite the theoretical potential of IMUs to capture similar biomechanical events, the specific variables selected did not align closely with those measured by the force plate. The second research question aimed to evaluate the effectiveness of IMU variables as predictors for force plate variables in multilevel linear models. Most of the regression models demonstrated high explanatory power, accounting for a significant percentage of the variance in the force plate metrics, meaning that the second hypothesis was supported.

The correlational analysis results from the current study do not align with the previous studies that evaluated the relationship between the same IMU-derived metrics and other joint kinetic metrics during dynamic tasks. A study by Sigward and Pratt found a strong correlation between peak resultant shank angular velocity and knee power and a moderate correlation between peak resultant shank angular velocity and knee extension moment.[45] Jones et al. reported strong negative correlations between the area under the tibial acceleration curve and knee extension moment and stiffness.[47]

However, others have reported only weak to moderate correlations between shank-mounted accelerometer metrics and knee moments, which more closely aligns with the results of this study.[152, 157] Ekdahl et al. examined the correlations between peak axial accelerations with peak knee abduction moment during a single-leg hop, deceleration task, and run cut and found no strong correlations.[157] Morgan & O'Connor examined similar correlations during landings and also found no correlations with knee abduction moment, but they found strong correlations between both peak anterior and posterior tibial accelerations and knee extension moment.[152]

The differences in findings between this study and the previous ones can be attributed to the fundamental distinctions between joint-specific metrics captured by the combination of force plates and motion capture systems, such as knee power, stiffness, and moments, and the overall forces exerted by the body on the ground, as captured by force plate metrics. Knee power, stiffness, and moments are all calculated using joint kinematics, which involves measurements of joint angles, angular velocities, and angular accelerations. Knee power is the product of joint moment and joint angular velocity, both derived from kinematic data. Knee stiffness, calculated as the ratio of change in joint moment to change in joint angle, and knee moments, determined by the forces applied and the moment arm derived from body segment positions, also rely on kinematic measurements.

This inclusion of joint kinematics in knee power, stiffness, and moments offers a clearer relationship with IMU metrics because these metrics are directly derived from the same type of data that IMUs capture. IMUs provide data on angular velocities and accelerations of body segments, which align closely with the kinematic measurements used in joint-specific metrics.[158, 159] This direct measurement alignment ensures that IMU data and joint-specific metrics capture the same underlying joint dynamics, leading to more precise correlations.

In contrast, force plate metrics such as impulse, loading rate, contact time, peak landing force, and force at zero velocity capture the cumulative forces exerted by the entire body during movement. These metrics provide a broader view of movement dynamics but do not isolate the contributions of individual joints. This lack of specificity means they do not provide detailed insights into the mechanics of individual joints. Consequently, the overall forces measured by force plates reflect the combined output of multiple joints and muscle groups, resulting in a broader, global perspective that does not align as closely with the specific segment-level data captured by IMUs.

An important observation in the study is the presence of significant p-values despite weak correlations. A p-value indicates the likelihood that the observed correlation occurred by chance. However, statistical significance does not necessarily imply practical significance. The magnitude of r reflects the strength and direction of a linear relationship between two variables. In these findings, while many correlations are statistically significant ($p < 0.01$), their low coefficients indicate weak relationships. This discrepancy can be caused by large sample sizes, which increase the power of the statistical test, making it easier to detect even small effects as significant.

Two studies have aimed to predict GRF metrics using linear regressions and IMUs and their results partially support this study's findings.[160, 161] Chaaban et al. evaluated simple linear regressions between each task predictor and response variables.[161] The predictor variable was then selected from each IMU with the highest R^2. This did not lend to models with good fits, with the most accurate from the shank accelerometer being the GRF, with an R^2 of 0.58. All but one model from the current study across all conditions had higher levels of accuracy than the most accurate model in the previous study’s analysis. This demonstrates the relevance of the specific IMU metrics chosen in the current study and the inclusion of random intercepts to account for individual variation.

Chaaban et al. also analyzed stepwise linear regressions to develop models using multiple-sensor, multiple-feature data in the same study. [161] These models included accelerometer metrics and combined accelerometer and gyroscope metrics. Hyperparameters were optimized through a grid search on p-value thresholds for the F-test of the change in the sum of squared errors when adding or removing terms. The final models were selected based on maximizing R^2 while minimizing the number of features, targeting a maximum of 41 features. Again, the only model that predicted a kinetic metric with accuracy was GRF, with an R^2 of 0.87. It should be noted that this study did not specify which metrics were included in any of the fitted models.[161] Even with only one accelerometer on each limb and four predictor variables, the models from the current study had similar levels of accuracy during the drop cut and jump. This is an important difference because it makes this type of analysis more accessible and less complicated.

On the other hand, the findings from the current study were more consistent with the results of Alcantara et al. [160] who aimed to predict whole-body kinetics through simple linear regressions using an accelerometer on the sacrum during running. Alcantara et al. reported using the accelerometer metrics and the runner's body mass, speed, and step frequency as predictors it was possible to predict impulse, peak vertical GRF, and contact time with an $R^2 > 0.85$. The overall fit of these models was more accurate than the same models for the countermovement jump but had very similar fits for those same models for the drop jump and drop cut. However, this study required more predictor variables, which also required additional equipment.

Previous studies predicting biomechanical variables with wearable devices often employ multiple IMUs or machine learning algorithms, such as artificial neural networks.[161-163] Despite achieving low prediction errors, these approaches can be challenging to interpret. Consequently, the high cost of using multiple IMUs or the significant computational demands of these models may limit the practical application of these findings.[160]

Certain limitations should be noted with the current study. First, the movements were performed with standardized lab shoes onto force plates. Therefore, the results may not reflect those collected from an average rehabilitation setting. Another limitation could be that the movement onset for the IMU variables was synchronized with the force plate onset. In practical settings, force plates might not be available for synchronization, limiting the generalizability and practical utility of the research. The final limitation was that IMU variables from the first landing period were used to predict force plate values across the whole trial. Having variables from the same movement phase for both devices could have led to better results.

The findings from this study demonstrate the efficacy of multilevel linear regression models in accurately describing the relationship between predictor variables and GRF variables during drop jumps and cuts. While prior research has often relied on multiple IMUs or complex machine learning algorithms like artificial neural networks to predict biomechanical variables, such approaches can be challenging to interpret despite low prediction errors and are costly to implement. The financial burden of multiple IMUs and the computational demands of advanced algorithms may limit the practical application of these methods. By utilizing a single accelerometer, this study showed that biomechanical variables relevant to ACLR rehabilitation could be predicted with high accuracy. This reduces financial costs and simplifies the methodology, making it more accessible for broader applications in practice.

Even though these models were found to be accurate, future research is necessary before implementing these protocols into practice. For example, future research should test these models in clinical populations to see if the findings hold true across diverse groups. Additionally, there should be research that adds more predictor variables to see if that improves the accuracy of the models. These predictor variables could come from data requiring minimal or no additional equipment, such as body weight. Alternatively, they could come from additional IMU variables, such as the area under the curve in each sensor for the second landings and the peaks from each of the respective landings, rather than the universal one used in this study. There should also be research comparing the accuracy of these multilevel linear regressions to other statistical models using the same variables as the current study. This will help determine whether using a different statistical model is worth the extra complexity in the analysis and the potential decrease in practicality. Finally, future studies should separate the drop cut into braking and propulsive phases as done for the other movments in this study.

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Appendices

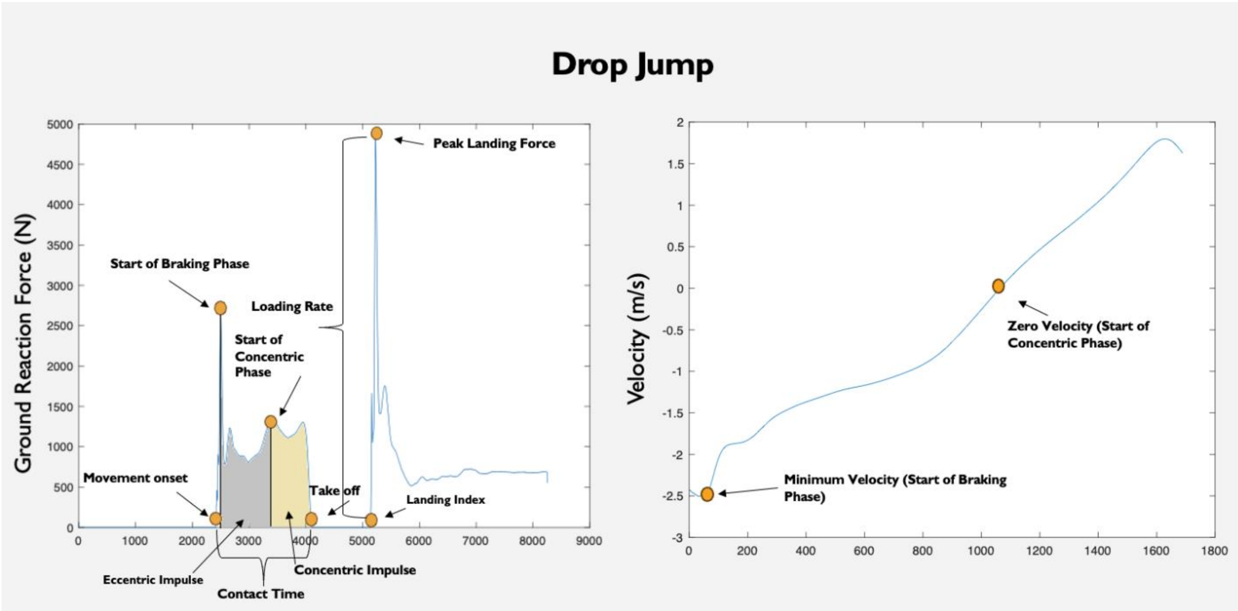

Figure 1. GRF-frame curve (a) and Velocity-frame curve (b) show how events were picked and metrics calculated in MATLAB for the Drop Jump.

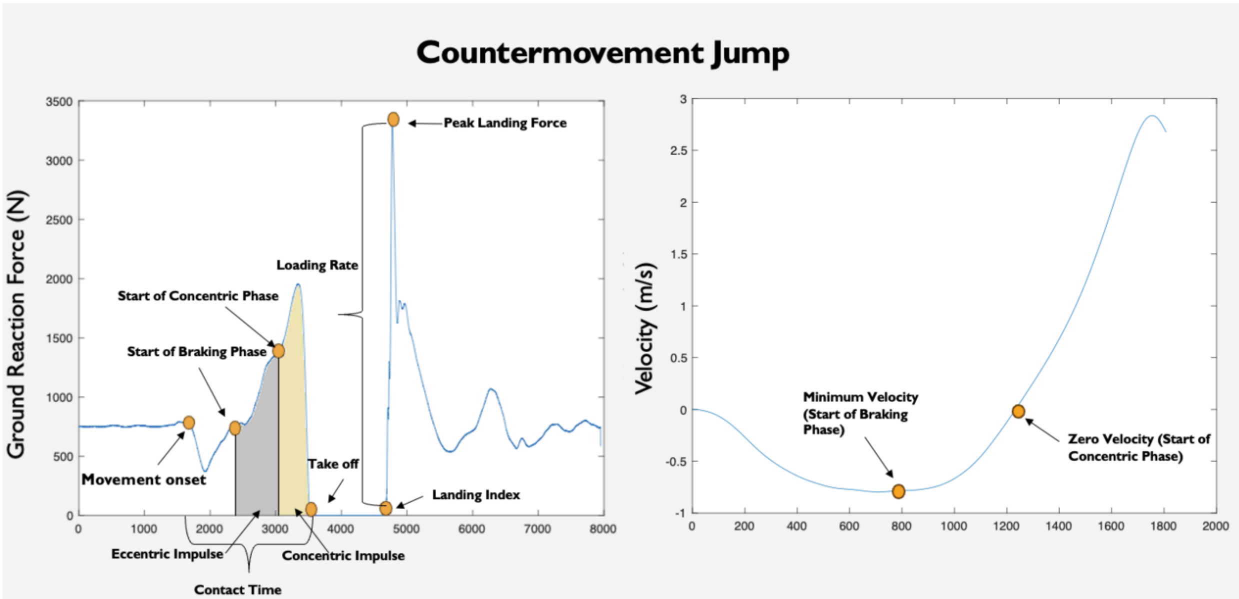

Figure 2. GRF-frame curve (a) and Velocity-frame curve (b) show how events were picked and metrics calculated in MATLAB for the Countermovement Jump.

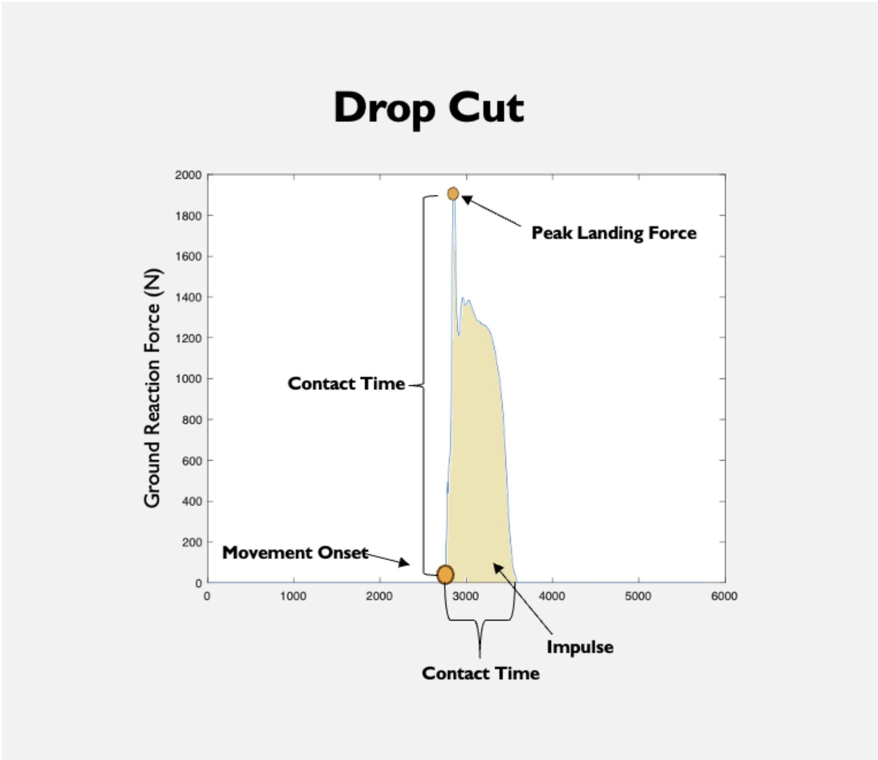

Figure 3. GRF-frame curve shows how events were picked and metrics calculated in MATLAB for the Drop Cut.

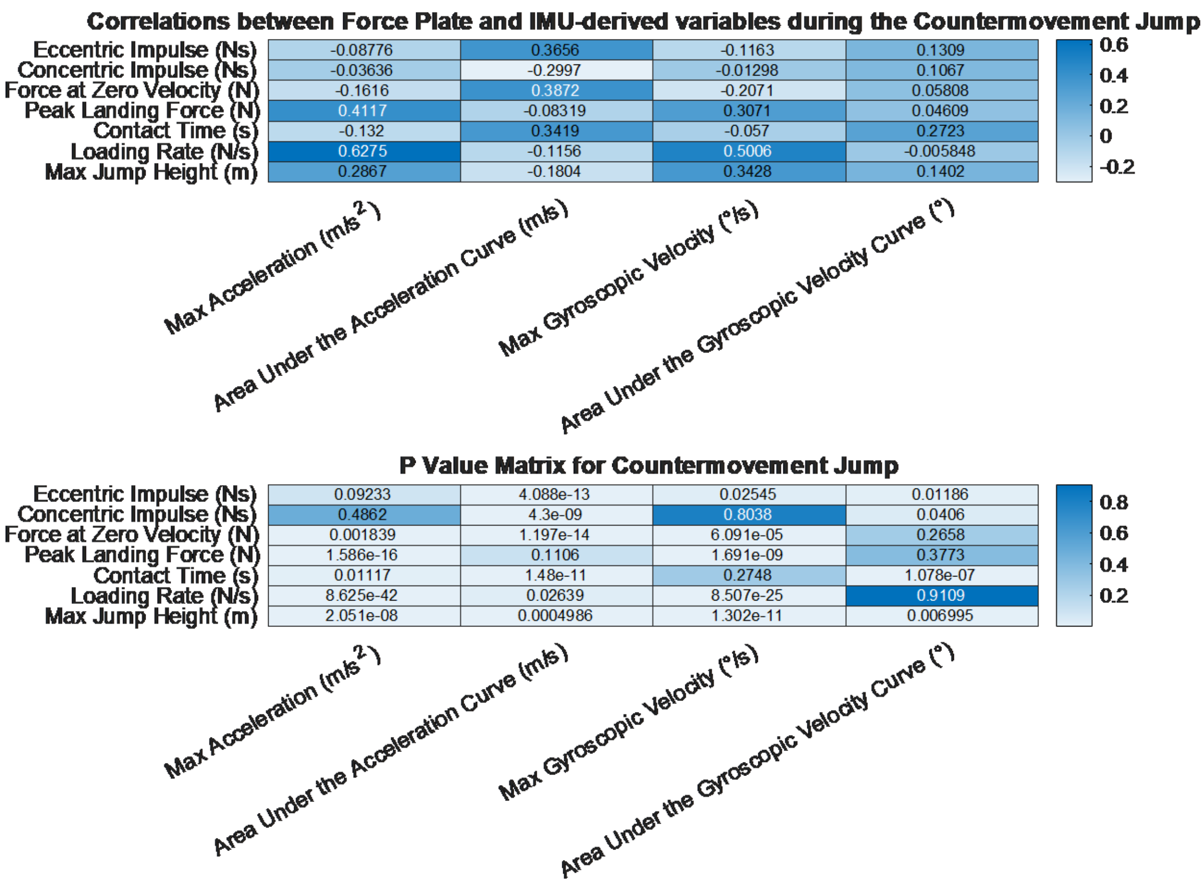

Figure 4. Correlation heatmap (a) and p-value heatmap (b) between each force plate and IMU variable for the countermovement jump.

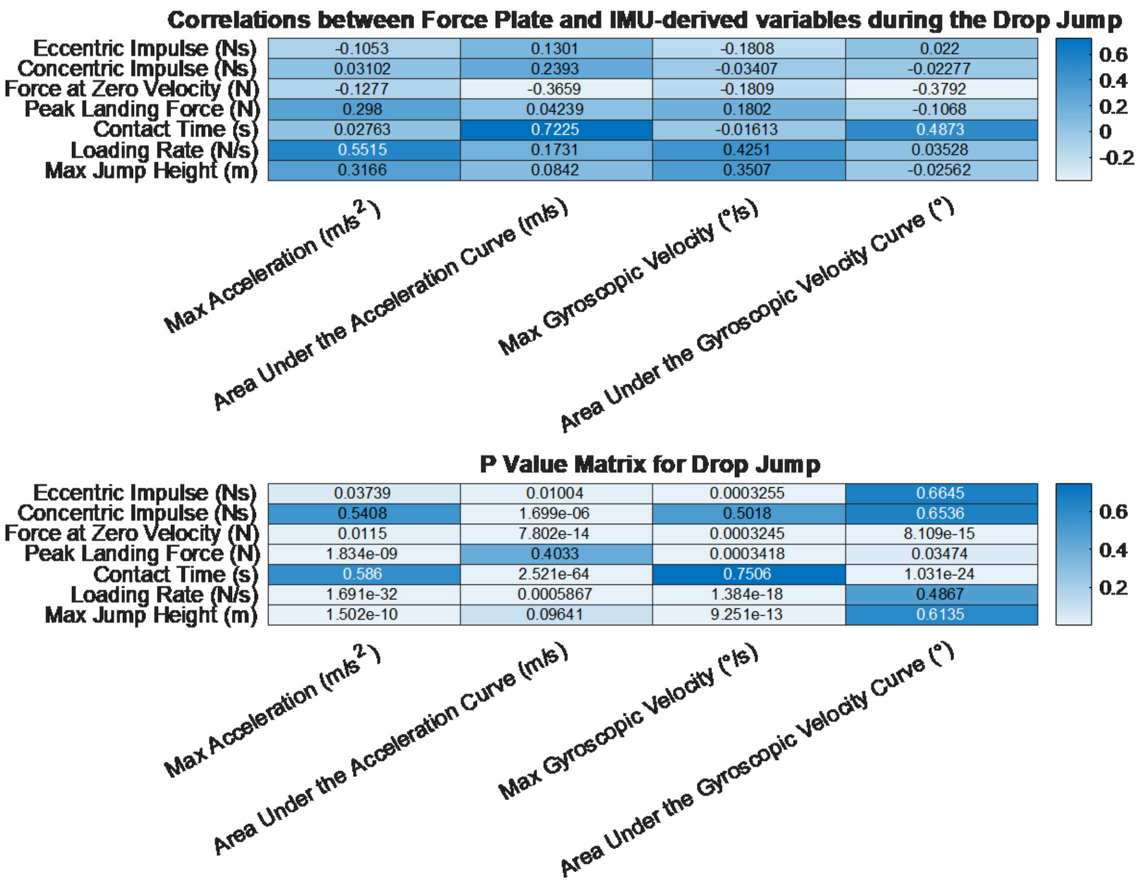

Figure 5. Correlation heatmap (a) and p-value heatmap (b) between each force plate and IMU variable for the drop jump.

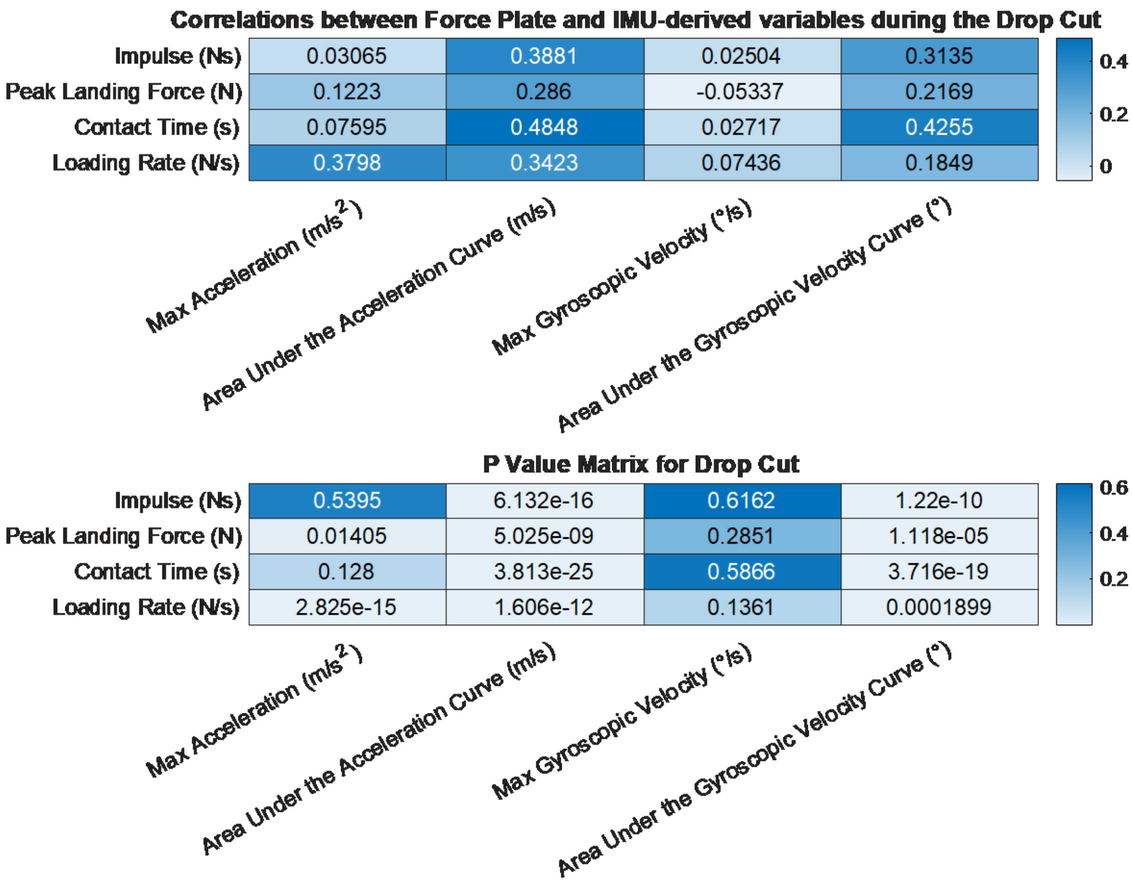

Figure 6. Correlation heatmap (a) and p-value heatmap (b) between each force plate and IMU variable for the drop cut.

Table 1. Countermovement Jump Fitted Multilevel Model Results. Coefficient ± Standard Error.

	Intercept	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve (°)	R^2
Eccentric impulse (Ns)	6.97±0.66***		0.11±0.09***	-0.001±0.00043*	0.038±0.008***	0.656
concentric Impulse (Ns)	599.09±35.95***		-9.42±1.53***			0.676
orce at Zero Velocity (N	318.57±34.75***		9.97±1.40***	0.06±0.029*		0.523
Peak Landing Force (N)	1622.56±117.03***	1.478±0.28***				0.776
Contact Time (s)	1.44±0.08***		0.022±0.0036***			0.586
Loading Rate (N/s)	153.68±13.20***	0.166±0.033***				0.762
Max Jump Height (m)	0.297±0.0154***			2.66±7.18**	0.00052±0.00012***	0.921

* $p<0.05$
** $p<0.01$
*** $p<0.001$

Table 1. Drop Jump Fitted Multilevel Model Results. Coefficient ± Standard Error

	Intercept	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve (°)	R^2
Eccentric impulse (Ns)	13.52±0.44***		0.095±0.023***			0.992
concentric Impulse (Ns)	173.49±14.26***		4.79±1.68***		-0.45±0.178*	0.890
'orce at Zero Velocity (N	932.37±57.91***		-8.01±3.48*			0.857
Peak Landing Force (N)	1570.93±126.65***	1.85±0.35***				0.774
Contact Time (s)	0.30±0.029***	0.00009±0.000043*	0.026±0.0018***	-0.000031±0.000011**		0.893
Loading Rate (N/s)	152.9±13.82***	0.17±0.036***				0.811
Max Jump Height (m)	0.2404±0.015**	0.000016±0.0000005***				0.944

* p<0.05
** p<0.01
*** p<0.001

Table 2. Drop Cut Fitted Multilevel Model Results. Coefficient ± Standard

	Intercept	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve (°)	R^2
Impulse (Ns)	363.15±21.9***		7.38±1.053***			0.958
Peak Landing Force (N)	1682.98±150.03***		44.32±10.96***			0.804
Contact Time (s)	0.3±0.017***	-0.000071±0.000022**	0.0052±0.0016**	0.0053±0.0002**		0.896
Loading Rate (N/s)	31320.58±5799.69***	99.14±19.47***				0.585

* p<0.05
** p<0.01
*** p<0.001

Table 3. Average values across all trials for each force plate variable during the drop jump and countermovement jump.

	Eccentric Impulse (Ns)	Concentric Impulse (Ns)	Force at Zero Velocity (N)	Peak Landing Force (N)	Contact Time (s)	Loading Rate (N/s)	Max Jump Height (m)
Drop Jump	223.74	208.80	821.40	1960.40	0.62	42903.21	0.25
Countermovement Jump	111.94	459.69	416.64	1906.64	1.78	42604.37	0.28

Table 4. Average values across all trials for each force plate variable during the drop cut.

	Impulse (Ns)	Peak Landing Force (N)	Contact Time (s)	Loading Rate (N/s)
Drop Cut	446.26	2181.74	0.38	50993.00

Table 5. Individual participant's average values for each force plate variable during the countermovement jump.

	Eccentric Impulse (Ns)	Concentric Impulse (Ns)	Force at Zero Velocity (N)	Peak Landing Force (N)	Contact Time (s)	Loading Rate (N/s)	Max Jump Height (m)
S01	134.64	401.73	691.80	2080.41	1.95	41233.01	0.30
S02	96.38	645.85	471.23	2689.24	1.75	133307.67	0.33
S03	133.98	464.35	283.93	1290.18	2.08	18223.60	0.28
S06	79.71	711.73	416.16	1718.90	1.94	23112.00	0.43
S07	83.21	424.96	403.22	1103.15	1.88	17643.13	0.33
S08	121.50	564.50	393.74	2691.44	1.69	52674.99	0.30
S09	118.09	683.60	577.62	2055.63	1.72	25769.69	0.39
S10	111.29	168.99	635.96	2366.75	1.53	51176.39	0.27
S11	92.08	497.28	236.56	3443.37	1.60	202892.28	0.36
S12	113.35	362.69	304.71	1091.47	1.72	12265.65	0.14
S13	79.52	289.20	457.90	1641.38	1.28	51074.37	0.23
S14	158.73	513.85	543.33	2382.02	1.66	37018.16	0.22
S15	80.09	533.39	352.69	1320.54	1.87	21306.35	0.35
S16	115.74	486.38	341.60	2626.83	1.57	46886.84	0.40
S17	101.62	271.63	460.39	1062.40	1.82	11522.26	0.18
S18	67.73	278.17	396.54	3029.18	1.16	205608.09	0.37
S20	90.22	462.05	337.54	1008.02	1.76	12361.31	0.19
S22	114.81	601.84	427.28	1898.94	1.63	24089.20	0.23
S23	80.23	690.37	439.35	2654.47	1.94	49962.99	0.38
S24	141.18	172.11	648.48	2243.01	1.61	50051.66	0.13
S25	107.48	563.10	357.98	1764.63	2.04	21986.80	0.31
S26	379.27	959.26	564.63	3257.46	2.68	60432.78	0.29
S27	102.89	289.68	359.46	1162.39	1.17	14524.83	0.20
S28	42.52	299.61	205.94	1480.56	1.37	21571.41	0.32
S29	57.73	248.81	390.06	1461.65	1.16	25684.60	0.23
S30	77.09	425.39	312.74	2356.22	1.74	45767.10	0.31
S31	75.36	408.83	315.54	1861.76	1.41	31951.31	0.34
S32	91.68	550.47	460.39	1479.18	1.68	24131.15	0.34
S33	88.01	449.25	270.80	1199.92	1.73	13180.82	0.27
S34	79.63	310.84	262.14	796.40	2.14	9129.17	0.15
S35	164.64	507.99	475.73	2062.54	2.00	38709.66	0.25
S36	160.59	371.78	527.81	2564.40	2.10	57230.44	0.32
S37	121.44	377.21	460.61	1829.55	1.67	25285.32	0.30
S38	121.69	661.56	359.69	2927.39	2.09	63513.36	0.31
S39	102.97	130.87	606.65	1160.89	2.20	15526.48	0.25
S40	70.66	526.01	374.15	1646.20	1.70	21004.21	0.31
S41	202.69	355.94	459.12	1589.59	1.79	19903.83	0.09
S43	132.01	768.61	367.47	1856.16	2.77	25797.50	0.35

Table 6. Individual participant's average values for each IMU variable during the countermovement jump.

	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve
S01	196.51	19.36	722.56	67.36
S02	225.05	12.27	803.26	66.73
S03	229.52	16.14	783.81	71.79
S06	222.11	15.38	977.08	81.30
S07	195.85	14.71	855.30	67.13
S08	208.12	14.71	761.58	89.25
S09	151.18	11.96	702.73	72.99
S10	219.04	18.81	937.22	88.63
S11	551.90	12.55	1751.98	68.60
S12	147.37	13.23	580.42	65.47
S13	239.32	11.88	654.05	69.88
S14	213.77	15.81	455.15	71.05
S15	179.12	12.98	802.33	73.58
S16	257.42	13.66	1014.72	55.11
S17	171.75	18.10	674.62	57.33
S18	462.17	12.37	1616.79	55.41
S20	137.65	13.46	519.81	59.32
S22	95.51	12.06	378.22	49.02
S23	135.48	8.42	524.90	40.40
S24	107.02	17.49	483.71	57.06
S25	100.15	11.88	552.86	67.31
S26	128.98	18.24	490.64	63.93
S27	106.69	19.43	442.88	51.00
S28	225.06	13.01	977.94	71.23
S29	175.53	12.37	616.36	52.21
S30	259.56	14.42	1100.96	73.11
S31	203.53	12.09	598.06	50.10
S32	194.94	12.91	771.66	57.99
S33	206.52	13.21	1049.27	70.00
S34	134.40	17.51	597.99	64.50
S35	202.09	16.50	890.01	63.42
S36	230.08	21.52	861.97	67.91
S37	178.81	16.52	787.36	63.34
S38	129.47	14.20	662.71	76.41
S39	126.61	21.41	540.54	70.33
S40	111.63	11.98	501.66	77.15
S41	223.95	16.52	1158.18	56.01
S43	273.54	16.79	1174.47	85.19

Table 7. Individual participant's average values for each force plate variable during the drop jump.

	Eccentric Impulse (Ns)	Concentric Impulse (Ns)	Force at Zero Velocity (N)	Peak Landing Force (N)	Contact Time (s)	Loading Rate (N/s)	Max Jump Height (m)
S01	188.036	270.112	1061.222	2392.734	0.492	49476.730	0.461
S02	271.992	279.774	850.329	2809.206	0.647	46937.618	0.313
S03	213.731	200.033	703.835	1454.137	0.778	17697.237	0.216
S06	282.393	273.861	1085.606	1724.735	0.653	23452.640	0.335
S07	166.277	202.739	900.805	1338.301	0.509	27497.013	0.386
S08	265.833	210.573	849.636	2257.636	0.586	38538.657	0.156
S09	329.560	285.089	986.927	2368.400	0.684	33037.729	0.308
S10	188.951	173.674	674.161	2553.956	0.639	53809.034	0.276
S11	227.140	259.891	802.245	3720.118	0.688	261903.208	0.399
S12	227.268	164.994	478.681	1030.858	0.798	10993.041	0.101
S13	137.979	187.055	690.313	1496.549	0.590	43188.762	0.302
S14	413.664	284.395	893.499	2594.868	0.875	38782.605	0.215
S15	194.725	222.686	853.670	1164.995	0.630	17553.999	0.316
S16	296.434	231.201	820.450	2487.983	0.697	38523.684	0.309
S17	198.273	189.106	521.819	963.972	0.755	10529.628	0.154
S18	175.863	189.449	678.313	2767.154	0.616	174450.632	0.342
S20	185.982	223.078	649.623	1126.161	0.712	14373.718	0.213
S21	258.948	238.624	1176.401	1815.903	0.473	23665.725	0.196
S22	208.511	210.286	1416.426	2988.011	0.346	59444.755	0.309
S23	258.143	178.324	636.732	2106.538	0.806	53309.703	0.145
S24	211.267	211.510	685.609	1716.920	0.670	22132.192	0.259
S25	345.609	320.646	1269.905	2860.162	0.724	47374.208	0.271
S26	218.922	192.908	640.394	1134.088	0.762	16171.979	0.132
S27	158.783	171.001	654.251	1530.920	0.684	21231.350	0.304
S28	167.729	173.360	785.846	1680.738	0.574	29070.364	0.247
S29	197.472	171.883	915.334	2208.398	0.523	43018.450	0.246
S30	192.550	183.513	842.006	2631.346	0.478	92477.506	0.308
S31	212.782	221.198	931.261	1808.687	0.477	35392.290	0.352
S32	189.416	193.824	706.375	1336.508	0.651	15222.724	0.256
S33	161.978	148.930	430.537	882.363	0.848	11121.978	0.126
S34	220.056	213.342	990.800	2456.916	0.512	56021.368	0.248
S35	198.762	243.240	878.099	2368.477	0.549	51250.086	0.343
S36	202.879	164.573	1001.511	1789.014	0.437	30387.713	0.178
S37	222.411	224.673	874.467	2943.365	0.607	63855.687	0.259
S38	138.215	120.285	662.698	1230.958	0.453	15871.114	0.137
S39	206.628	208.470	941.331	1665.231	0.514	23551.137	0.295
S40	353.354	108.096	620.068	1650.440	0.543	14117.207	0.001
S42	218.665	198.414	736.631	1699.458	0.645	23895.914	0.251
S43	218.665	198.414	736.631	1699.458	0.645	23895.914	0.251

Table 8. Individual participant's average values for each IMU variable during the drop jump.

	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve
S01	211.473	13.457	961.122	59.891
S02	202.019	14.337	749.216	68.509
S03	208.757	15.769	831.259	86.073
S06	217.313	14.987	906.084	76.856
S07	223.322	12.410	963.382	69.444
S08	215.293	14.507	863.899	75.907
S09	175.173	13.507	647.925	74.784
S10	208.422	14.877	919.081	79.829
S11	404.679	16.604	1628.996	83.715
S12	174.875	15.105	671.370	80.895
S13	270.222	14.414	1145.635	66.506
S14	244.072	17.467	647.657	72.135
S15	209.183	14.841	659.770	73.021
S16	231.206	15.487	1059.448	64.977
S17	199.102	15.917	712.416	61.751
S18	475.575	15.237	1601.501	68.774
S20	191.594	15.867	733.388	70.206
S21	115.250	9.418	485.346	38.893
S22	137.710	9.220	570.319	48.786
S23	114.175	14.052	462.724	59.118
S24	113.978	12.172	612.546	69.008
S25	132.627	14.482	468.994	70.895
S26	155.148	14.350	577.318	77.320
S27	218.721	15.257	995.755	83.278
S28	240.855	14.220	858.197	64.534
S29	238.636	12.310	944.569	71.157
S30	276.803	13.771	779.081	52.730
S31	236.765	12.868	972.954	66.971
S32	191.384	14.394	945.469	79.405
S33	209.927	15.793	776.538	73.904
S34	236.701	13.009	756.559	55.624
S35	222.627	14.357	905.600	72.589
S36	170.070	12.247	706.566	59.935
S37	141.748	13.706	561.035	68.991
S38	118.354	9.725	530.326	70.877
S39	131.897	10.259	611.077	60.785
S40	216.394	10.832	886.247	69.803
S42	265.553	14.375	1311.751	81.981
S43	265.553	14.375	1311.751	81.981

Table 9. Individual participant's average values for each force plate variable during the drop cut.

	Impulse (Ns)	Peak Landing Force (N)	Contact Time (s)	Loading Rate (N/s)
S01	512.07	2178.90	0.36	23267.70
S02	584.94	2993.56	0.44	86962.46
S03	342.68	1808.41	0.36	51089.57
S04	229.27	1740.72	0.22	42537.53
S05	788.02	3910.16	0.54	95445.55
S06	868.38	2553.94	0.75	54034.07
S07	379.27	1889.97	0.33	43082.63
S08	480.17	2841.68	0.37	62737.65
S09	680.48	1905.65	0.55	72272.31
S10	416.05	1971.66	0.40	79328.08
S11	409.09	1844.95	0.35	31686.89
S12	344.22	1723.04	0.38	35047.46
S13	376.45	2421.72	0.41	147627.57
S14	581.66	2767.51	0.43	81638.33
S15	384.14	1597.73	0.37	25074.34
S16	553.43	2169.94	0.37	15669.76
S17	340.70	1826.03	0.35	35637.34
S18	377.40	1854.31	0.35	90083.85
S19	333.31	2253.18	0.34	50187.21
S20	384.58	2164.00	0.34	98950.52
S22	435.80	2817.16	0.28	32594.74
S23	530.68	2111.77	0.35	22791.14
S24	397.17	1793.74	0.39	35322.43
S25	466.00	2449.71	0.40	54872.49
S26	551.48	3394.24	0.34	70498.00
S27	328.33	1712.35	0.34	31877.07
S28	307.62	1679.37	0.34	31769.86
S29	377.14	1744.78	0.37	29133.40
S30	406.96	1837.81	0.34	20452.89
S31	511.91	3530.72	0.45	122456.87
S32	436.94	2079.09	0.34	23548.00
S33	355.45	1640.80	0.33	12968.28
S34	308.77	1165.62	0.44	20124.23
S35	501.32	2229.39	0.38	60537.09
S36	493.18	2747.38	0.39	70624.39
S37	490.80	2241.06	0.42	47595.41
S38	426.35	2103.07	0.35	38594.69
S39	370.38	1841.21	0.42	43047.51
S40	425.13	1951.20	0.31	23341.72
S41	386.92	1697.21	0.34	17719.25
S42	501.87	2711.11	0.37	82326.64
S43	391.20	1893.39	0.34	25351.20

Table 10. Individual participant's average values for each IMU variable during the drop cut.

	Max Acceleration (m/s^2)	Area Under the Acceleration Curve (m/s)	Max gyroscopic Velocity (°/s)	Area Under the Gyroscope Curve
S01	208.24	8.27	1106.46	65.77
S02	319.18	14.39	1124.90	80.74
S03	203.75	12.16	841.58	73.11
S04	186.20	6.56	793.26	39.88
S05	209.98	13.60	994.23	90.13
S06	228.56	15.99	1042.42	103.74
S07	229.64	12.36	1177.69	71.08
S08	177.31	10.56	769.49	61.83
S09	129.24	10.23	663.51	76.83
S10	275.73	13.31	987.91	74.72
S11	243.80	13.50	1098.35	80.09
S12	180.50	10.79	730.58	64.63
S13	409.72	13.90	1245.17	72.90
S14	205.12	12.70	761.83	74.88
S15	205.98	12.98	962.37	81.53
S16	179.87	12.99	1031.28	79.60
S17	189.77	11.85	898.78	60.11
S18	252.21	12.00	1265.70	65.42
S19	151.40	8.32	669.75	142.15
S20	210.10	10.90	748.32	60.35
S22	108.73	9.44	767.39	57.93
S23	122.45	7.17	629.97	41.62
S24	100.58	9.23	649.86	60.44
S25	114.56	10.73	728.43	76.40
S26	128.55	10.12	618.25	59.85
S27	131.39	9.22	627.11	56.72
S28	210.58	12.83	1101.71	79.23
S29	161.02	10.76	793.89	60.21
S30	341.91	11.99	1556.36	76.19
S31	279.39	14.34	823.83	75.22
S32	164.60	11.47	845.13	64.63
S33	175.54	10.12	754.68	55.08
S34	201.50	12.26	950.05	71.99
S35	197.74	12.14	866.39	69.47
S36	293.69	14.10	1115.56	89.40
S37	168.41	13.16	816.72	72.27
S38	110.11	9.53	680.21	63.90
S39	105.34	7.99	527.23	54.53
S40	143.59	10.06	723.22	49.56
S41	194.66	8.10	1177.63	66.70
S42	232.20	12.01	887.40	67.96
S43	227.84	11.75	1366.05	68.20

Appendix E. Informed Consent

Consent for Research Participation

Research Study Title: Inertial Measurement Units as Proxy Measures for Knee Kinetic Variables Associated With ACL Re-Injury

Researcher(s): Joshua T. Weinhandl, PhD, University of Tennessee, Knoxville
Kavon Bonakdar, BS, University of Tennessee, Knoxville

Why am I being asked to be in this research study?

We are asking you to be in this research study because you are between the ages of 18 to 35 years old and are recreationally active. We also ask you to be in this research study because you are healthy. We define being recreationally active as being physically active at least 3 times per week for a minimum of 30 minutes each session. Individuals who are not between the ages of 18 to 35, not recreationally active 3 times a week, or have sustained a lower limb injury within the past year will not be asked to be in this research study.

How long will I be in the research study?

If you agree to be in the study, your participation will involve 1 study visit lasting approximately 45 minutes.

What will happen if I say "Yes, I want to be in this research study"?

If you agree to participate in this study, we will ask you to come to the Biomechanics Lab for one testing session.

Preparation:

1. You will fill out a fitness activity questionnaire and a musculoskeletal health history form to determine if you can participate in the study. The researcher will also record your height, weight, and date of birth on a data collection sheet.
2. You will then change into compression shorts and generic T-shirt, which will be provided if you do not have them.
3. Your height and weight will be recorded.
4. You will then complete a 2-minute warm-up jog at a self-selected pace and 5 bodyweight squats.
5. After the warm-up, a strap with a sensor attached will be wrapped on the top of your ankles to track your movement for each trial.

Testing Procedures:

1. You will complete 5 familiarization trials before each of the tasks.
2. You will then complete 5 successful trials of the jump and cut tasks.
    - Drop Jump Task: You drop from a box and immediately jump as high as possible.
    - Drop Cut Task: You will drop from a box on to your testing limb, and immediately cut to the opposite side.
    - Countermovement Jump Test: You will stand with your feet on the force plate, squat to a self-selected depth and then jump as high as possible.
3. The testing session will then be completed.

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What happens if I say “No, I do not want to be in this research study”?

Being in this study is up to you. You can say no now or leave the study later. Either way, your decision won’t affect your grades, your relationship with your instructors, or standing with the University of Tennessee, Knoxville.

What happens if I say “Yes” but change my mind later?

Even if you decide to be in the study now, you can change your mind and stop at any time. If you decide to stop before the study is completed, you can tell the PI and/or co-PI that you want to withdraw from the study at any time. If you decide to withdraw from the study, your information will be kept de-identified and kept in a locked drawer in the Biomechanics lab. Only study personnel will have access to any forms and data that have already been collected.

Are there any possible risks to me?

It is possible that someone could find out you were in this study or see your study information, but we believe this risk is small because of the procedures we use to protect your information. These procedures are described later in this form.

Possible risks include lower extremity injury during the landings of the study movements. These risks will be minimized. You will complete a required warm-up before data collection, so your muscles are ready to move and provided ample opportunity to familiarize yourself with the movements to reduce the risk of injury. Further, breaks will be provided between each trial and task. The movements included are movements you should be familiar with, due to you being recreationally active. In the *very* unlikely case that you hurt yourself during the study visit, emergency services will be contacted immediately for assistance. If you realize afterward that you have become injured, you should seek medical assistance.

Are there any benefits to being in this research study?

We do not expect you to benefit from being in this study. Your participation may help us learn more about the relationship between limb accelerations collected from inertial measurement units are similar to ground reaction forces measured from a force plate. This information will hopefully confirm that using these sensors is appropriate in the rehabilitation setting.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by keeping all forms in a locked drawer or on a password-encrypted computer drive in the Biomechanics lab, which is locked every day. Only the investigators conducting this study will have access to your personal information. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used. We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information, we collect about you. These include people at the University of Tennessee, Knoxville who oversee research to make sure it is conducted properly. Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of

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the research. If a law or court requires us to share the information, we would have to follow that law or final court ruling.

What will happen to my information after this study is over?

We will keep your information to use for future research. Your name and other information that can directly identify you will be kept secure and stored separately from your research data collected as part of the study. We will not share your research data with other researchers.

Will I be paid for being in this research study?

You will not be paid for being in this study.

Will it cost me anything to be in this research study?

It will not cost you anything to be in this study.

What else do I need to know?

We may need to stop your participation in the study without your consent if you do not follow the study instructions, no longer meet the study's eligibility requirements, if your safety comes into question, or if the study is stopped for any reason.

The University of Tennessee does not automatically pay for medical claims or give other compensation for injuries or other problems should you realize you are injured outside of the study visit.

Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Dr. Joshua Weinhandl (**jweinhan@utk.edu**, 865-974- 9556) or Kavon Bonakdar (**kbonakdar@vols.utk.edu**, 865-974-2091)

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board

The University of Tennessee, Knoxville

Phone: 865-974-7697

Email: **utkirb@utk.edu**

STATEMENT OF CONSENT

I have read this form, and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

Name of Adult Participant	Signature of Adult Participant	Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

Name of Research Team Member	Signature of Research Team Member	Date

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Appendix F. Fitness Activity Questionnaire

SUBJECT: ________ DATE: ________________

FITNESS ACTIVITY QUESTIONNAIRE
Please describe your current participation in the following types of exercise:

1. Aerobic (aerobic classes, walking, jogging, stair climbing, hiking, cycling, etc.)
Frequency (# of days per week): __________________
Duration (time spent per session): __________________minutes
Intensity (difficulty level): light somewhat hard hard very hard
How long have you been participating in aerobic activity as described above?
__________________Years

2. Anaerobic (weight training, sprinting, etc.)
Frequency (# of days per week): __________________
Duration (time spent per session): __________________minutes
Intensity (difficulty level): light somewhat hard hard very hard
How long have you been participating in anaerobic activity as described above?
__________________Years

3. Organized or Recreational sports
Type of sport(s):____
Frequency (# of days per week): __________________
Duration (time spent per session): __________________minutes
Intensity (difficulty level): light somewhat hard hard very hard
How long have you been participating in sports activity as described above?
__________________Years

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SUBJECT: ________   DATE: _________________

MUSCULOSKELETAL QUESTIONNAIRE

Spine / Low Back / Sacroiliac Joint:

Have You Ever Suffered An Injury To Your Spine / Low Back / Sacroiliac Joint?

☐ YES ☐ NO

- List Time Missed ______________________________
- Please Describe ______________________________

____

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Been Hospitalized For A Spine / Low Back / Sacroiliac Joint Injury?

☐ YES ☐ NO

- When? ______________________________
- Please Describe ______________________________

____

Have You Ever Had Surgery of Any Kind on Your Spine / Low Back / Sacroiliac Joint?

☐ YES ☐ NO

- When? ______________________________
- Please Describe ______________________________

____

Have You Ever Had Numbness/Tingling Down One (1) or Both Legs?

☐ YES ☐ NO

- Date(s)/Time Missed? ______________________________

____

- Please Describe? ______________________________

____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Spine, Low Back, or SI Joint Injury?

☐ YES ☐ NO

- Please Describe ______________________________

____

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Hip/Groin

Have You Ever Suffered An Injury To Your Hip / Groin *(including hernias and/or sports hernias)?* ☐ YES ☐ NO

- List Time Missed ______
- Please Describe ______

______

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Had Surgery For A Hip / Groin Injury?

☐ YES ☐ NO

- When? ______
- Please Describe ______

______

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Hip and/or Groin Injury?

☐ YES ☐ NO

- Please Describe ______

______

Thigh / Hamstring / Quadriceps:

Have You Ever Suffered An Injury To Your Thigh, Hamstring, and/or Quadriceps?

☐ YES ☐ NO

- List Time Missed ______
- Please Describe ______

______

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Been Hospitalized For A Thigh, Hamstring, and/or Quadriceps Injury?

☐ YES ☐ NO

- When? ______
- Please Describe ______

______

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Have You Ever Had Surgery For A Thigh, Hamstring, and/or Quadriceps Injury?

☐ YES ☐ NO

- When? ____________________
- Please Describe ____________________

____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Thigh, Hamstring, or Quadriceps Injury?

☐ YES ☐ NO

- Please Describe ____________________

____

Knee / Patella:

Have You Ever Suffered An Injury To Your Knee and/or Patella (kneecap)?

☐ YES ☐ NO

- List Time Missed ____________________
- Please Describe ____________________

____

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Been Hospitalized For A Knee and/or Patella Injury?

☐ YES ☐ NO

- When? ____________________
- Please Describe ____________________

____

Have You Ever Had Surgery For A Knee and/or Patella Injury?

☐ YES ☐ NO

- When? ____________________
- Please Describe ____________________

____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To A Knee / Patella Injury?

☐ YES ☐ NO

- Please Describe ____

____

Have You Ever/Do You Presently Wear A Knee Brace?

☐ YES ☐ NO

- Which Knee? ________________ Brand / Model of Brace? ________________

____

- Reason for Wearing ? ____

____

Ankle / Lower Leg:

Have You Ever Suffered An Injury To Your Ankle / Lower Leg?

☐ YES ☐ NO

- List Time Missed ____
- Please Describe ____

____

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Been Hospitalized For An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

____

Have You Ever Had Surgery For An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- When? ____
- Please Describe ____

____

Have You Ever Been Advised Not To Participate In Athletic Activities Due To An Ankle / Lower Leg Injury?

☐ YES ☐ NO

- Please Describe ______________________________

Do You Presently ☐ Tape Your Ankle(s) ☐ Use Ankle Brace(s) ☐ Other

- Please Describe ______________________________

Foot / Toes:

Have You Ever Suffered An Injury To Your Foot / Toe(s)?

☐ YES ☐ NO

- List Time Missed ______________________________
- Please Describe ______________________________

Were Any Diagnostic Tests Performed? (check all that apply)

☐ X-Rays ☐ MRI ☐ CT-Scan ☐ Bone Scan

Have You Ever Had Surgery For A Foot / Toe Injury?

☐ YES ☐ NO

- When? ______________________________
- Please Describe ______________________________

Have You Ever Been Advised Not To Participate In Athletic Activities Due To An Foot and/or Toe Injury?

☐ YES ☐ NO

- Please Describe ______________________________

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Appendix H. IRB Approval Letter

THE UNIVERSITY OF
TENNESSEE
KNOXVILLE

January 30, 2024

Joshua Trueblood Weinhandl,
UTK - Coll of Education, Hlth, & Human - Kinesiology, Recreation & Sports Studies

Re: UTK IRB-24-07978-XP
Study Title: INERTIAL MEASUREMENT UNITS AS PROXY MEASURES FOR KNEE KINETIC VARIABLES ASSOCIATED WITH ACL RE-INJURY

Dear Joshua Trueblood Weinhandl:

The UTK Institutional Review Board (IRB) reviewed your application for the above referenced project. It determined that your application is eligible for expedited review under 45 CFR 46.110(b)(1), categories (4) and (7). The IRB has reviewed these materials and determined that they do comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects.

Therefore, this letter constitutes full approval by the IRB of your application (Version 2.1) as submitted, including:

Documents Stamped:

- Consent for Research Participation – (Version 1.1)
- In-Person Recruitment Announcement - (Version 1.1)
- Verbal In-person Instructions - (Version 1.2)
- Thesis Recruitment Email - (Version 1.1)
- Data Collection Sheet - (Version 1.0)
- Musculoskeletal_Questionnaire - (Version 1.0)
- Fitness_Activity_Questionnaire - (Version 1.0)

That have been dated and stamped IRB approved. You are approved to enroll a maximum of 100 participants. Approval of this study will be valid from 01/30/2024 to 01/29/2025.

Any revisions in the approved application, consent forms, instruments, recruitment materials, etc., must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems involving risks to subjects or others in the manner required by the local IRB policy.

Finally, re-approval of your project is required by the IRB in accord with the conditions specified above. You may

Institutional Review Board | Office of Research & Engagement
1534 White Avenue Knoxville, TN 37996-1529
865-974-7697 865-974-7400 fax irb.utk.edu

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Flagship Campus of the University of Tennessee System

Vita

Kavon Bonakdar was born and raised in Jackson, Tennessee, in 1999 to Hamid Bonakdar and Farzaneh Kaveh. He is the youngest of two children and has an older sister, Shida. Kavon graduated from the University School of Jackson in 2018. He then graduated from the University of Tennessee with a B.S. in Kinesiology and Hispanic Studies. He stayed at the University of Tennessee to complete an M.S. in Kinesiology with an emphasis in Biomechanics in the Summer of 2024. He plans on continuing his education at Washington University to pursue his Doctor of Physical Therapy degree.