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CONNECTING THE PIECES: HOW LOW BACK PAIN ALTERS LOWER
EXTREMITY BIOMECHANICS AND SHOCK ATTENUATION IN ACTIVE
INDIVIDUALS

DISSERTATION

A dissertation submitted in partial fulfillment of the
requirements for the degree of Doctor of Philosophy in the
College of Health Sciences
at the University of Kentucky

By

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Lexington, Kentucky

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Clinical Nutrition

and

Dr. Arthur J. Nitz, Professor of Physical Therapy

Lexington, Kentucky

2019

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ABSTRACT OF DISSERTATION

CONNECTING THE PIECES: HOW LOW BACK PAIN ALTERS LOWER EXTREMITY BIOMECHANICS AND SHOCK ATTENUATION IN ACTIVE INDIVIDUALS

Low back pain in collegiate athletes has been reported at a rate of 37% from a wide array of sports including soccer, volleyball, football, swimming, and baseball. Whereas, in a military population the prevalence of low back pain is 70% higher than the general population. Compensatory movement strategies are often used as an attempt to reduce pain. Though compensatory movement strategies may effectively reduce pain, they are often associated with altered lower extremity loading patterns. Those who suffer from chronic low back pain tend to walk and run slower and with less trunk and pelvis coordination and variability. Individuals with low back pain also tend to run with more stiffness in their knees. Moving with less joint coordination and more stiffness are potential compensatory movement patterns acting as a guarding mechanism for pain.

Overall the purpose of this project was to determine how chronic low back pain influences lower extremity biomechanics and shock attenuation in active individuals compared to healthy individuals and examine how the altered lower extremity biomechanics are related to clinical outcome measures. We hypothesized that individuals who present with chronic low back pain are more likely to exhibit higher vertical ground reaction forces and less knee flexion excursion during landing, compared to healthy individuals. We also hypothesized that individuals with chronic low back pain will have a reduced ability to attenuate shock during landing compared to the healthy individuals.

This study was a case control design in which physically active individuals suffering from chronic low back pain were matched to healthy controls. All participants reported for one testing session to assess self-perceived knee function in the form of the Knee Osteoarthritis Outcomes Score (KOOS), lower extremity strength and mechanics during three landing tasks. Isometric strength was assessed using an isokinetic dynamometer during hip abduction, hip extension, and knee extension. The landing tasks included a drop vertical jump, a single leg hop, and a crossover hop. A three-dimensional motion analysis system with two in-ground force plates and four inertial measurement units were used to assess lower extremity mechanics during the landing tasks.

Individuals with low back pain presented with reduced KOOS scores compared to healthy individuals in four of the five subscales, including Symptoms ($p=0.007$), Pain ($p=0.002$), Activities of Daily Living ($p=0.021$), and Quality of Life ($p=0.003$).

Alternatively, while there were some strength, kinematic, and kinetic between limb asymmetries noted in the low back pain group, there were not between group differences with the healthy individuals. In the low back pain group, individuals presented with greater dominant limb knee extension strength ($p=0.039$) and greater dominant limb ankle plantarflexion at initial contact during the drop vertical jump, compared to the non-dominant limb ($p=0.022$). Individuals with low back pain also presented with greater non-dominant limb tibia impact during the single limb hop ($p=0.008$).

While we did not identify any mechanical differences between individuals suffering from chronic low back pain and those who do not, we did identify that an active population suffering from low back pain does present with decreased self-perceived knee function compared to active individuals without low back pain. As these groups biomechanically perform similarly, they do not clinically perform the same, specifically, in terms of the KOOS. Such differences should not be overlooked when treating active populations with low back pain. If this population is presenting with altered self-perceived knee function at a young age, it is likely that it will continue to decline and negatively affect their function.

KEYWORDS: Low Back Pain, Biomechanics, Active, Lower Extremity, Inertial
Measurement Units

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11/15/2019

Date

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DEDICATION

To Raza, three degrees, four states, lots of road trip miles, several tears and countless laughs, undoubted dedication, and endless love.

My mental health stability, without you, I would not have made it.

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CHAPTER 1. BACKGROUND

Epidemiology of Low Back Pain

Chronic low back pain is a debilitating disease globally and is the leading cause of physician visits and utilization of healthcare services.[1] It is also a leading cause of activity limitation in individuals under 45.[2] It has been previously reported that in the US, low back pain incurs total health care expenditures up to \$91 billion per year.[3] In the general population low back pain affects about 80% of individuals at least once in their lives,[4, 5] and during any given year 15-20% of individuals will present with low back pain.[3]

In athletes, low back pain is also a common occurrence. It has been estimated that unspecified low back pain occurs in up to 15% of adolescent athletes,[6] though prevalence of low back pain is higher in some sports. Low back pain in adolescent gymnasts has been reported to be upwards of 86%.[7] Low back pain has been reported in 37% of adult athletes in a wide array of sports including soccer, volleyball, football, swimming, wrestling, tennis, gymnastics, and baseball.[8, 9] Approximately 30% of college football players have reported low back pain[10] while 30% have lost playing time.[8] Degenerative disc disease and spondylolysis are the most common diagnoses in athletes with low back pain.[8] Greene et al.[11] reported that college athletes with a history of low-back injury had three times the risk for subsequent pain compared to those who had never experienced pain. Additionally, factors such as presence of low back pain, missed training, and time to return to regular training and/or competition were all significant predictors of a low back injury in the following season .[11]

It has been indicated that the prevalence of low back pain in military personnel is 70% higher than the general population.[12] In a sample of Marines queried on their past injury history, 32% reported an injury while 25% of those injuries were localized to the low back.[13] The data were consistent with other high incidences of LBP reported in Special Operations Forces commands.[14] Low back pain is the leading cause of lost duty days[15, 16] and caused the greatest 5-year risk of disability from the military at rates approaching 20%.[17]

Etiology of Low Back Pain

It is common for an individual to experience idiopathic low back pain that can persist for long periods of time. Physically active individuals, such as athletes and military personnel are no exception to this cause. Chronic unspecified low back pain is defined as experienced pain between T12 and gluteal folds with no radiographic or diagnostic explanation. It is commonly reported that chronic unspecified low back pain may be due to weak core muscles,[6] hip muscle tightness,[18-20] or general overuse pain.[8] Individuals who experience low back pain may have a clear diagnosis, based on either a defining event or a specific injury, such as end plate fractures or disc herniation.[21] Overuse injuries, such as spondylolysis, spondylolisthesis, sacroiliac or vertebral joint inflammation, posterior element overuse syndrome, overuse disc degeneration and herniation, and vertebral body apophyseal avulsion fractures are common diagnoses in patient with low back pain. Additional reasons for low back pain, albeit less common, include infections, tumors, and cysts[21] and musculoskeletal deformities such as scoliosis or hip dysplasia. Despite the reason for the pain it is common that low back pain leads to changes in one's movement patterns.[22-24] These compensations are often an attempt to

reduce or avoid pain and become learned behaviors that may result in lasting damage to the lower extremity joints.[25, 26] In a highly active population, such as the military or athletes, compensatory lower extremity joint mechanics may alter joint loading and increase the risk for secondary lower extremity injury.

Low Back Pain Movement Strategies

Regional Interdependence

Regional interdependence is defined as clinical observations related to the relationship purported to exist between regions of the body, specifically with respect to the management of musculoskeletal disorders.[27] The human body may alter the way it absorbs and transmits forces through the ankle, knee, hip, back, and neck, or how pain in a proximal location may affect mechanics at distal and proximal joints. Individuals with low back pain tend to exhibit behaviors associated with regional interdependence, as they present with altered movement patterns potentially due to the back pain they are experiencing.[26, 28] Regional interdependence is similar to the concept of the kinetic chain in that the body is connected in a way that an injury at one joint location can have lasting affects at other locations due to potential compensations.[29] Regional interdependence is demonstrated by individuals with low back pain, as they have presented with altered mechanics, specifically at the knee.[28] Most patient reported outcomes are specific to a type of injury, disease, or an area of location on the body. Individuals with both osteoarthritis at the hip or knee and LBP have been shown to report worse function, higher pain, and greater stiffness on patient reported outcomes than those with hip or knee osteoarthritis without low back pain.[26] The Knee Osteoarthritis Outcomes Score

(KOOS) is used to identify knee related challenges that patient's experience, and have been linked to identifiers leading to possible knee injuries[30] and knee osteoarthritis.[31] Granan et al.[30] found that individuals with lower KOOS quality of life scores were at a 33% greater risk of a secondary ACL tear. Similarly Long et al.[32] reported that individuals who had sustained a lower limb injury reported knee quality of life scores that were correlated with higher loading during gait, and related to knee osteoarthritis severity. Therefore, it is possible that additional patient reported outcomes, not specific to one's injury, but interdependent, may provide a more complete understanding of how the injury affects the patient's overall pain and function.

Neuromuscular Deficits

Nadler et al.[25] found that females who reported low back pain in the last year had a greater strength discrepancy in the hip extensors (15%) than females who did not report low back pain in the last year (5.3%). This hip muscle asymmetry has been shown to be predictive of whether or not female athletes will seek treatment for low back pain within the following year.[18, 19] Similarly, Kendall et al.[33] found that patients with low back pain demonstrated 31% less hip abduction force output than patients without low back pain and Hides et al.[34] found that strength deficit patterns associated with athletes with low back pain may have negative consequences on performance. Additional research suggested that those with lower hip flexor and hip adductor strength were more likely to present with low back pain.[20] Overall, those with low back pain tend to present with weaker lower extremity musculature, which in other populations, such as anterior

cruciate ligament reconstruction, have been known to increase their risk for secondary injuries.[35-37]

Gait Biomechanics

Previous research has demonstrated the interdependence link between LBP and altered lower extremity biomechanics. The human body may alter the way it absorbs and transmits ground reaction forces through the ankle, knee, hip, back, and neck.[38] These alterations may be due to proximal pain leading to poor distal joint mechanics. Individuals with low back pain have presented with altered biomechanics at the knee, highlighting regional interdependence.[39, 40] Individuals with LBP that presented with lower vertical ground reaction forces during walking, also landed with a more extended knee at initial contact.[41] This decrease in vertical ground reaction force may be attributed to the fact that patients with low back pain tend to walk slower.[42, 43] Voloshin and Wosk suggested that individuals with LBP had a diminished ability to absorb shock during walking by up to 20%.[44]

Joint coordination is the coupling of segments to produce efficient movement.[45] Inter-segmental or inter-joint coordination uses the segments and the joints, as well as temporal spatial organization to identify coordinative movement patterns.[46] Joint coordination has been used to explain motor control changes of the lower extremity, in many pathological populations including those with patellofemoral pain,[47] iliotibial band syndrome,[48] hip arthroplasty.[49] Defining the phase movements of coordination is necessary for assessing motor control mechanisms during gait.[50] In healthy individuals, a faster walking speed tends to lead to a transition of antiphase movement,

meaning that their trunk and pelvis move in opposite directions in the transverse plane. Whereas, during slower walking speeds healthy individuals move in more in-phase movement patterns, meaning that the trunk and the pelvis are moving in the same direction in the transverse plane as they step forward.[51, 52] Individuals with low back pain do not transition to the antiphase movement associated with faster walking speeds like healthy individuals, suggesting a more rigid or “guarded” gait pattern.[22, 23, 53, 54] In the transverse plane trunk pelvis coordination of healthy individuals evolves from in phase to antiphase coordination walking velocity increases.[52] This suggests that their trunk and pelvis move in opposite directions in the transverse plane, while individuals who suffer from chronic low back pain walk with less trunk and pelvis joint coordination and variability. The lack of ability to move into an out of phase movement suggests that those with low back pain move in a stiffer pattern as a guarding mechanism for pain.[55, 56]

Sung et al.[57] identified that individuals with low back pain present with a dominant limb dependence strategy during walking. Individuals with low back pain may be more likely to demonstrate poor loading mechanics on the non-dominant limb, potentially relying on the dominant limb for effective performance and pain avoidance. Individuals suffering from low back pain may become dominant limb dependent by increasing their lumbar spine rotation on the dominant side [58, 59], and decreased time in single limb stance on the dominant side, during walking [57]. Spending less time on the dominant limb, especially during more functional tasks such as running or jumping, may inhibit proper load distribution throughout the lower extremity leading to further injury.

A combination of higher vertical ground reaction forces and smaller knee excursions causes force to be transmitted in other directions throughout the body.[60, 61] Landing with greater knee extension at initial contact may be indicative of diminished lower extremity strength and proprioception,[62] and can sometimes lead to the knee converting shock absorption into the frontal plane.[63] Excessive knee valgus motion during landing forces the tibia to translate in an inappropriate manner leading to potential destruction of the ligaments within the knee, most notably the anterior cruciate ligament.[64] Also, landing with higher vertical ground reaction forces and less knee flexion results in a greater knee extension joint moment.[60, 65, 66] This torque exposes the knee to potential injury. In individuals with low back pain there is a lack of information on the interdependence of lower extremity biomechanics in active populations, in which complex dynamic movements are more relevant. Hamill et al.[39] found that individuals with low back pain exhibit greater knee stiffness during running. Haddas et al.[67] identified that females land with greater knee flexion at initial contact, smaller vertical ground reaction forces, and greater knee flexion moments compared to males. The lack of research surrounding lower extremity functional mechanics in active individuals with low back pain demonstrates a critical need to further understand how these individuals compensate for pain.

Modalities for Biomechanical Analysis

Three Dimensional Motion Capture

Three dimensional motion capture has been regarded as the gold standard technology for collecting human movement biomechanical data,[68-70] especially during sport specific movements.[71, 72] Retroreflective markers are placed specific bony landmarks to create a model of the human body, and additional markers are placed on the segments of interest to track the human movement. This allows for the calculation of joint range of motion and position during certain events in a movement, referred to as kinematics. Additionally, kinetic data is collected using force plates, measuring the direct amount of force applied to the ground during contact. Three dimensional motion capture has been deemed as valid and reliable for assessing many different types of human movement, including jumping and landing tasks.[69-72] Unfortunately, three dimensional motion capture can be extremely restrictive, as the cameras are very expensive and the data must be collected in a controlled laboratory setting, thus performance coaches, clinicians, and researchers alike have expressed the need to capture field data, for more translatable research.

Inertial Measurement Units

Inertial Measurement Units (IMUs) are sensors that are able to collect biomechanical data. They are typically small, portable, and collect data via Bluetooth signal. IMUs may include three different types of data captured from magnetometers (position in global space), gyroscopes (rotational accelerations), and accelerometers (linear accelerations). Multiple combinations of this data have begun to be used in

biomechanical data collections for human movement, mainly because of their lesser expense and portable use, providing a greater ability for field data collection.[73-76] Many attempts have been made to validate IMUs, though there are many inconsistencies in how the data are collected, processed, and analyzed.[74, 76-78] IMUs are able to analyze joint kinematics,[74, 77] segment impacts identified by peak accelerations,[76, 79], segment angular velocities,[75, 80] and shock attenuation in the frequency domain[78, 81, 82] and in the time domain, commonly referred to as acceleration reduction.[78, 82] Thus, the validation of IMUs for clinical use maybe task and study specific.

IMU data have been used in conjunction with three dimensional motion capture data, as well as standalone. Elvin et al.[79] determined that increases in knee contact angle during landing influences not only vertical ground reaction forces but also segment accelerations, with the strongest positive relationship to peak pelvis impacts. Further, joint angular velocity has been related to three dimensional mechanics that are associated with poor loading mechanics following anterior cruciate ligament reconstruction.[75, 80] Early research by Shorten and Winslow[81] identified that as running speed increases, tibial impact is increased, then actively attenuated throughout the body to adapt for increases in load. Similarly prior shock attenuation research has also demonstrated that during a fatiguing run the body loses its shock absorption capabilities.[83] In the time domain, Derrick et al.[78] demonstrated that healthy individuals present with poor acceleration reduction following a fatiguing run.

Significance and Specific Aims

Low back pain is a common occurrence among active individuals including athletes and military service members, and has been known to be a primary cause of reduced levels of activity, a loss of sport participation, and lost duty days. Pain avoidance movement strategies are documented in a commonly older low back pain population, but lacks evidence in these typically younger and more active populations. These changes in biomechanical strategies (i.e. inadequate shock attenuation and/or increased sagittal and frontal plane knee joint moments) may be associated with a higher risk of a secondary lower extremity injury. Active individuals consistently sustain impacts greater than those experienced during activities of daily living, due to the nature of sport, activity, or even occupation. If active individuals are able to maintain function despite biomechanical compensations from low back pain, they may be putting themselves at greater risk for reduced long-term musculoskeletal health outcomes. By evaluating lower extremity biomechanics and shock attenuation in active individuals who present with low back pain, we can begin to understand the magnitude this commonly unspecified condition may have on a more active population. Therefore, the primary purpose of this project is to determine how chronic low back pain influences lower extremity biomechanics and shock attenuation in active individuals compared to healthy individuals and examine how the altered lower extremity biomechanics are related to clinical outcome measures.

Specific Aim 1: To determine the relationship between lower extremity kinematics and kinetics and shock attenuation in healthy individuals during a functional landing task.

Hypothesis 1.1: Higher vertical ground reaction forces will be associated with poor shock attenuation during landing.

Hypothesis 1.2: Lower joint excursion would be associated with poor shock attenuation during landing.

Specific Aim 2: To determine the effects of chronic low back pain on strength, lower extremity biomechanics, and shock attenuation during landing compared to a healthy population

Hypothesis 2.1: Individuals with LBP would have weaker hip and quadriceps strength compared to healthy control groups.

Hypothesis 2.2: Individuals with LBP would have altered lower extremity biomechanics including decreased knee excursion, and increased knee joint loading compared to healthy individuals during landing.

Hypothesis 2.3: Individuals with LBP would have an inability to attenuate shock through the kinetic chain to the trunk during landing, compared to healthy individuals.

Specific Aim 3: To determine how lower extremity biomechanics during landing relate to clinical outcome measures in individuals with chronic low back pain

Hypothesis 3.1: Altered lower extremity mechanics and shock attenuation would be related to higher Oswestry Low Back Pain Disability scores and lower Knee Injury and Osteoarthritis Outcome Scores

Hypothesis 3.2: Altered lower extremity mechanics and shock attenuation would be related to reduced lower extremity isometric strength

CHAPTER 2. THE RELATIONSHIP BETWEEN LOWER EXTREMITY MECHANICS AND LOAD CHARACTERIZED BY INERTIAL SENSORS DURING LANDING

Introduction

An individual's ability to absorb forces plays a critical role in injury risk reduction and physical performance optimization. Mechanical shock, generated by ground contact, is often attenuated or reduced by the body and controlled by factors not limited to eccentric muscle control, soft tissue absorption, and increased joint excursion. During locomotion, the lower extremities are often the prime shock absorbers of the body.[61, 84, 85] in which the lower extremities can be manipulated to lessen, or better attenuate load during ground contact. For example an increase in knee flexion at initial contact reduces vertical ground reaction forces.[61, 86] Also, landing with greater knee flexion excursion has been shown to mitigate impact stress compared to landing with a stiffer knee.[61, 87-89] The likelihood of lower extremity injury, like an ankle sprain or an anterior cruciate ligament injury, increases when the load of ground contact becomes larger than what the lower extremities can sufficiently attenuate.[90]

Functional tasks, such as sport specific movements like hopping and jumping, are clinically applicable tools for clinicians and researchers to assess patient movement strategies, progress, and return to sport participation as they incorporate multiple goals within the task. Functional tasks have the ability to highlight injury risk and the capability to measure muscle strength, power, proprioception, and neuromuscular control among other constructs.[91] Single limb hops are consistently used by clinicians to determine lower extremity function, as only one limb is available to absorb the load of the entire body

and must decelerate the center of mass in both horizontal and vertical directions.[92] The crossover hop provides similar functional assessment capabilities as a single limb hop, while adding a level of complexity by incorporating lateral stability requiring the individual to hop over a line during three continuous hops. This requires increased hip strength and neuromuscular control for not only vertical shock absorption but also lateral stability. Assessing functional tasks that require different levels of complexity better simulate game like situations for athletes. It is common for healthy individuals to exhibit high levels of variability when performing landing tasks.[93] It has also been suggested that greater levels of movement variability, such as inconsistent ankle, knee, and hip moments, may contribute to a greater injury risk.[93] Further, Nordin and Dufek[94] found that healthy individuals change their load absorption strategies and movement variability based on task demand. Also, landing with greater vertical ground reaction forces has been associated with a greater upright landing posture, for example reduced hip flexion and a more rigid trunk, and an increase in risk for injury[61, 90] via greater quadriceps demand.[95, 96] Thus, assessing more than one functional task may be important when analyzing landing strategies.

Clinicians often use a battery of tests to return athletes to sport participation that include a combination of jumping and hopping.[97] Clinicians may benefit from having more information than distance or height jumped, such as lower extremity loading parameters and joint kinematics in order to make more informed decisions. While three dimensional (3D) motion capture systems provide us with important mechanical information on the way an individual moves, they are knowingly expensive, require a controlled laboratory space, and require a greater technical understanding, thus most

clinics do not have the personnel, physical, and fiscal resources to accommodate 3D motion capture systems. Alternatively, inertial measurement units (IMUs) are a less expensive technology that may provide the necessary information for clinicians. Derrick et al.[78] demonstrated that healthy individuals present with poor acceleration reduction following a fatiguing run. In addition, Elvin et al.[79] determined that knee contact angle during landing influences not only vertical ground reaction forces but also segment accelerations, with the strongest positive relationship to peak pelvis impacts. IMUs are portable and have wireless capabilities, which may provide advantages to being used in a clinic or field setting. Though IMUs appear to be useful for clinicians, information surrounding the use of IMUs during functional tasks collected in clinic settings are limited.[75, 77, 98] . Therefore, the purpose of this study was to determine the relationship between lower extremity kinematics and kinetics and shock attenuation in healthy individuals during a functional landing task. We hypothesized that higher vertical ground reaction forces and lower joint excursion would be associated with poor shock attenuation during landing.

Methods

Participants

Healthy individuals were recruited from university sports teams, fitness centers, and the general population for a cross sectional study. Individuals were eligible to participate in this research study if they had not experienced a bout of low back pain (LBP) lasting more than 48 hours, had never undergone back or lower extremity surgery, had not

experienced a lower extremity injury in the last year, had not been diagnosed with a musculoskeletal deformity (such as scoliosis or hip dysplasia), were not pregnant, and scored less than a 10% on the Modified Oswestry Low Back Pain Disability Index[99] (Appendix A). Additionally, these individuals must also have scored a minimum of a 5 on the Tegner Physical Activity Scale (Appendix B).[100] Scoring a 5 on the Tegner indicates that they were moderately physically active running at least two times a week on uneven ground, or competitive cycling, or taking part in heavy construction work. Limb dominance was recorded for all individuals as which leg they would choose to kick a soccer ball. All individuals read and signed an informed consent approved by the University's Institutional Review Board. Participation in this study included one visit to the University of Kentucky's Sports Medicine Research Institute to complete a biomechanical analysis of functional hop tasks.

Procedures

Hop Testing

Biomechanical outcome measures were collected during a single limb hop, and a cross over hop. Individuals were allowed as many practice trials as they deemed necessary to feel comfortable, then five successful trials on both the dominant and the non-dominant limb were recorded for biomechanical analysis. The single limb hop and crossover hop (Figure 2.1) followed Noyes et al.[97] specifications. In the single limb hop, the starting line was placed so individuals must stick the landing on one of two in-ground force plates. Individuals were instructed to jump forward as far as possible and stick the landing, identified by no double hops, pivoting, shifting, touching the other foot to the ground, or

touching the ground with their hands. A single limb hop in which any part of the foot did not land on the force plate was considered a bad trial and individuals were asked to give another effort. During the crossover hop the starting line was placed to capture the second landing on one of two force plates for kinetic measurements. The crossover hop consisted of three continuous single limb hops over a 15cm divide. Any trial in which the individual's foot did not fully cross the divide, or the individuals did not land with their foot fully on one force plate during the second hop, they were asked to give another effort while still allowing for adequate rest in between trials. Individuals must also have stuck the final landing, which was identified by no double hops, pivoting, shifting, touching the other foot to the ground, or touching the ground with their hands.

Three Dimensional Motion Capture

Three dimensional motion capture was used to examine lower extremity biomechanics during the hop tasks. Trunk and lower extremity segments were defined and tracked using 14mm markers placed at 7th cervical vertebrae, bilateral acromion processes, the sternum, xiphoid process, the 12th thoracic vertebrae, as well as bilateral iliac crests, anterior superior iliac spines, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, proximal and distal heel, mid-foot distal to the lateral malleoli, and the head of 1st and 5th metatarsals. Additionally, tracking clusters, created by rigid thermoplastic with four 9.5mm markers affixed, were placed over the pelvis at the posterior superior iliac spine and bilaterally on the lateral thighs and shanks (Figure 2.2). After the static calibration, medial and lateral markers placed on anatomical landmarks of the lower extremity were removed and only tracking markers were used for the data collection during the dynamic movements. Kinematics were collected at 200Hz

with a 14-camera 3D motion capture system (Vicon, Centennial, CO) and ground reaction forces captured at 2000Hz, on two in-ground force plates (Bertec Corporation, Columbus, OH).

Inertial Measurement Units

Lower extremity accelerations were collected using three, 9-axis, inertial measurement units (IMUs; I Measure U, Vicon, Centennial, CO), sampled at 500Hz. IMUs were placed on the sacrum (directly underneath the pelvis cluster used for motion capture), and approximately 2cm superior to the medial malleolus on the shank (Figure 2.3).

Data Analysis

3D motion capture data were analyzed using Visual 3D (C-Motion, Germantown, MD). Marker trajectories were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 6Hz. Ground reaction force data were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 50Hz. Lower extremity kinematics were calculated using Euler angles and kinetics were calculated from inverse dynamics methods following International Society of Biomechanics guidelines.[69, 101] Peak vertical ground reaction forces (PVGRF), and peak lower extremity joint angles of the hip, knee, and ankle were identified during the landing phase of all tasks. Joint angle at initial contact for the hip, knee, and ankle were identified at the initiation of ground contact on the force plates when the vertical ground reaction force exceeded 10N. Landing phase sagittal plane joint excursion was calculated for the hip, knee, and ankle. Joint excursion is the difference between the peak joint angle during landing and the joint angle at initial contact. Loading rate was assessed as the mean of the derivative of the vertical ground reaction force curve

from ground contact, indicated by when the force plates recorded a reading greater than 50N[102] to PVGRF.

Acceleration data were collected in the Vicon Nexus software for timing synchronization to motion capture data. IMU data were analyzed in Visual 3D to calculate acceleration reduction and peak impacts in the vertical direction during landing. In the single limb hop, the landing phase was defined as initial contact with the force plate to 100ms after landing. In the crossover hop the second hop was recorded over the force plates and the landing phase was defined as initial contact with the force plates to peak knee flexion. Peak tibia and peak pelvis impact during landing were determined as the peak positive acceleration from initial contact with the force plates to peak knee flexion. Acceleration reduction was the percentage of peak impact between the pelvis and the tibia sensor as $(\text{Peak Pelvis Impact})/(\text{Peak Tibia Impact}) \times 100$. [78, 82] Thus, the reported number is the percentage of the amount of impact that was not absorbed from the tibia to the pelvis, indicating that a larger percentage indicates less favorable shock absorption during landing.

Statistics

Paired t-tests were run to assess differences between limbs were completed to assess dominant to non-dominant limb differences. Pearson Correlation Coefficients were used to assess relationships between IMU and biomechanical variables. Correlation coefficients were interpreted as little to no relationship with a correlation coefficient between 0.00-0.25, a fair relationship between 0.25-0.50, a moderate to good relationship between 0.50-0.75, and a good to excellent relationship above 0.75, in which a correlation

coefficient of 1.00 indicates a perfect relationship.[103] An alpha value of $p=0.05$ was set for all correlations, using SPSS (SPSS 22, IBM Corporation, Armonk, NY)

Results

Twenty-six individuals completed this study (20F, 6M; height: $1.70\pm0.07\text{m}$; mass: $70.65\pm11.12\text{kgs}$; age 24.1 ± 4.1 yrs). All individuals in this study were right leg dominant. Self-reported physical activity was recorded in via the Tegner Physical Activity Scale with an average score of 7.1 ± 1.1 , equivalent to sports such as competitive tennis, and recreation soccer, football, and rugby.

Dominant vs Non-Dominant Limb Differences

In both the crossover hop and the single limb hop there were no differences between the dominant and the non-dominant limb in any IMU or 3D motion capture variables (Table 2.1). Due to the lack of differences between the dominant and the non-dominant limb, variables across limbs were pooled to assess the relationships between the IMU variables and the 3D motion capture variables, additionally, dominant limb only relationships were assessed.

Crossover Hop

Acceleration reduction was not significantly correlated to ALR, PVGRF, or hip, knee, or ankle excursions (Table 2.2). Alternatively, peak tibia and peak pelvis impact demonstrated a statistically significant correlation to ALR (peak tibia: $r=0.422$, $p=0.004$;

peak pelvis: $r=0.335$, $p=0.026$). PVGRF was moderately correlated to peak tibia impact ($r=0.537$, $p=0.0001$) and correlated to peak pelvis impact ($r=0.419$, $p=0.005$; Figure 2.5). Dominant limb only relationships were assessed (Table 2.3), in which acceleration reduction was not significantly correlated to ALR, PVGRF, and hip, knee, or ankle excursions. Though, peak pelvis impact was correlated to PVGRF ($r=0.408$, $p=0.043$; Figure 2.4), and dominant limb peak tibia impact was moderately correlated to both PVGRF ($r=0.591$, $p=0.001$) and ALR ($r=0.522$, $p=0.006$; Figure 2.4).

Single Limb Hop

Acceleration reduction was significantly correlated to PVGRF ($r=0.381$, $p=0.020$), but not to any other 3D motion capture variable (Table 2.2). Peak tibia impact was not significantly correlated to ALR, PVGRF, and hip, knee, or ankle excursion. Peak pelvis impact was fairly correlated to PVGRF ($r=0.468$, $p=0.004$, Figure 2.6). Dominant limb only relationships were assessed (Table 2.3), in which there were no significant correlations between acceleration reduction, peak tibia impact, and peak pelvis impact with ALR, PVGRF, and hip, knee, and ankle excursion.

Discussion

The overall purpose of this study was to determine the relationship between lower extremity kinematics and kinetics with shock attenuation and acceleration reduction in healthy individuals during different functional landing tasks. Our hypotheses were partially supported, in that acceleration reduction was not correlated with lower extremity joint excursions in either task, though those with higher ALR and PVGRF had less

favorable acceleration reduction from the tibia to the pelvis during landing in the single limb hop.

Specifically, the crossover hop provided moderate relationships from IMU variables to 3D motion capture variables. During the crossover hop, individuals who landed during the second hop with higher PVGRF were more likely to exhibit less favorable shock absorption, or a higher percentage of acceleration reduction. Although acceleration reduction provided no significant relationships to PVGRF in the crossover hop, those who exhibited higher peak tibia impacts and peak pelvis impacts also landed with greater PVGRF and faster ALR. It is possible that the relationships were stronger with single sensor outputs, like tibia impact, and not to acceleration reduction because the crossover hop is a multi-dimensional task. With the complexity of the crossover hop, incorporating a lateral component, shock is also being absorbed in the frontal plane during landing, most likely in the knee.[97, 104-106] Therefore, greater tibia impacts may be related to increased PVGRF and ALR because the sensor was placed distal to the knee, experiencing ground contact before the body had the opportunity to absorb the shock. Further, with the lateral component of the crossover hop and our primarily female population, this relationship may have the ability to identify poor landing strategies. Prior research has displayed that females have exhibited greater lateral forces and knee adduction moments during cutting tasks,[107] that are also commonly seen to lead to higher vertical ground reaction forces due to inappropriate landing strategies.[64] While frontal plane accelerations were not measured in this study, by doing so may provide important injury risk evidence. Absorbing force in the frontal plane has been indicated as

a significant risk for injury,[64, 108, 109] especially in relation to anterior cruciate ligament injuries.[64, 110]

Similarly, in the crossover hop there was a relationship between the increase in ALR related to higher peak tibia and pelvis impacts. This landing and loading phase built into one short ground contact time requires multiple constructs of the human performance including, power, proprioception, and neuromuscular control.[91] It has been indicated that the rate at which forces are absorbed by the lower extremity may be more important than peak forces experienced.[111] Thus, the relationship of loading rate to peak tibia and pelvis impacts provides more in-depth information about how individuals may be handling the large amounts of load, on a single limb, over a short period of time, while still trying to optimize performance. Loading rate has been implicated as a factor for the progression or development of injuries,[111] such as osteoarthritis and low back pain.[112] With relationships to loading rate, IMUs may provide more informed decisions about both acute and long term chronic load bearing overuse injuries.

Similarly, during the single limb hop individuals who had greater landing forces tended to absorb less shock during landing. Peak pelvis impact during the single limb hop showed a similar relationship to PVGRF as acceleration reduction. Of the many variables involved during landing from high impact tasks, especially on one leg, there are two interrelated factors that play a large part in absorbing forces associated with ground contact, the time to absorb the force, and the amount of joint flexion excursion experienced during landing.[61, 84, 88] Elvin et al.[79] found peak trunk accelerations to be associated with knee contact angle during jump landings, whereas in our case, knee flexion excursion did not indicate a relationship with IMU variables during landing. It is possible that we did

not see similar relationships as they did because they conducted a bilateral jumping task, and their sensors were placed on the proximal tibia, contrary to this study in which they were placed on the distal tibia.[79] It is also possible that the speed of landing may be controlling the shock absorption factor of the landing more so than the lower extremity joint angles.[113, 114]

Although there are significant relationships reported between the IMU variables and the 3D motion capture variables, the correlation coefficients were still considered only moderate to good at best. One reason there may not be strength in correlations between the IMUs and 3D motion capture variables in this study, especially joint excursions, may be because shock absorption was measured across the two largest shock absorbing joints, the knee and the hip.[61, 85, 88] It is possible that stronger relationships may be present if there were a sensor placed on the thigh, allowing us to look at multiple levels of acceleration reduction throughout the body, in which the hip and the knee do not combined into one shock absorbing mechanism. In a pathological population this may be particularly important based on injury location.

This study is not without limitations, as we had to exclude some data from this analysis due to the limits of the IMU sensors. The commercially available sensors that were used in this study had an upper limit of 16Gs, and in the case of the landing, some individuals exceeded that limit. Further, as IMUs measure accelerations in multiple directions as well as rotational velocity, we chose to assess straightforward accelerometer variables and their relationships for the sake of clinical translatability. As the purpose of this study was to assess the relationship between IMUs and loading mechanics such as sagittal plan joint kinematics, PVGRF, and ALR, assessing the vertical direction was the

most appropriate. Additionally, these measures avoid complicated analyses and calculations that may not be feasible in a clinical setting as they require additional time and expertise.

Ultimately, we believe that impact during landing may be assessed using inertial measurement units (IMUs). IMUs can provide clinicians with a more objective assessment of their patients, especially when it comes to returning athletes to sports after injuries. Although IMUs may be considered expensive, they are more affordable than force plates or fully integrated 3D motion capture systems. IMUs are smaller and can be more user friendly with data collection possible through applications on a tablet giving immediate feedback to clinicians and patients alike. This work not only provides the relationships between vertical ground reaction forces and impacts during landing, but also can also be beneficial for clinicians as a resource for normative landing impacts measured with IMUs. Clinicians can also use this mechanical loading information to guide treatment strategies and evaluate treatment effectiveness.

Table 2.1 Mean \pm Standard Deviation for all inertial measurement unit data and 3D motion capture data, including p values for dominant vs. non-dominant comparisons in both the crossover hop, and single limb hop.

	Dominant Limb	Non-Dominant Limb	P Value
Crossover Hop			
Acceleration Reduction (%)	38.8 \pm 16.1	39.4 \pm 13.6	0.543
Peak Tibia Impact (G)	13.1 \pm 1.8	12.5 \pm 2.1	0.095
Peak Pelvis Impact (G)	4.9 \pm 1.9	4.7 \pm 1.6	0.244
PVGRF (N/kg)	28.3 \pm 4.0	28.3 \pm 3.5	0.987
ALR (N/kg/s)	661.3 \pm 345.4	697.8 \pm 365.2	0.448
Hip Excursion (°)	5.7 \pm 4.7	5.1 \pm 3.4	0.629
Knee Excursion (°)	38.3 \pm 6.3	36.8 \pm 3.3	0.328
Ankle Excursion (°)	26.7 \pm 9.7	24.1 \pm 10.3	0.099
Single Limb Hop			
Acceleration Reduction (%)	51.8 \pm 21.9	49.1 \pm 18.2	0.636
Peak Tibia Impact (G)	14.2 \pm 1.6	13.7 \pm 1.5	0.733
Peak Pelvis Impact (G)	7.2 \pm 3.2	6.7 \pm 2.8	0.526
PVGRF (N/kg)	33.6 \pm 5.2	33.1 \pm 3.7	0.730
ALR (N/kg/s)	939.1 \pm 268.6	971.2 \pm 403.7	0.733
Hip Excursion (°)	10.8 \pm 4.6	10.1 \pm 4.4	0.554
Knee Excursion (°)	39.3 \pm 5.4	37.4 \pm 4.9	0.285
Ankle Excursion (°)	11.7 \pm 9.9	8.3 \pm 5.2	0.311

Table 2.2 Pooled dominant and non-dominant limb correlations of inertial measurement unit variables to 3D motion capture variables. *Indicates significant correlation coefficient of an alpha value of $p < 0.05$.

	Peak Tibia Impact	Peak Pelvis Impact	Acceleration Reduction
Crossover Hop			
PVGRF (N/kg)	0.537 (0.0001)*	0.419 (0.005)*	0.086 (0.577)
ALR (N/kg/s)	0.422 (0.004)*	0.335 (0.026)*	0.018 (0.905)
Hip Excursion (°)	0.249 (0.107)	-0.94 (0.549)	-0.176 (0.259)
Knee Excursion (°)	0.142 (0.363)	0.85 (0.587)	0.063 (0.688)
Ankle Excursion (°)	-0.023 (0.885)	-0.144 (0.358)	-0.007 (0.966)
Single Limb Hop			
PVGRF (N/kg)	0.197 (0.243)	0.468 (0.004)*	0.381 (0.020)*
ALR (N/kg/s)	0.272 (0.103)	0.135 (0.424)	0.034 (0.841)
Hip Excursion (°)	0.232 (0.166)	-0.301 (0.070)	-0.319 (0.054)
Knee Excursion (°)	-0.078 (0.646)	-0.315 (0.058)	-0.280 (0.093)
Ankle Excursion (°)	0.261 (0.131)	-0.116 (0.507)	-0.106 (0.543)

Table 2.3 Dominant limb correlations of inertial measurement unit variables to 3D motion capture variables, represented as: r value (p value). *Indicates significant correlation coefficient of an alpha value of $p < 0.05$.

	Peak Tibia Impact	Peak Pelvis Impact	Acceleration Reduction
Crossover Hop			
PVGRF (N/kg)	0.591 (0.001)*	0.408 (0.043)*	0.075 (0.720)
ALR (N/kg/s)	0.522 (0.006)*	0.346 (0.090)	0.024 (0.910)
Hip Excursion (°)	0.372 (0.067)	0.041 (0.847)	-0.061 (0.773)
Knee Excursion (°)	0.339 (0.097)	0.263 (0.204)	0.118 (0.573)
Ankle Excursion (°)	-0.112 (0.593)	-0.198 (0.344)	-0.008 (0.968)
Single Limb Hop			
PVGRF (N/kg)	0.341 (0.153)	0.305 (0.204)	0.228 (0.347)
ALR (N/kg/s)	0.364 (0.15)	-0.030 (0.904)	-0.116 (0.637)
Hip Excursion (°)	0.223 (0.359)	-0.154 (0.530)	-0.222 (0.360)
Knee Excursion (°)	-0.154 (0.530)	0.110 (0.655)	0.159 (0.516)
Ankle Excursion (°)	-0.249 (0.319)	-0.276 (0.268)	-0.244 (0.330)



Figure 2.1 Visual representation of the single limb forward hop for distance and the single limb crossover hop for distance from Noyes et al.[97]



Figure 2.2 Posterior view of marker placement.



Figure 2.3 Inertial measurement unit placement on the medial distal tibia, and attachment strap representation.

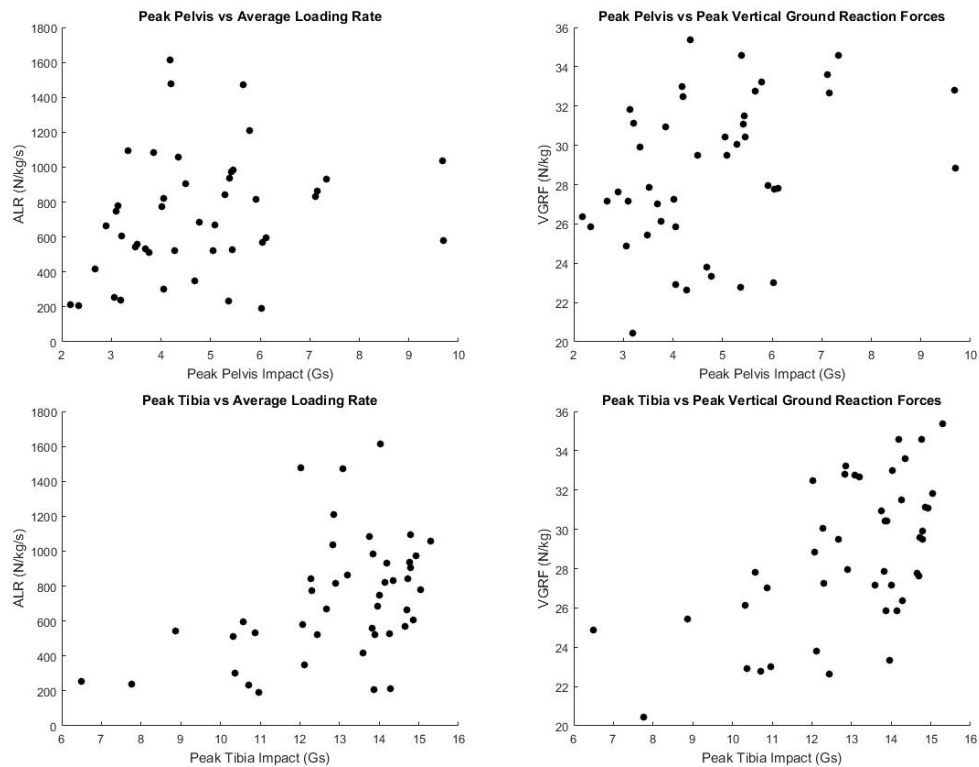


Figure 2.4 Crossover hop pooled data scatter plots of significant relationships between IMU variables and 3D motion capture variables.

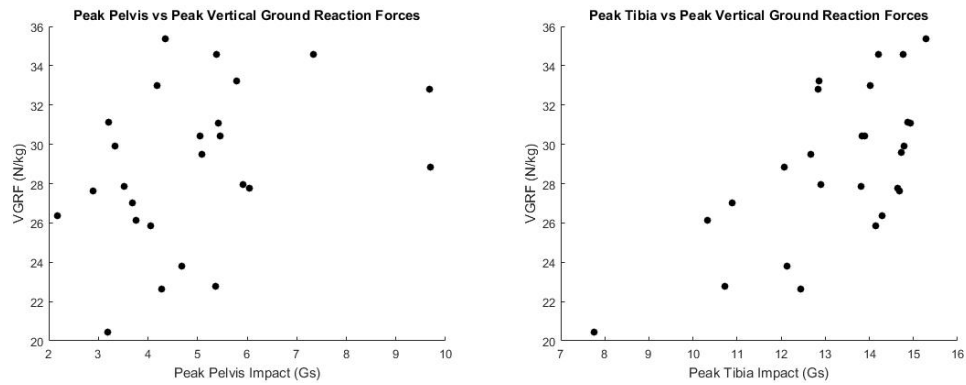


Figure 2.5 Crossover hop dominant limb data scatter plots of significant relationships between IMU variables and 3D motion capture variables.

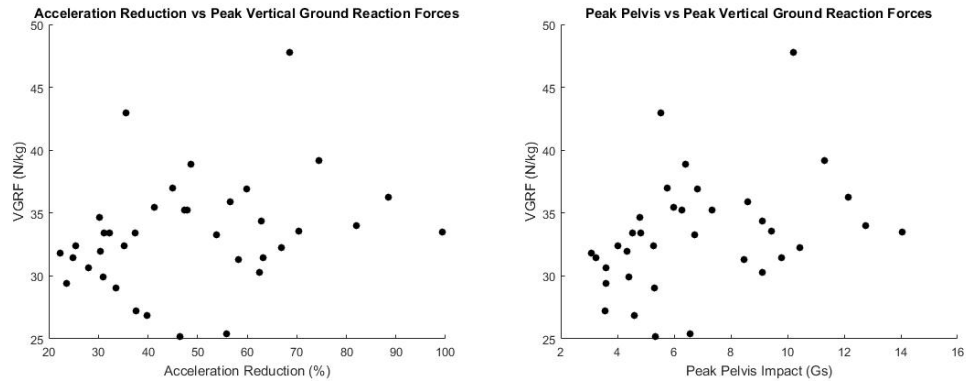


Figure 2.6 Single limb hop pooled data scatter plots of significant relationships between IMU variables and 3D motion capture variables.

CHAPTER 3. LOWER EXTREMITY STRENGTH AND BIOMECHANICAL DIFFERENCES BETWEEN ACTIVE INDIVIDUALS WITH AND WITHOUT CHRONIC LOW BACK PAIN

Introduction

More than 80% of individuals in the general population experience an episode of low back pain (LBP) at some point during their lifetime.[4, 5] In active populations, up to 37% suffer from LBP,[9] and military populations report 70% higher prevalence than the general population.[12] Individuals with LBP develop musculoskeletal deficits such as weaker trunk strength,[115] and poor behavioral mechanisms.[116] They also tend to adapt their movement patterns to compensate for and/or avoid pain[117] with factors such as altered trunk and pelvis joint coordination.[23, 118] Compensation strategies often becomes a learned behavior that can result in lasting damage to the joints of the lower extremities caused by altered loading mechanics.[28, 119, 120] In highly active populations chronic altered lower extremity joint mechanics, for instance increased lower extremity loading and reduced joint excursion, may increase the risk for secondary lower extremity injury.[9]

Movement analyses in individuals experiencing LBP may help understand how and where compensations of altered mechanics are affecting the body, increasing the risk of injury. When conducting movement analyses in individuals with LBP, it has been recommended that the spine and lower extremities be examined together.[121] Although the majority of research on LBP has focused on the biomechanics of the back and trunk, independent of the lower extremities, little research[39, 43] has considered the lower extremities in those with LBP. Individuals with LBP who present with lower vertical ground reaction forces during walking, also landed with a more extended knee at initial

contact.[41] Lower vertical ground reaction force may be attributed to demonstrated slower walking speeds in patients with LBP.[42, 43] Landing with increased knee extension at initial contact may be indicative of reduced lower extremity strength and proprioception.[62] Similar research found that patients with current LBP ran with greater knee stiffness profiles compared to those without LBP.[39] Individuals suffering from LBP may become dominant limb dependent by increasing their lumbar spine rotation on the dominant side [58, 59], and decreased time in single limb stance on the dominant side, during walking [57]. Spending less time on the dominant limb, especially during more functional tasks such as running or jumping, may inhibit proper load distribution throughout the lower extremity leading to further injury.[122]

Regional interdependence explains how pain or an injury in one location can lead to pain or injury beyond the original location of pain.[27, 123] Regional interdependence has become a more widely accepted explanation for secondary injuries and/or pain beyond the location of original pain.[27, 123] The human body may alter the way it absorbs and transmits ground reaction forces through the ankle, knee, hip, back, and neck, or how pain in a proximal location may affect distal joint mechanics.[124] Regional interdependence is demonstrated by individuals with LBP, as they have presented with altered mechanics, specifically at the knee, while also having indicated subsequent hip and knee osteoarthritis.[28] Individuals with both osteoarthritis at the hip or knee and LBP have been shown to report worse function, higher pain, and greater knee stiffness during running than those with hip or knee osteoarthritis without LBP.[26] As research has begun to link the interdependence between LBP and altered lower extremity mechanics,[39] there is a

lack of information on complex dynamic movements that are more relevant to an active population where LBP is so common.

While research studying the interdependence between the lower extremities and the trunk together is warranted,[121] shock attenuation through the kinetic chain to the trunk is increasingly important to identify key absorption strategies. Inertial measurement units (IMUs), small portable sensors that have high clinical applicability, have begun to be used to assess shock absorption strategies.[80, 125, 126] Previous research suggests that LBP is associated with an inability to attenuate shock by 20% from the femur to the forehead during walking.[44] Further, persons with lower limb amputation also commonly present with LBP[127, 128] and have shown to exhibit a reduced ability to attenuate shock which may lead to increased joint moments.[129, 130] Due to their ease of use, IMUs may be able to provide effective information for clinicians to make more informed rehabilitation decisions, especially in active populations. Therefore, additional research is necessary to determine a connection between shock attenuation and lower extremity mechanics. Mechanical compensation strategies may be associated with an increased risk of a secondary lower extremity injury.

Active individuals, including first responders, athletes and military personnel consistently sustain impacts greater than those typically seen during activities of daily living, due to the nature of sport, activity, or even occupation. Prolonged performance with mechanical compensations from LBP may increase the risk for reduced long-term musculoskeletal health outcomes. Evaluation of lower extremity mechanics and shock attenuation in active individuals who present with LBP may identify the magnitude this commonly unspecified condition has on a more active population. Therefore, the purpose

of this project was to determine the how individuals who suffer from chronic LBP present with altered lower extremity strength, biomechanics and shock attenuation during landing compared to a healthy population. We hypothesize that individuals with LBP will have weaker hip and quadriceps strength compared to a healthy control group. We also hypothesize that individuals with LBP will have altered lower extremity biomechanics including decreased knee excursion, increased knee joint loading, as well as an inability to attenuate shock through the kinetic chain to the trunk during landing, compared to healthy individuals.

Methods

Participants

This study is a cross sectional case control study examining the differences between individuals who suffer from low back pain (LBP) and healthy controls (CTRL). Individuals with LBP were recruited from local sports medicine clinics, athletic trainers, fitness centers, and the general population. Patients were eligible to participate in this study if they currently suffering from chronic LBP, which has persisted for at least four months, had not sustained a lower extremity injury in the past year, had never undergone lower extremity surgery, were not experiencing radicular symptoms, had not been diagnosed with a musculoskeletal deformity, such as scoliosis or hip dysplasia, scored greater than a 10% disability on the Oswestry Disability Index (ODI, Appendix 1), a patient reported outcome geared toward understanding an individual's self-perceived low back function in different areas of life.[99, 131] Individuals must also have scored a minimum of a 5 on the

Tegner Physical Activity Scale (Appendix 2). CTRL participants were recruited from an active population and included if they did not have a history of LBP injury, or surgery, a history of lower extremity surgery or lower extremity injury within the last year, or scored lower than a 10% on the ODI. Healthy control participants were matched to the LBP patients based on mass (± 5 kgs), age (± 2 years), Tegner physical activity score (± 2 points but no less than 5), dominant leg (used to kick a ball), and sex (assigned at birth). All individuals read and signed an informed consent approved by the University's Institutional Review Board. Participation in this study included one visit to the University of Kentucky's Sports Medicine Research Institute to complete clinical outcome measures that included patient reported outcomes, lower extremity strength measures, and a biomechanical analysis of movement strategies during different functional tasks.

Clinical Outcomes Measures

Strength

Lower extremity hip and knee strength was measured using a Biodex System 4 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY). Isometric hip abduction was measured side lying, with the hip abducted at 0 degrees (Figure 3.1).[65] Isometric hip extension was measured in the supine position with the hip flexed at 60 degrees (Figure 3.1). Isometric knee extension was measured in a seated position with the knee flexed at 60 degrees (Figure 3.1). For each test, individuals received two practice trials warming up at 50% effort, then one practice trial at 100% effort, before completing three maximal effort trials with 30 seconds rest between each trial. Verbal encouragement was provided

throughout the test. Peak torque across the trials for each joint direction and limb was recorded and normalized to body mass.

Biomechanical Outcome Measures

Functional Tasks

Biomechanical outcome measures were collected during three functional tasks, a single limb forward hop for distance, single limb crossover hop for distance, and a drop vertical jump. All single limb tasks were completed on both legs, whereas the drop vertical jump is a bilateral task. Individuals were allowed as many practice trials as necessary to feel comfortable with each movement. Five successful trials of each functional task were recorded for the biomechanical analysis. The order of testing of each functional task was randomized to prevent fatigue or leaning bias. The single limb hop and crossover hop followed Noyes et al.[97] specifications (Figure 3.2). For the single limb hop the starting line was placed so individuals must stick the landing on a force plate. During the crossover hop the starting line was placed to capture the second landing on a force plate for kinetic measurements. A single limb hop in which any part of the foot did not land on the force plate was considered a bad trial and individuals were asked to give another effort. For the crossover hop, anytime in which the individuals foot did not fully cross the 15cm divide, or during the second hop in which the individuals did not land with their foot fully on the force plates, the individuals were asked to give another effort. Adequate rest time was always provided between trials. For both single limb hops, individuals must also have stuck the final landing, which was identified by no double hops, pivoting, shifting, touching the other foot to the ground, or touching the ground with their hands. Individuals

were instructed to complete the single limb hops by jumping as far as possible while still being able to stick the landing.

The drop vertical jump was completed according to Paterno et al.[132] specifications (Figure 3.3). Individuals stood on a 45cm box and leaned forward until they naturally dropped off the box. At ground contact they performed a maximal effort vertical jump. A vertec vertical jump measurement tool was provided for visual feedback and maximal effort motivation for the participant. Individuals were placed on the box so that they landed with one foot on each in ground force plate, though this information was not divulged to the participant to preserve natural mechanics. If at any time, any part of the foot did not land on the force plate, or two feet landed on one plate, the trial was considered a bad trial and individual was asked to perform a replacement trial. Pain was evaluated after every task using a visual analog scale, rating pain from 0-10.[133]

Three Dimensional Motion Capture

Three dimensional motion capture was used to examine trunk and lower extremity biomechanics during the three different functional tasks. Trunk and lower extremity segments were defined and tracked using 14mm markers (Figure 3.4) placed at 7th cervical vertebrae, bilateral acromion processes, the sternum, xiphoid process, the 12th thoracic vertebrae, as well as bilateral iliac crests, anterior superior iliac spines, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, proximal and distal heel, mid-foot distal to the lateral malleoli, and the head of 1st and 5th metatarsals. Additionally, tracking clusters, created by rigid thermoplastic with four 9.5mm markers affixed, were placed over the pelvis at the posterior superior iliac spine and bilaterally on the lateral thighs and shanks (Figure 3.4). After the static calibration, medial and lateral

markers placed on anatomical landmarks of the lower extremity were removed and only tracking markers were used for the data collection during the dynamic movements. Kinematics were collected at 200Hz with a 14 camera three dimensional motion capture system (Vicon, Centennial, CO) and ground reaction forces captured at 2000Hz, on two in ground force plates (Bertec Corporation, Columbus, OH).

Inertial Measurement Units

Lower extremity vertical accelerations were collected using three, 9- axis, inertial measurement units (IMUs; I Measure U, Vicon, Centennial, CO). Acceleration data were collected in the Vicon Nexus software for timing synchronization to motion capture data, sampled at 500Hz. IMUs were placed on the sacrum (directly underneath the pelvis cluster used for motion capture), and approximately 2cm superior to the medial malleolus on the shank (Figure 3.5).

Data Analysis

Three-dimensional motion capture data was analyzed using Visual 3D (C-Motion, Germantown, MD). Marker trajectories were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 6Hz. Ground reaction force data were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 50Hz. Lower extremity kinematics were calculated using Euler angles, while kinetics were calculated from inverse dynamics methods following International Society of Biomechanics guidelines. Peak vertical ground reaction forces (PVGRF), and peak lower extremity joint angles of the hip, knee, and ankle were identified during the landing phase of all tasks. Joint angle at initial contact for the hip, knee, and ankle were identified at the initiation of

ground contact on the force plates when the vertical ground reaction force exceeded 10N. Landing phase sagittal plane joint excursions were calculated at the hip, the knee, and the ankle. Joint excursion was calculated as the difference between peak joint angle at landing and angle at initial contact. Loading rate was assessed as the mean of the derivative of the vertical ground reaction force curve from ground contact (greater than 50N to PVGRF).[102]

Raw IMU data were analyzed in Visual 3D to calculate acceleration reduction and peak impacts in the vertical direction during landing. In the drop vertical jump, the landing phase of the jump was defined as initial contact with the force plate (greater than 20N) to peak knee flexion. In the single limb hop, landing was defined as initial contact with the force plate to 100ms after landing. In the crossover hop the second hop was recorded over the force plates and the landing phase was defined as initial contact with the force plates to peak knee flexion. Peak tibia and peak pelvis impacts during landing were determined as the peak positive acceleration from initial contact with the force plates to peak knee flexion. Acceleration reduction was the percentage of peak impact between the pelvis and the tibia sensor as $\text{peak pelvis impact} / \text{peak tibia impact} * 100$ [78, 82]. Thus, the reported number is the percentage of the amount of impact that was not absorbed from the tibia to the pelvis, indicating that a larger percentage indicates less favorable shock absorption during landing.

Statistical Analysis

Sample size estimations were based on prior data examining biomechanical characteristics, like those proposed in Aim 2, in an unspecified LBP population who

completed a stop jump task. Based on sample size estimations (G*Power 3.0.10, Germany) using peak knee flexion data during landing of the stop jump, with a Cohen's d effect size of 0.61, a sample size of at least 43 LBP participants was needed to attain 80% statistical power for independent samples t-test with an alpha level of 0.05. An interim power analysis was calculated from the data on our first 12 subjects. Based on this analysis, a sample size of at least 25 LBP participants would be needed 80% power and an alpha level of 0.05.

A 2x2 (group, limb) repeated measures Analysis of Variance (ANOVA) was used to determine the differences in peak knee extension strength, peak hip strength, hip, knee, and ankle excursions and peak joint moments, PVGRF, and shock attenuation in the LBP and the CTRL group, as well as the dominant limb to the non-dominant limb. Conducting a 2x2 repeated measures ANOVA also allowed us to understand if there is a group x limb interaction. Post-hoc paired t-tests were run for variables with significant limb differences or limb x group interactions, while independent t-tests were run for variables with significant group differences or limb x group interactions. Statistical significance was set at $p \leq 0.05$. All statistical analyses were performed in SPSS (version 24; IBM Corp, Armonk, NY).

Results

Twenty-eight individuals who suffered from low back pain (LBP) and twenty-eight healthy control (CTRL) individuals completed this study. There were no significant differences between groups in demographic variable such as height (LBP: $1.7 \pm 0.07\text{m}$, CTRL: $1.7 \pm 0.08\text{m}$; $p=0.896$), mass (LBP: $72.06 \pm 12.1\text{kg}$, CTRL: $72.66 \pm 13.2\text{kg}$; $p=0.861$), age (CTRL: 24.6 ± 4.5 , LBP: $25.0 \pm 4.8\text{yrs}$; $p=0.785$) and activity

level (CTRL: 7.0 ± 1.1 , LBP: 6.7 ± 1.1 ; $p=0.359$). While subjects between groups were matched in terms of sex, there were more females (21) than males (7) that presented with LBP. Individuals had suffered from LBP for an average of 4.7 ± 3.0 years. As expected ODI scores was significant between groups, as was a requirement for inclusion in the study. Every individual in the CTRL group reported an ODI score of 0%, while in the LBP group scores ranges from 10-34%, with the average ODI score being 16.4%.

Strength

Limb asymmetries in peak knee extension strength were identified ($p=0.0001$; Table 3.1) in the repeated measures ANOVA, and post hoc analysis paired t-tests indicated that both the CTRL group ($p=0.005$) and the LBP group ($p=0.039$) demonstrated greater peak knee extension strength in the dominant limb compared to the non-dominant limb (Figure 3.6). No significant differences existed between groups, limbs, or a group x limb interaction in peak hip abduction strength or peak hip extension strength (Table 3.1).

Drop Vertical Jump

During the drop vertical jump (Table 3.2), RM ANOVA revealed within subject limb differences for average loading rate ($p=0.026$) and ankle flexion angle at initial contact ($p=0.005$). The CTRL group had a greater asymmetry in their average loading rate during landing ($p=0.007$), indicating that their dominant limb experienced higher average loading rates than their non-dominant limb, while no significant differences between limbs in average loading rate were found in the LBP group (Figure 3.7). Post hoc paired t-tests

also indicated that individuals in the LBP group landed with greater ankle plantar flexion on their dominant leg compared to their non-dominant leg ($p=0.022$; Figure 3.7).

Crossover and Single Limb Hops

In the single limb hop (Table 3.4), a limb x group interaction for hip flexion angle at initial contact ($p=0.022$; Figure 3.8) was identified. Interestingly, though there was a limb x group interaction, post hoc t-tests did not identify either a group difference or a limb difference in hip flexion at initial contact. There was also a limb x group interaction for peak tibia impact during landing ($p=0.012$, Figure 3.8). Individuals with LBP exhibited higher peak tibia impact on the non-dominant limb compared to their dominant limb ($p=0.008$; Figure 3.8). A similar difference was not present in the CTRL group. Additionally, the RM ANOVA model indicated there was a limb difference in average loading rate ($p=0.048$) during the single limb hop. Post hoc analysis identified no between limb differences in either group in average loading rate during landing. The RM ANOVA model also indicated there was a limb difference in ankle excursion ($p=0.010$; Figure 3.8) during the single limb hop. Specifically, the CTRL group experienced greater ankle excursion in the dominant limb compared to the non-dominant limb ($p=0.032$). Asymmetrical ankle excursion was not present in the LBP group. In addition, in the crossover hop (Table 3.3) no significant group or limb differences were identified.

Discussion

The overall purpose of this study was to determine how individuals who suffer from chronic LBP present with altered lower extremity strength, biomechanics and shock attenuation during landing compared to a healthy population. The goal was to identify how individuals suffering from chronic LBP may compensate for pain in a way that has the potential to put them at greater risk for a secondary injury or reduced long-term musculoskeletal health. Our hypotheses were partially supported in that both the CTRL group and the LBP group exhibited asymmetrical quadriceps strength between the dominant and non-dominant limbs, though there were no quadriceps strength differences between groups. Interestingly, both groups' levels of asymmetries were within the healthy range, greater than 90% symmetry.[134, 135] The CTRL group demonstrated 92% strength symmetry, and the LBP group demonstrated 95% strength symmetry between the dominant and non-dominant limbs. While there are significant differences between limbs in both groups, neither the LBP nor the CTRL groups' case would these be considered clinically meaningful asymmetries. Hip abduction nor hip extension strength were significantly different between the CTRL group and the LBP group. The lack of differences in hip strength is not in agreement with previous research that found athletes who suffer from LBP demonstrated 31% less hip abduction force compared to those who did not suffer from LBP.[33] Nadler et al.[25] also reported that athletes who presented with LBP the prior year also demonstrated greater hip extension strength asymmetry. It is possible that we did not report the same differences in hip strength as previous research due to the nature of our population. Our LBP group was a younger, continuously active population, potentially not allowing for strength declines due to LBP that are commonly seen.

In the drop vertical jump our hypotheses were partially supported, as individuals with LBP presented with increased ankle plantar flexion angles at initial ground contact on their dominant limb compared to their non-dominant limb. This asymmetrical limb interaction was not present in the CTRL group. There were no other lower extremity kinematic differences between limbs or groups. This is in contrast with Hamill et al.[39] in which they identified that during running, individuals suffering from LBP exhibited greater levels of knee stiffness compared to those who did not suffer from LBP, suggesting that those with LBP may not adequately attenuate foot-ground impact. While not significantly different from the CTRL group, our LBP group presented with higher PVGRF's and loading rates, but did present with significantly asymmetrical ankle kinematics. It is possible that individuals with LBP use compensatory kinematic patterns, like increased ankle plantarflexion, to reduce ground impact and the resultant pain.

Further, the drop vertical jump is a bilateral task with multiple goals; landing successfully immediately followed by a maximal vertical jump. It was unexpected to identify asymmetrical loading rate patterns in healthy individuals, though there were not asymmetrical loading patterns in individuals with LBP. It is possible that the individuals in the LBP group landed with increased dominant limb ankle plantar flexion at initial contact, which may be a compensation strategy this is allowing them to land with symmetrical ALR. It is possible the LBP group does not exhibit loading rate asymmetries due to their compensations strategies. Individuals with LBP may try to reduce the amount of load placed on, or experienced by their low back, as a compensation strategy to avoid pain.[41] Though, this potential compensation strategy may lead to symmetrical load distribution it may put them at a higher risk of an ankle sprain, as the mechanism of injury

for a lateral ankle sprain includes increased plantar flexion combined with inversion and internal rotation.[136]

During the single limb hop, individuals suffering from LBP presented with higher non-dominant limb peak tibia impacts during landing compared to their dominant limb, an asymmetry not exhibited in the CTRL group. Sung et al.,[57] identified that during walking individuals with LBP present with a dominant limb dependence strategy. Individuals with LBP may be relying on the dominant limb for effective performance. Because they rely on the dominant limb for effective performance, landing on the non-dominant limb may be unfamiliar and less able to absorb load. The increase in tibia impact, but not similar increases in lower extremity joint kinematics, suggests greater load on the tissues, bones, and joints in the non-dominant limb. If individuals are unfamiliar with landing on their non-dominant limb this may lead to reduced neuromuscular control,[137-139] and the body's ability to properly absorb shock from the ground, potentially relating back to their proprioception and postural control.[140]

IMUs could be used to assess impact in individuals with LBP. This increase in impact has also been associated with chronic overuse injuries,[125] such as stress fractures,[141, 142] as well as altered loading patterns in individuals who have undergone anterior cruciate ligament reconstruction.[80, 126] Long-term altered loading mechanics can lead to osteoarthritis,[143-145] commonly seen in those with LBP.[26, 28] Further, the ability for an IMU to assess these loading alterations during landing is beneficial for clinicians, as IMUs are portable and may provide easier translatability and accessibility in a clinical setting.

Future work assessing mechanics in individuals with LBP may benefit from accounting for the side in which an individual experiences back pain, as well as how long they have been experiencing pain. In this study, data were compared between groups and limbs as dominant and non-dominant limb comparisons based on previous LBP literature following similar methods.[23, 25, 39] Though LBP is not typically a traumatic injury and identifying a specific side of pain may be difficult for some individuals, as the pain is not always localized. Despite that some individuals describe midline LBP, it is possible that accounting for pain location, such as an involved side and uninvolved side, similar to lower extremity pathology research, may provide additional beneficial information in the way an individual with LBP mechanically compensates for pain.

There are limitations to this study, one being the unbalanced sample size in sex, that our sample is primarily female. Females are more likely to present with LBP, [146-148] and suffer from compensations from LBP, than males.[25, 67] Sex differences may have been apparent with a more even dispersion of males and females, providing additional insight to possible compensation mechanisms in individuals with LBP. Another limitation of this study may be the unspecified presentation of LBP. Inclusion of this study required individuals to present with chronic LBP for a minimum of four months. In an active population a diagnosis of a specific injury may be more beneficial to identify mechanical alterations during functional activities. Because compensation strategies are learned, it is possible that additional mechanical alterations may develop in individuals who have been suffering from pain for a longer period. In this study individuals were included if they had been suffering from LBP for four months or greater, though suffering from LBP may consist of lifelong pain. As there have been many definitions of the timeline that defines

chronic, based on the CDC's National Health Interview Survey, and the majority of people experiencing pain over three months, we defined chronic as four months or greater.[149] It is possible that those suffering for many years may have adapted different types of compensations as those who have been more recently suffering in the last year.

While there are no differences in knee extension strength or hip strength between healthy individuals and those suffering from LBP, there were some movement discrepancies during functional tasks. The type of task identified different landing strategies, in terms of bilateral and unilateral landing tasks, thus it is important for clinicians to understand different tasks may produce different strategies in individuals with LBP, and this should be taken into account when assessing different functional tasks. In a more landing explicit task, such as the single limb hop, individuals with LBP present with asymmetrical ground impacts during landing. Such movement strategies may be considered compensation strategies to avoid pain. Rehabilitation for athletes with LBP tends to focus on trunk and core musculature,[8] both of which are beneficial for landing control, though, it may be advantageous of clinicians to ensure individuals present with lower extremity landing mechanics that support absorption of forces from ground contact.

Table 3.1 Mean \pm standard deviations for peak strength variables in the dominant and non-dominant limb in both the CTRL and the LBP groups. ‡ Denotes significant limb differences via the RM ANOVA.

Strength	CTRL		LBP	
	Dominant	Non- Dominant	Dominant	Non-Dominant
Knee Strength‡	$2.30 \pm 0.4^*$	2.12 ± 0.3	$2.25 \pm 0.5^*$	2.14 ± 0.5
Hip Abduction	1.62 ± 0.4	1.61 ± 0.3	1.72 ± 0.3	1.71 ± 0.5
Hip Extension	2.05 ± 0.6	1.88 ± 0.5	1.87 ± 0.5	1.84 ± 0.6

Table 3.2 Mean \pm standard deviations for biomechanical variables measured during the drop vertical in the dominant and non-dominant limb in both the CTRL and the LBP groups. † Denotes significant group x limb interactions via the RM ANOVA. ¥ Denotes significant limb differences via the RM ANOVA. *Denotes significant differences via post hoc analysis. Significance determined by $p \leq 0.05$.

	CTRL		LBP	
	Dominant	Non- Dominant	Dominant	Non-Dominant
Peak Tibia Impact	13.96 \pm 1.3	13.85 \pm 1.7	13.17 \pm 1.6	12.76 \pm 1.7
Peak Pelvis Impact	6.55 \pm 2.8		5.89 \pm 2.8	
Acceleration Reduction	50.15 \pm 17.8	50.34 \pm 16.3	37.23 \pm 15.5	38.51 \pm 15.4
VGRF	21.14 \pm 4.1	19.13 \pm 3.3	22.09 \pm 5.0	22.09 \pm 5.7
Peak Loading Rate	1045.46 \pm 284.2	925.89 \pm 215.0	1086.42 \pm 329.9	1082.57 \pm 326.6
Average Loading Rate †	372.79 \pm 112.9*	326.60 \pm 105.5	371.34 \pm 163.48	391.6 \pm 177.3
Hip Excursion	41.40 \pm 17.9	41.68 \pm 18.7	41.99 \pm 17.9	42.29 \pm 17.3
Knee Excursion	59.97 \pm 10.4	59.30 \pm 9.6	60.61 \pm 14.7	60.98 \pm 13.4
Ankle Excursion	39.4 \pm 7.3	37.73 \pm 6.3	40.61 \pm 7.9	39.8 \pm 6.8
Hip at Initial Contact	43.16 \pm 11.7	43.50 \pm 10.6	39.76 \pm 8.5	40.20 \pm 9.7
Knee at Initial Contact	29.52 \pm 7.7	31.42 \pm 6.3	28.69 \pm 6.1	29.77 \pm 6.6
Ankle at Initial Contact ¥	-16.16 \pm 5.8	-13.97 \pm 5.3	-14.32 \pm 5.8*	-12.33 \pm 5.7
Peak Hip Extension Moment	-3.81 \pm 1.2	-3.62 \pm 1.2	-4.12 \pm 1.8	-4.25 \pm 1.8
Peak Hip Flexion Moment	2.61 \pm 0.9	2.49 \pm 0.8	2.87 \pm 0.8	2.88 \pm 0.9
Peak Knee Extension Moment	2.29 \pm 0.3	2.26 \pm 0.4	2.46 \pm 0.5	2.40 \pm 0.4

Table 3.3 Mean \pm standard deviations for biomechanical variables measured during the crossover hop in the dominant and non-dominant limb in both the CTRL and the LBP groups. ¥ Denotes significant limb differences via the RM ANOVA. *Denotes significant differences via post hoc analysis. Significance determined by $p \leq 0.05$.

	CTRL		LBP	
	Dominant	Non- Dominant	Dominant	Non-Dominant
Peak Tibia Impact	12.89 \pm 1.9	12.37 \pm 2.2	12.35 \pm 2.1	12.56 \pm 1.9
Peak Pelvis Impact	4.96 \pm 2.0	4.56 \pm 1.4	5.41 \pm 2.4	5.28 \pm 2.3
Acceleration Reduction	38.97 \pm 16.5	37.76 \pm 11.6	44.23 \pm 18.5	41.85 \pm 16.3
VGRF	30.45 \pm 4.7	30.58 \pm 6.0	30.17 \pm 5.4	21.54 \pm 5.8
Peak Loading Rate	1650.01 \pm 451.8	1694.05 \pm 609.3	1715.31 \pm 562.5	1776.08 \pm 603.5
Average Loading Rate	857.8 \pm 331.8	878.70 \pm 430.5	888.65 \pm 387.1	964.23 \pm 422.9
Hip Excursion	6.94 \pm 4.9	5.71 \pm 3.3	7.47 \pm 4.0	7.28 \pm 5.4
Knee Excursion	38.40 \pm 5.0	36.43 \pm 3.1	37.75 \pm 7.5	37.30 \pm 6.9
Ankle Excursion	24.02 \pm 7.7	21.51 \pm 8.5	22.04 \pm 7.6	23.08 \pm 9.8
Hip at Initial Contact	42.93 \pm 9.4	41.31 \pm 10.2	41.94 \pm 8.9	40.35 \pm 8.1
Knee at Initial Contact	19.08 \pm 3.7	19.18 \pm 4.7	19.43 \pm 3.9	18.96 \pm 4.8
Ankle at Initial Contact	-1.79 \pm 8.9	0.42 \pm 9.8	1.06 \pm 6.7	0.58 \pm 10.4
Peak Hip Extension Moment	-4.97 \pm 1.2	-4.67 \pm 1.2	-4.83 \pm 1.6	-4.89 \pm 1.6
Peak Knee Extension Moment	3.00 \pm 0.5	2.97 \pm 0.5	2.96 \pm 0.6	2.86 \pm 0.6

Table 3.4 Mean \pm standard deviations for biomechanical variables measured during the single limb hop in the dominant and non-dominant limb in both the CTRL and the LBP groups. †Denotes significant group x limb interactions via the RM ANOVA. ¥ Denotes significant limb differences via the RM ANOVA. *Denotes significant differences via post hoc analysis. Significance determined by $p \leq 0.05$.

	CTRL		LBP	
	Dominant	Non- Dominant	Dominant	Non-Dominant
Peak Tibia Impact †	13.87 \pm 1.7	13.59 \pm 1.5	12.84 \pm 1.9*	14.24 \pm 1.3
Peak Pelvis Impact	7.22 \pm 3.3	6.73 \pm 2.8	7.23 \pm 3.9	7.28 \pm 3.7
Acceleration Reduction	51.81 \pm 21.9	49.16 \pm 18.2	57.95 \pm 31.1	51.10 \pm 25.1
VGRF	35.94 \pm 5.8	34.77 \pm 4.0	34.53 \pm 4.9	35.08 \pm 4.9
Peak Loading Rate	2012.05 \pm 573.7	2005 \pm 540.0	1976.97 \pm 406.1	2103.40 \pm 456.9
Average Loading Rate ¥	1029.48 \pm 256.0	1080.85 \pm 372.1	1011.61 \pm 260.8	1092.21 \pm 342.4
Hip Excursion	14.30 \pm 7.0	13.93 \pm 7.2	12.64 \pm 4.9	14.61 \pm 6.4
Knee Excursion	41.38 \pm 5.3	39.53 \pm 5.7	39.54 \pm 7.5	39.11 \pm 5.3
Ankle Excursion ¥	12.63 \pm 8.3*	9.74 \pm 5.4	12.41 \pm 6.6	10.32 \pm 7.47
Hip at Initial Contact †	43.99 \pm 10.4	45.87 \pm 10.4	47.09 \pm 10.6	45.06 \pm 8.4
Knee at Initial Contact	16.38 \pm 5.7	17.22 \pm 5.7	17.69 \pm 4.9	17.89 \pm 4.8
Ankle at Initial Contact	-6.35 \pm 8.1	-4.86 \pm 6.3	-5.53 \pm 6.2	-3.78 \pm 8.0
Peak Hip Extension Moment	-6.11 \pm 1.8	-5.88 \pm 1.5	-6.14 \pm 2.0	-6.29 \pm 1.9
Peak Knee Extension Moment	3.04 \pm 0.5	2.95 \pm 0.4	3.12 \pm 0.7	3.09 \pm 0.7

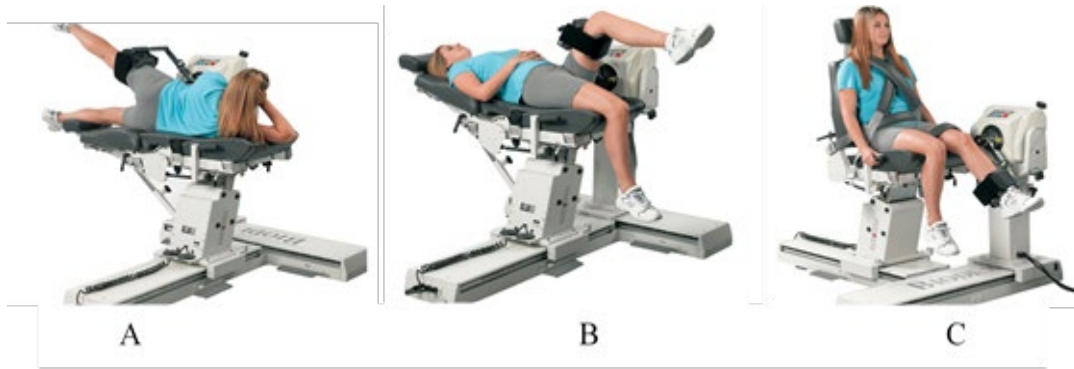


Figure 3.1 A: Biodex setup for hip abduction strength. B: Biodex setup for hip extension strength. C: Biodex setup for knee extension strength.

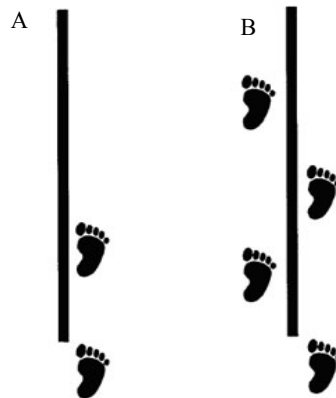


Figure 3.2 Visual representation of the single limb forward hop for distance and the single limb crossover hop for distance from Noyes et al.[97]

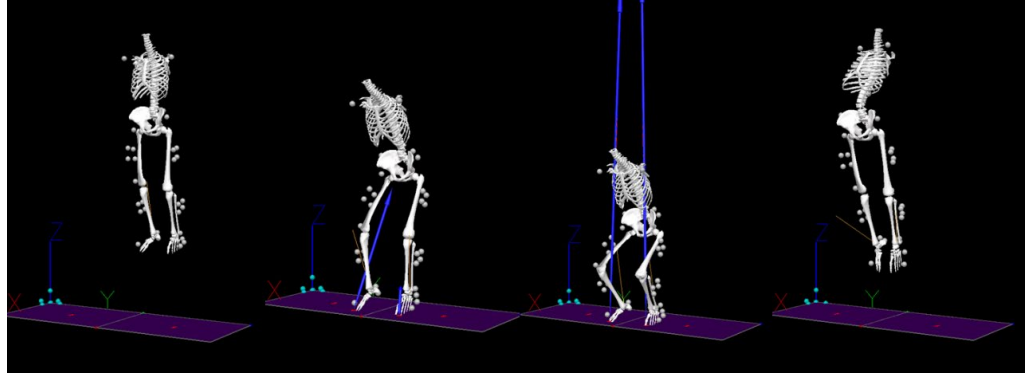


Figure 3.3 Visual representation of the drop vertical jump.



Figure 3.4 Posterior view of marker placement.



Figure 3.5 Inertial measurement unit placement on the medial distal tibia, and attachment strap representation.

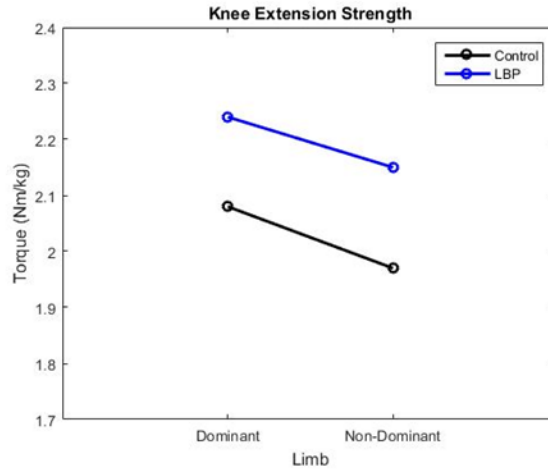


Figure 3.6 Knee extension strength differences between dominant and non-dominant limbs, as well as between CTRL and LBP groups. The black line represents the CTRL differences between limbs, and the blue line represents the LBP differences between limbs

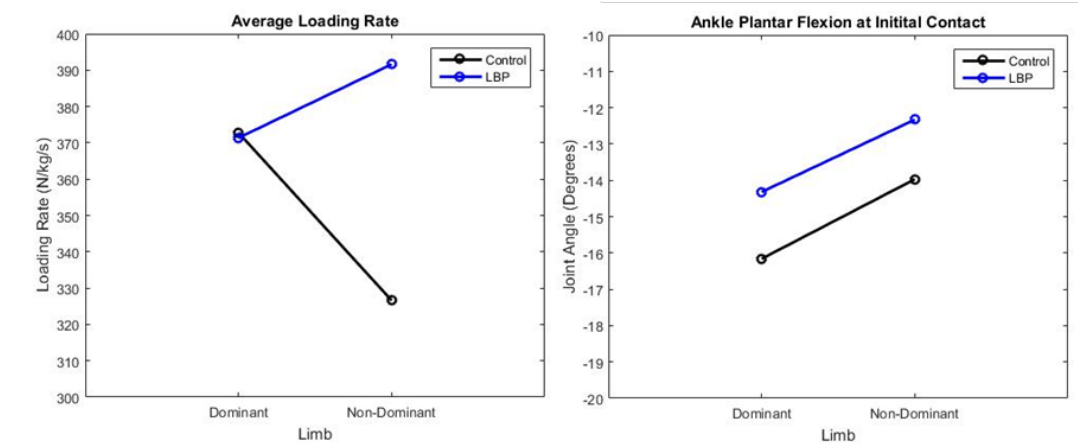


Figure 3.7 Average loading rate and ankle plantar flexion angles at initial contact differences between limbs and groups. The black line represents the CTRL differences between limbs, and the blue line represents the LBP differences between limbs.

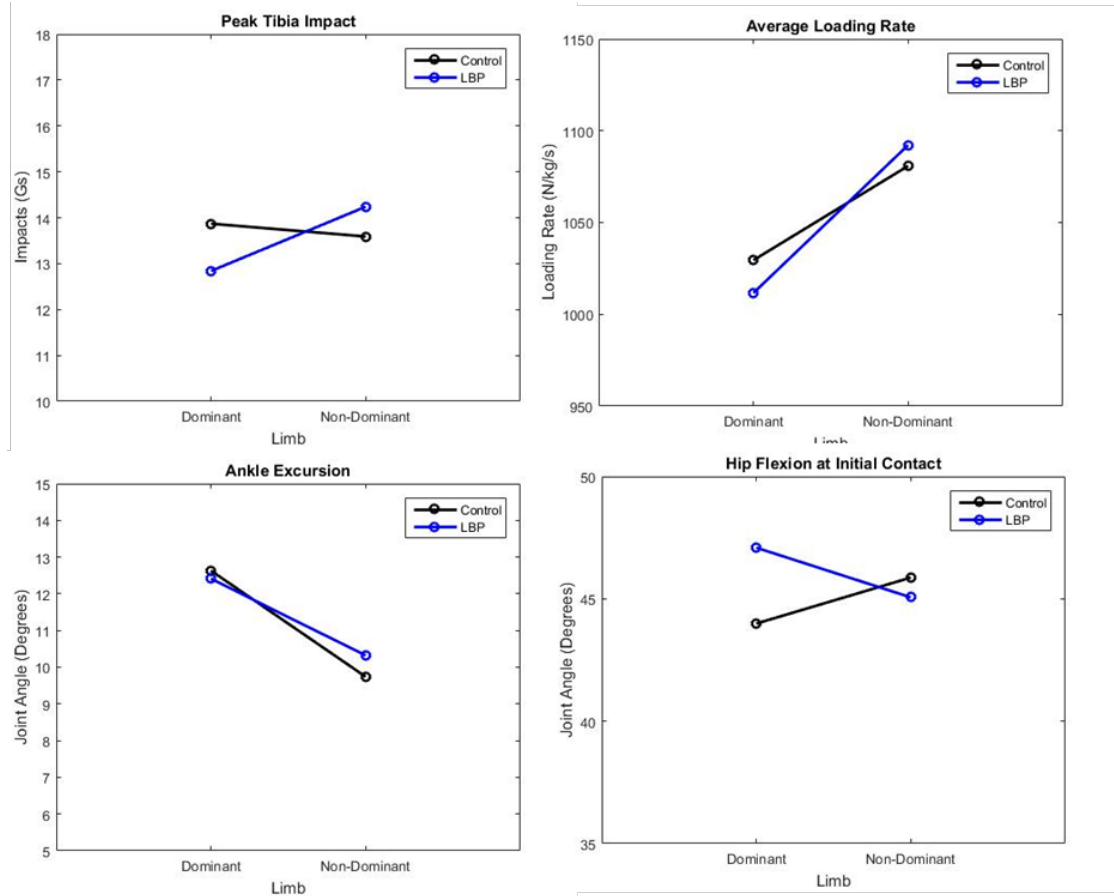


Figure 3.8 Peak tibia impact, average loading rate, ankle excursion, and hip flexion at initial contact differences between limbs and groups. The black line represents the CTRL differences between limbs, and the blue line represents LBP differences between limbs.

CHAPTER 4. CLINICAL DIFFERENCES BETWEEN ACTIVE INDIVIDUALS WITH AND WITHOUT CHRONIC LOW BACK PAIN AND THEIR RELATION TO LOWER EXTREMITY MECHANICS

Introduction

Low back pain (LBP) is one of the most prevalent conditions in the general population, and is also a common medical presentation among physically active individuals, affecting up to 37% of athletes in a single year.[9] LBP affects athletes of all ages,[7, 9, 21] and of varying sports including soccer, gymnastics, rowing, handball, ice hockey, field hockey, basketball, and rugby[8, 150]. Factors such as high training volume, physical loads, repetitive motions, strains, forced body positions, and contact may influence the prevalence of LBP in athletes.[150] In the athletic population, LBP of a lesser intensity may not disrupt participation despite significant discomfort, which can increase the risk for further injury.[11] LBP of a greater intensity may have negative consequences on performance and result in time loss from participation.[8, 11] It has been well documented that a previous history of LBP is a risk factor for later occurrences of LBP,[11, 151] and secondary lower extremity musculoskeletal deficits, such as knee laxity. [9, 25]

Patient reported outcomes, such as the Oswestry Disability Index, is often used to help clinicians understand how LBP effects individuals lives.[99, 131] Most patient reported outcomes are specific to an area of location on the body, injury, or disease. Regional interdependence is the theory that pain or an injury in one location can lead to pain or injury beyond the original location of pain.[27, 123] In this case, individuals who suffer from LBP may have pain or develop injuries in their lower extremities, such as knee pain. It possible that some patient reported outcomes, not specific to one's injury, may

help clinicians understand how their patients' current injury may affect additional body locations and overall function. Regional interdependence has been identified in individuals with LBP who have presented with altered mechanics at the knee.[39] Similarly interdependence results in subsequent hip and knee osteoarthritis.[28] Individuals with both osteoarthritis at the hip or knee and LBP have been shown to report worse overall function, higher pain, and greater joint stiffness than those with hip or knee osteoarthritis without LBP.[26] The Knee Osteoarthritis Outcomes Score (KOOS) is used to identify knee related challenges, and has been linked to identifiers leading to possible knee injuries[30] and knee osteoarthritis.[152] Individuals with lower KOOS quality of life scores are at a 33% greater risk of a secondary ACL tear.[30] Similarly, individuals with a lower limb injury had a high rate leading to knee osteoarthritis, and those with knee osteoarthritis presented with lower KOOS scores.[32] Therefore, patient reported outcomes for the knee may provide insight to into how individuals with LBP may suffer beyond the low back.

It has also been documented that individuals with LBP present with reduced vertical ground reaction forces during walking[41], potentially a pain avoidance strategy. For example, individuals with LBP have presented with a 20% reduced ability to absorb shock during walking.[44] Additionally, individuals with LBP run with greater knee stiffness[39], and less trunk and pelvis joint coordination.[23, 118] Despite these findings, there has not been a well-informed way for clinicians to identify such altered mechanics. Three dimensional (3D) motion capture systems are the gold standard for assessing these altered mechanics in human movement. As most rehabilitation clinics do not have access to the 3D motion capture systems to assess loading mechanics, therefore patient reported

outcomes may help identify LBP patients with reduced knee function and indirectly altered mechanics.[32] Patient perceived function and muscle strength have demonstrated significant relationships with loading mechanics in a variety of different clinical populations.[35, 153-157] Understanding these relationships may be beneficial for clinicians to make more informed decisions when treating a LBP population. Thus, the purpose of this study was two-fold, first to determine how individuals with LBP present clinically compared to healthy individuals, and second to determine how lower extremity mechanics during landing relate to clinical outcome measures in individuals with LBP. We first hypothesized that individuals with LBP would present with less favorable clinical outcome measures than healthy individuals. Additionally, we hypothesized that altered lower extremity mechanics and shock attenuation will be related to lower Knee Injury and Osteoarthritis Outcome Scores and reduced lower extremity isometric strength.

Methods

Participants

This study was a cross sectional case control examining the differences between individuals who suffer from low back pain (LBP) and healthy controls (CTRL), and further the way in which individuals with LBP move, may be related to clinical outcome measures. Individuals with LBP were recruited from local sports medicine clinics, athletic trainers, fitness centers, and the general population. Patients were eligible to participate in this study if they were currently suffering from chronic LBP, which has persisted for at least four months, had not sustained a lower extremity injury in the past year, had never undergone

lower extremity surgery, were not experiencing radicular symptoms, had not been diagnosed with a musculoskeletal deformity, such as scoliosis or hip dysplasia, scored greater than a 10% disability on the Oswestry Disability Index (ODI), and a minimum of a 5 on the Tegner Physical Activity Scale (Appendix 2). CTRL participants were recruited from an active population and included if they did not have a history of LBP, injury, or surgery, a history of lower extremity surgery, had a lower extremity injury within the last year, scored lower than a 10% on the ODI. Healthy control participants were matched to the LBP patients based on mass (± 5 kgs), age (± 2 years), Tegner physical activity score (± 2 points), dominant leg (choose to kick a soccer ball), and sex (assigned at birth). All individuals read and signed an informed consent approved by the University's Institutional Review Board. Participation in this study included one visit to the University of Kentucky's Sports Medicine Research Institute to complete clinical outcome measures that included patient reported outcomes, lower extremity strength measures, and a biomechanical analysis of movement strategies during different functional tasks.

Clinical Outcome Measures

All individuals completed clinical outcome measures prior to the biomechanical analysis. Individuals completed two patient reported outcome questionnaires and maximal lower extremity strength measurements.

Patient Reported Outcomes

Individuals completed the ODI (Appendix 1), a patient reported outcome geared toward understanding an individual's self-perceived low back function in different areas of life.[99, 131] The ODI has ten sections including pain intensity, personal care, lifting,

walking, sitting, standing, sleeping, sex life, social life and traveling, allowing individuals to identify where they fall for each section on a five-point Likert scale. A total score of 50 points is possible on the ODI. To calculate the total score, the following equation is used: $(\text{Individual Score} / \text{Total Possible Score of 50}) \times 100$ in which the score is reported in a percent format i.e. ODI = 22%.[99] A higher score ODI indicates greater pain and greater functional disability due to the pain. A score from 0-20% indicates minimal disability, 21-40% is a moderate disability, 41-60% is a severe disability, 61-80% is considered crippled, and 81-100% is indicated as bedridden or an exaggeration (Appendix 1).[99]

Individuals also completed the Knee Injury and Osteoarthritis Outcomes Score (KOOS; Appendix 3), a patient reported outcome geared toward understanding an individual's self-perceived knee function in different areas of life.[152, 158, 159] The KOOS provides separate scores for each of the five sections that include knee symptoms (SYM), pain, quality of life (QOL), activities of daily living (ADL), and sport and recreation (SR). The KOOS uses a 5-point Likert scale for each question. To score each section of the KOOS the number of questions in the section over the score on the Likert scale was converted to out of 100 points. KOOS section scores range from 0-100, with a 100 being considered a perfectly healthy score, indicating no knee problems.[152, 158]

Lower Extremity Strength

Lower extremity hip and knee strength was measured using a Biodex System 4 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY). Isometric hip abduction was measured side lying, with the hip abducted at 0 degrees (Figure 4.1).[65] Isometric hip extension was measured in the supine position with the hip flexed at 60 degrees (Figure 4.1). Isometric knee extension was measured in a seated position with the knee flexed at

60 degrees (Figure 4.1). For each test, individuals received two practice trials warming up at 50% effort, then one practice trial at 100% effort, before completing three maximal effort trials with 30 seconds rest between each trial. Verbal encouragement was provided throughout the test. Peak torque across the trials for each joint direction and limb was recorded and normalized to body mass.

Biomechanical Outcome Measures

Functional Tasks

Biomechanical outcome measures were collected during a single limb forward hop for distance (Figure 4.2). Individuals were allowed as many practice trials as they deemed necessary to feel comfortable with each movement, then five successful trials of each functional task, were recorded for biomechanical analysis. The single limb hop followed Noyes et al.[97] specifications (Figure 4.2). In the single limb hop the starting line was placed so individuals must stick the landing on one of two in ground force plates. During the crossover hop the starting line was placed to capture the second landing on one of two force plates for kinetic measurements. A single limb hop in which any part of the foot did not land on the force plate was considered a bad trial and individuals were asked to give another effort. Individuals must also have stuck the final landing, which was identified by no double hops, pivoting, shifting, touching the other foot to the ground, or touching the ground with their hands. Individuals were instructed to complete the single limb hops by jumping as far as possible while still being able to stick the landing. Pain was evaluated after every task using a visual analog scale, rating pain from 0-10.[133]

Three Dimensional Motion Capture

Three dimensional motion capture was used to examine trunk and lower extremity biomechanics during the three different functional tasks. Trunk and lower extremity segments were defined and tracked using 14mm markers placed at 7th cervical vertebrae, bilateral acromion processes, the sternum, xiphoid process, the 12th thoracic vertebrae, as well as bilateral iliac crests, anterior superior iliac spines, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, proximal and distal heel, mid-foot distal to the lateral malleoli, and the head of 1st and 5th metatarsals (Figure 4.3). Additionally, tracking clusters, created by rigid thermoplastic with four 9mm markers affixed, were placed over the pelvis at the posterior superior iliac spine and bilaterally on the lateral thighs and shanks. After the static calibration, medial and lateral markers placed on anatomical landmarks of the lower extremity were removed and only tracking markers were used for the data collection during the dynamic movements. Kinematics were collected at 200Hz with a 14 camera three dimensional motion capture system (Vicon, Centennial, CO) and ground reaction forces captured at 2000Hz, on two in ground force plates (Bertec Corporation, Columbus, OH).

Inertial Measurement Units

Lower extremity vertical accelerations were collected using three, 9- axis, inertial measurement units (IMUs; I Measure U, Vicon, Centennial, CO). Acceleration data were collected in the Vicon Nexus software for timing synchronization to motion capture data, sampled at 500Hz. IMUs were placed on the sacrum (directly underneath the pelvis cluster used for motion capture), and approximately 2cm superior to the medial malleolus on the shank (Figure 4.4).

Data Analysis

Three dimensional motion capture data was analyzed using Visual 3D(C-Motion, Germantown, MD). Marker trajectories were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 6Hz. Ground reaction force data were filtered using a 4th order lowpass Butterworth filter with a cutoff frequency of 50Hz. Lower extremity kinematics were calculated using Euler angles. Only lower extremity variables that effect knee loading during landing were included in this analysis, as the goal was determine their relationship to a knee osteoarthritis health related patient reported outcome. Peak vertical ground reaction forces (PVGRF), average loading rate, and peak knee and ankle flexion angles were identified during the landing phase of the single limb hop. Joint angle at initial contact for the knee and ankle were identified at the initiation of ground contact on the force plates when the vertical ground reaction force exceeded 10N. Landing phase sagittal plane knee and ankle excursions were calculated as the difference between the peak joint angle during landing and the joint angle at initial contact. Average loading rate (ALR) was assessed as the mean of the derivative of the vertical ground reaction force curve from ground contact, indicated by when the force plates recorded a reading greater than 50N,[102] to PVGRF.

Raw IMU data were analyzed in Visual 3D to calculate acceleration reduction and peak impacts in the vertical direction during landing. In the single limb hop, landing was defined as initial contact with the force plate (greater than 20N) to 100ms after landing. Peak tibia and peak pelvis impacts during landing were determined as the peak positive acceleration from initial contact with the force plates to peak knee flexion. Acceleration

reduction was the percentage of peak impact between the pelvis and the tibia sensor as $(\text{peak pelvis impact}/\text{peak tibia impact})/100$ [78, 82]. Thus, the reported number is the percentage of the amount of impact that was not absorbed from the tibia to the pelvis, indicating that a larger percentage indicates less favorable shock absorption during landing.

Statistics

Group differences were assessed to determine if clinical outcome measures beyond the back may provide benefit to clinicians in guiding rehabilitation. Independent t-tests were run to determine differences in KOOS scores, peak hip abduction strength, peak hip extension strength, and peak knee extension strength, between the LBP and the CTRL group. An alpha value of $\alpha=0.05$ was used to determine group differences using SPSS (SPSS 25, IBM Corporation, Armonk, NY).

In the LBP group, relationships between loading mechanics variables and clinical outcome measures were assessed using Pearson Correlation Coefficients, confirmed via scatterplots. Correlation coefficients were indicated as little to no relationship with a correlation coefficient between 0.00-0.25, a fair relationship between 0.25-0.50, a moderate to good relationship between 0.50-0.75, and a good to excellent relationship above 0.75, in which a correlation coefficient of 1.00 indicates a perfect relationship.[103] Multiple linear forward regressions were used to determine if KOOS scores and lower extremity strength variables could predict loading mechanics during the single limb hop. Only loading mechanics variables that exhibited a relationship with clinical outcome measures were included in the regression analysis. Additionally, sex, as a variable was

entered into every regression model to account for well-known sex differences in individuals with LBP.[18, 25, 67] Models were compared and the highest adjusted R² was identified as the best model. An alpha value of $\alpha=0.05$ was used for all regression models, using SPSS (SPSS 25, IBM Corporation, Armonk, NY).

Results

Twenty-eight individuals who suffered from low back pain (LBP) and twenty-eight healthy control (CTRL) individuals completed this study. There were no significant differences between groups in demographic variable such as height (LBP: $1.7\pm0.07\text{m}$, CTRL: $1.7\pm0.08\text{m}$; $p=0.896$), mass (LBP: $72.06\pm12.1\text{kgs}$, CTRL: $72.66\pm13.2\text{kgs}$; $p=0.861$), age (CTRL: 24.6 ± 4.5 , LBP: $25.0\pm4.8\text{yrs}$; $p=0.785$) and activity level (CTRL: 7.0 ± 1.1 , LBP: 6.7 ± 1.1 ; $p=0.359$). While subjects between groups were matched in terms of sex, there were more females (21) than males (7) that presented with LBP. ODI scores were significantly different between groups, as expected due to the fact that a minimum ODI score of 10% was a requirement for inclusion in the study. Every individual in the CTRL group reported an ODI score of 0%, while in the LBP group scores ranges from 10-34%, with the average ODI score being 16.4%.

Individuals with LBP presented with worse perceived knee function, lower KOOS scores, compared to the CTRL group in four KOOS subscales (Table 4.1), including SYM ($p=0.007$), Pain ($p=0.002$), QOL ($p=0.021$), and ADL ($p=0.003$). There were no significant differences between the two groups in the KOOS Sports and Recreation Score. Strength including, peak knee extension strength, peak hip abduction

strength or peak hip extension strength was not significantly different between groups for either the dominant or the non-dominant limb.

Seven loading mechanics variables exhibited a relationship with clinical outcome measures (Appendix 4). Dominant limb peak pelvis impact showed a moderate negative relationship to ADL scores ($r=-0.645$, $p=0.003$). Dominant limb acceleration reduction also showed moderate negative relationships to ADL scores ($r=-0.622$, $p=0.006$), dominant limb hip abduction strength ($r=-0.559$, $p=0.030$) and dominant limb hip extension strength ($r=-0.558$, $p=0.031$). Dominant limb PVGRF showed a fair relationship to dominant limb peak knee extension strength ($r=0.428$, $p=0.033$). Dominant limb ALR showed a fair relationship to dominant limb peak knee extension strength ($r=0.463$, $p=0.020$). Non-dominant acceleration reduction showed moderate negative relationships to QOL ($r=-0.521$, $p=0.032$), SR ($r=-0.500$, $p=0.021$), and non-dominant limb peak knee extension strength ($r=-0.534$, $p=0.027$). Non-dominant limb ALR showed a positive moderate relationship to non-dominant peak knee extension strength ($r=0.662$, $p=0.0001$) and a positive fair relationship to non-dominant peak hip extension strength ($r=0.451$, $p=0.035$). Non-dominant knee excursion showed a positive fair relationship to QOL ($r=0.452$, $p=0.016$). Linear regressions were used to predict loading mechanics from the clinical outcome measures that demonstrated significant correlations. Three of the seven linear regressions provided a significantly reliable model, including dominant limb peak pelvis impact, dominant limb acceleration reduction and non-dominant limb ALR. Sex and ADL significantly explained 35.6% of the variance peak pelvis impact while landing on the dominant limb ($r^2=0.356$, $p=0.014$, Figure 4.5A). Sex and ADL significantly explained 35.5% of the variance of acceleration reduction during landing on the dominant limb

($r^2=0.355$, $p=0.029$, Figure 4.5B). Sex and non-dominant peak knee extension strength explained 50.2% of the variance of non-dominant ALR ($r^2=0.502$, $p=0.001$, Figure 4.5C).

Discussion

The overall purpose of this study was to determine if individuals with LBP exhibited worse clinical outcome measures of knee function compared to healthy individuals, and determine if self-perceived levels of knee function could predict mechanical loading. Our first hypothesis was supported; KOOS scores were lower in individuals with LBP compared to healthy individuals. Individuals with LBP presented with worse self-reported knee function in terms of their symptoms, pain, quality of life, and activities of daily living. Understanding how LBP may influence self-perceived knee function could be important to clinicians when determining rehabilitation methods. Individuals in the LBP group report the lowest KOOS score in the QOL section, with the majority of individuals with LBP reporting that they are aware of knee problems at least monthly. Individuals with LBP are more likely to suffer from knee osteoarthritis than those without LBP.[26, 28] Assessing knee function in individuals with low back pain may provide the clinician with a more holistic view of the patients overall levels of physical function and performance. While total scores of individuals with LBP are not quite to the magnitude as individuals who have recently sustained a knee injury, they are worse than a healthy population, and seem to be progressively declining. Thus, beneficial for clinicians to assess self-perceived knee function to potentially delay the onset of any further complications. While the majority of rehabilitation for those with LBP focuses on core strength and back stability, using patient reported outcomes that focuses on knee function

may help identify individuals at the greatest risk of developing a secondary disease, such as knee osteoarthritis.

While group differences between KOOS scores were small, they were still significant. The minimal detectable change for a knee injury population in the KOOS ranges from 5 points to 12 points, depending on the sub scale.[159] While our LBP group did not meet the criteria for minimal detectable change, all differences were greater than the standard error of measurement in a knee injury group.[159] Our LBP group is highly active and still participating in sports, it is possible that as our population ages these scores may progress to what is considered clinically significant and should not be overlooked, as effect sizes were moderate to large (Table 4.1). Both KOOS Pain and QOL have strong effect sizes, at 0.92 and 0.84 respectively, and the narrowest 95% confidence intervals, not crossing zero. With this decrease in KOOS scores, it may be possible that the mechanical compensations from LBP may be affecting self-perceived knee function. Similarly, KOOS SYM and ADL indicated moderate effect sizes at 0.75 and 0.67 respectively, also with narrow 95% confidence intervals. These differences in clinical outcome measures of knee function may be explained by common fear avoidance or fear of re-injury strategies that are typically displayed by those with LBP. It is well documented that individuals with LBP are more likely to experience pain avoidance strategies in the form of reduced joint range of motion,[160] reduced lumbar extension strength,[115] and a fear of re-injury.[161] It is possible that individuals with LBP present with worse KOOS scores due to pain avoidance strategies. Overall, pain avoidance may drive mechanical compensations that include the lower extremities, which over time, could potentially influence how individuals with LBP perceive their knee function.

The second purpose of this study was to determine how lower extremity mechanics during landing related to clinical outcome measures in individuals with LBP. We hypothesized that altered lower extremity mechanics and acceleration reduction would be related to lower KOOS scores and reduced lower extremity isometric strength. On the non-dominant limb, greater knee extension strength and sex was predictive of 50.2% of the variance of higher non-dominant limb loading rates during the landing of the single limb hop (Figure 4.5). Additionally, lower KOOS ADL scores and sex were predictive of 36.5% of the variance of higher peak pelvis impact while landing on the dominant limb and 35.5% of the variance of poor acceleration reduction during the landing of the single limb hop (Figure 4.5). For the sake of simplicity, focusing on peak pelvis impact over acceleration reduction may be more clinically feasible, as peak pelvis impact is part of the acceleration reduction calculation, and it is easier to assess and interpret. Peak pelvis impact was assessed via a sensor attached directly on top of the sacrum and would be affected by both hip and knee kinematics. During a landing specific task, such as the single limb hop, hip and knee flexion are the primary lower extremity movements required to reduce pelvis impact.[61, 94]

On the other hand, peak tibia impact did not present with any significant relationships to clinical outcome measures. This may be due to the fact that the IMU placed on the tibia is inferior to the knee joint and though it represents the accelerations traveling to the knee, it does represent the accelerations attenuated at the knee, which may have a significant impact on knee function. It is possible that higher loads traveling through the knee and making it to the pelvis, are represented by poor self-perceived knee function. Further, ADL scores were predictive of mechanical pelvis loading when landing on the

dominant limb, while strength was the only predictive variable on the non-dominant limb. It is possible that in individuals with LBP the dominant limb drives their perception of function.

Previous literature has identified that individuals with LBP present with decreased hip extension and hip abduction strength,[18, 25, 33] this was conflicting to what we found, a lack of strength differences between individuals with LBP and the CTRL group. In addition, contrary to what was expected, we found that on the non-dominant limb, knee extension strength, which is primarily quadriceps strength, predicted higher average loading rates. Although the performance levels of the LBP group during the single limb hop was not included in this analysis, it is possible that they jumped higher and further, leading to higher loading rates. This relationship may be explained by the idea that individuals with LBP exhibit a dominant limb dependency strategy[57] and when required to use the non-dominant limb to complete a functional task they may not have adequate absorption strategies. The unfamiliarity of moving on their non-dominant limb may also help explain why ineffective absorption strategies were present.

Even though these individuals exhibit LBP it is also possible that they may not be exhibiting compensations during functional tasks, as the pain may not be great enough to alter motor pathways to cause compensations. Individuals who have undergone anterior cruciate ligament reconstruction, that are stronger have been shown to perform better on hop testing.[42]

It may be possible that in the LBP group, those that demonstrate greater knee extension strength have less LBP severity or may not have been suffering from LBP as

long. The lack of strength differences may explain why active individuals with LBP did not demonstrate the mechanical compensations that were expected.

While there were also no differences in KOOS SR scores, it is possible that individuals with LBP recognize their poor knee function in low level activities, not during sport situations. While this seems opposite than what we would expect with typical injuries, with LBP typically being a non-traumatic injury,[8] it may be that pain is exacerbated following sport participation more often than during sport participation, explaining why we see differences in the KOOS scores focusing on lower intensity functional tasks.

Biomechanical characteristics associated with the onset and progression of knee osteoarthritis are cumulative overtime.[145] Individuals suffering from LBP for longer periods of time, or may present with a greater levels of back pain, may demonstrate different mechanical compensation strategies, and may have increasingly worse self-perceived knee function. It would be advantageous for future research to examine if the amount of time an individual suffers from LBP or if their level of pain has an effect on their self-perceived knee function, especially in an athletic population where exposure to risk of injury is high. One limitation of this study is while we identified a minimum level of pain for inclusion in the LBP group, all individuals in this study, despite level of disability on the ODI were still actively participating in their sport. It is possible that the ODI may not be specific enough to an active population with LBP to tease out mechanical compensations that previous literature has reported, like weaker hip abduction and hip extension strength.

Another limitation of this study may have been the inability of individuals with LBP to separate knee and low back limitations, when interpreting KOOS questions. While it was emphasized that this survey asked questions related to their knee function and knee pain it is possible that some individuals had difficulty deciphering functional limitations specific to the knee. For example, the KOOS includes questions focused on difficulty during standing, sitting, and rising from sitting (Appendix 3), all of which may be affected by back pain and knee pain.

While assessing a non-traumatic injury population is not always clear, it was evident that individuals with LBP commonly present with decreased self-perceived knee function compared to active individuals without LBP. Further it was presented that self-perceived knee function during activities of daily living predicted a significant portion of loading variables, specifically pelvis impact and acceleration reduction on the dominant limb during a single limb hop. Identifying these relationships indicates clinicians may be able to use these patient reported outcomes as predictors of biomechanical outcomes for active individuals with LBP. Clinicians may be able to include these patient reported outcomes in order to identify potential lower extremity loading mechanics in active individuals with LBP. This may be able to enhance rehabilitation programs, delaying the onset of risk factors associated with secondary injuries and reduced long-term musculoskeletal health.

Table 4.1 Mean \pm standard deviations for clinical outcome measures in both the CTRL and the LBP groups. *Denotes significant differences between groups where significance determined by $p \leq 0.05$.

	CTRL	LBP	P-Value	Hedges G Effect Size (95% Confidence Interval)
KOOS – SYM*	98.2 \pm 5.1	93.3 \pm 7.5	0.007	0.75 (0.21, 1.30)
KOOS- Pain*	99.4 \pm 2.6	95.2 \pm 5.8	0.002	0.92 (0.37, 1.47)
KOOS- ADL*	99.8 \pm 0.8	98.5 \pm 2.6	0.021	0.67 (0.13, 1.20)
KOOS-SR	98.5 \pm 4.2	94.8 \pm 9.5	0.066	0.50 (-0.04, 1.03)
KOOS-QOL*	99.1 \pm 3.6	92.6 \pm 10.2	0.003	0.84 (0.29, 1.38)
Dominant Peak Knee Extension Strength	2.3 \pm 0.4	2.2 \pm 0.5	0.768	0.22 (-0.34, 0.78)
Non-dominant Peak Knee Extension Strength	2.1 \pm 0.3	2.1 \pm 0.5	0.864	0.00 (-0.56, 0.56)
Dominant Peak Hip Abduction Strength	1.6 \pm 0.4	1.7 \pm 0.3	0.426	-0.28 (-0.92, 0.35)
Non-dominant Peak Hip Abduction Strength	1.6 \pm 0.3	1.7 \pm 0.5	0.537	-0.23 (-0.87, 0.40)
Dominant Peak Hip Extension Strength	2.0 \pm 0.6	1.8 \pm 0.5	0.347	0.36 (-0.28, 1.00)
Non-dominant Peak Hip Extension Strength	1.8 \pm 0.5	1.8 \pm 0.5	0.817	0.00 (-0.63, 0.63)



Figure 4.1 A: Biodex setup for hip abduction strength. B: Biodex setup for hip extension strength. C: Biodex setup for knee extension strength.



Figure 4.2 Visual representation of the single limb forward hop for distance from Noyes et al.[97]

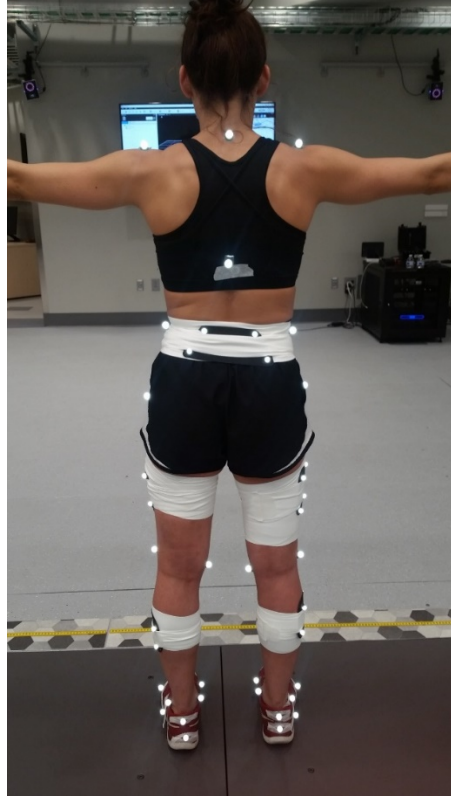


Figure 4.3 Posterior view of marker placement.



Figure 4.4 Inertial measurement unit placement on the medial distal tibia, and attachment strap representation.

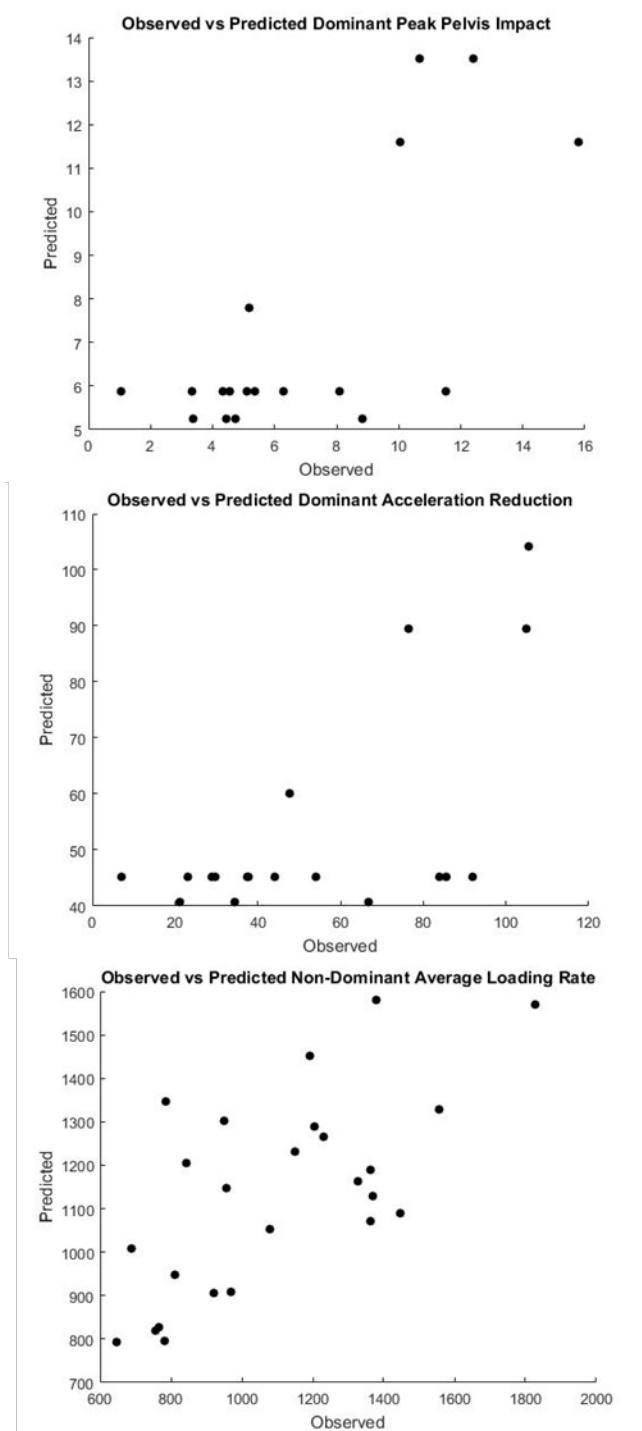


Figure 4.5 Observed data over predicted data using the model equations for the significantly predicted models of dominant limb peak pelvis impact, dominant limb acceleration reduction, and non-dominant average loading rate.

CHAPTER 5. CONCLUSION

Overall the purpose of this project was to determine how chronic low back pain influences lower extremity biomechanics and shock attenuation in active individuals compared to healthy individuals and examine how the altered lower extremity biomechanics are related to clinical outcome measures. The specific aims were as follows:

Specific Aim 1: To determine the relationship between lower extremity kinematics and kinetics and shock attenuation in healthy individuals during a functional landing task.

In aim 1, we found a moderate relationship between lower extremity impact and loading mechanics during landing, specifically that peak tibia and peak pelvis impact presented with significant relationships to loading rate and peak vertical ground reaction forces in a crossover hop. Thus, we believe that impact during landing may be able to be assessed using inertial measurement units (IMUs). Assessing impact during landing via IMUs may provide clinicians with a more objective assessment of their patients, especially athletes returning to sports following injuries. Although IMUs may be considered expensive, they are more affordable than force plates or fully integrated three dimensional motion capture systems, and as the technology advances are proving to be a practical tool in many environments to measure mechanics. IMUs are smaller and may be more user friendly during data collections, using tablet-based applications to provide immediate feedback to clinicians and patients alike. The results from aim 1 not only identify relationships between vertical ground reaction forces and impacts during landing, but may also act as a resource of preliminary normative landing impacts measured with IMUs for

clinicians. Clinicians can also use this mechanical loading information to guide treatment strategies and evaluate treatment effectiveness.

Specific Aim 2: To determine the effects of chronic low back pain on strength, lower extremity biomechanics and shock attenuation during landing compared to a healthy population

Next, in aim 2, we assessed group differences between active individuals with low back pain and healthy individuals. While there were no differences between the two groups, there were interlimb differences present in the low back pain group that did not exist in healthy individuals. Specifically, in the low back pain group peak tibia impact was higher in the non-dominant limb compared to the dominant limb during the single limb hop. Also, during the drop vertical jump individuals with low back pain landed with increased ankle plantar flexion angles at initial ground contact on their dominant limb compared to their non-dominant limb. These findings support the idea of dominant limb dependence strategies in individuals with low back pain, suggesting they present with poor mechanical strategies on their non-dominant limb, relying on the dominant limb for performance. This preliminary analysis of movement strategies during landing may indicate compensation strategies to either avoid pain, or possibly related to neuromuscular deficits. Both pain avoidance and neuromuscular deficits are types of alterations that clinicians should consider when treating individuals with low back pain.

Specific Aim 3: To determine how lower extremity biomechanics during landing relate to clinical outcome measures in individuals with chronic low back pain

In aim 3, active individuals with low back pain presented with worse KOOS scores compared to healthy adults. While the difference in KOOS scores is small between those with low back pain and healthy individuals, the reduction in scores provides important information for clinicians. Clinicians can use questionnaires about self-perceived knee function in individuals with low back pain as a possible way to target rehabilitation. As individualized medicine becomes increasingly important, using self-perceived knee function may be one way clinicians can help direct individualized rehabilitation. Specifically, self-perceived knee function during activities of daily living predict about 35% loading variables during a single limb hop. Identifying this relationship shows that clinicians may be able to utilize these patient reported outcomes as predictors of biomechanical outcomes for active individuals with low back pain, and possibly even in populations with more traumatic injuries in which loading mechanics are increasingly altered.

It is possible that we did not see as strong as relationships or group differences as we had hypothesized due to the active nature of the population. In our case, individuals suffering from low back pain were all still actively participating in their activity and/or sport thus potentially masking mechanical compensations. In addition, while we understand that the tasks that were included in this study were not tasks you would ask a typical low back pain population to experience, for an active population they provide clinical relevance. As these tasks are commonly carried out within sports medicine clinics for progress assessment and return to sport decisions in a number of lower extremity injuries, they may not be the best tasks to highlight compensations in individuals with low

back pain. Future research may benefit from assessing slightly lower intensity and repetitive tasks in which mechanical compensations may be more pronounced, such as running. It should also not be ignored that the first responder and tactical populations are also important in this discussion, as tactical populations present with low back pain at increasing rates compared to athletes. First responders and military personnel consistently sustain impacts greater than those typically seen during activities of daily living and also potentially different than active individuals. Further assessments highlighting tactical athlete's compensations to low back pain would be ideal, as this population doesn't get the chance to slow their activity participation like a typical active individual would. Thus their mechanical compensations may be more pronounced leading to a possible greater risk of secondary injury and reduced musculoskeletal health.

Ultimately, this research is a first step towards understanding a bigger goal of how individuals with low back pain compensate for their pain, how it may lead to secondary injuries or reduced long-term musculoskeletal health, and how we can work with clinicians to improve such outcomes. We first identified that IMUs may be able to be utilized to assess loading during a crossover hop, providing a preliminary analysis of technical yet clinical applicability when evaluating functional tasks. Next, we did not identify any strong mechanical differences between individuals suffering from chronic low back pain and those who do not. Though, we did identify that an active population suffering from low back pain does present with decreased self-perceived knee function compared to active individuals without low back pain. While these groups biomechanically perform similarly, they do not clinically perform the same, specifically, in terms of self-perceived knee function. Such differences should not be overlooked when treating active populations with

low back pain because if this population is presenting with altered self-perceived knee function at a young age, it is likely that it will continue to decline and negatively affect their function.

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APPENDICES

APPENDIX 1. OSWESTRY DISABILITY INDEX QUESTIONNAIRE

Oswestry Low Back Pain Disability Questionnaire

Sources: Fairbank JCT & Pynsent, PB (2000) The Oswestry Disability Index. *Spine*, 25(22):2940-2953.

Davidson M & Keating J (2001) A comparison of five low back disability questionnaires: reliability and responsiveness. *Physical Therapy* 2002;82:8-24.

The Oswestry Disability Index (also known as the Oswestry Low Back Pain Disability Questionnaire) is an extremely important tool that researchers and disability evaluators use to measure a patient's permanent functional disability. The test is considered the 'gold standard' of low back functional outcome tools^[1].

Scoring instructions

For each section the total possible score is 5: if the first statement is marked the section score = 0; if the last statement is marked, it = 5. If all 10 sections are completed the score is calculated as follows:

Example: 16 (total scored)
 50 (total possible score) x 100 = 32%

If one section is missed or not applicable the score is calculated:

 16 (total scored)
 45 (total possible score) x 100 = 35.5%

Minimum detectable change (90% confidence): 10% points (change of less than this may be attributable to error in the measurement)

Interpretation of scores

0% to 20%: minimal disability:	The patient can cope with most living activities. Usually no treatment is indicated apart from advice on lifting sitting and exercise.
21%-40%: moderate disability:	The patient experiences more pain and difficulty with sitting, lifting and standing. Travel and social life are more difficult and they may be disabled from work. Personal care, sexual activity and sleeping are not grossly affected and the patient can usually be managed by conservative means.
41%-60%: severe disability:	Pain remains the main problem in this group but activities of daily living are affected. These patients require a detailed investigation.
61%-80%: crippled:	Back pain impinges on all aspects of the patient's life. Positive intervention is required.
81%-100%:	These patients are either bed-bound or exaggerating their symptoms.

Oswestry Low Back Pain Disability Questionnaire

Instructions

This questionnaire has been designed to give us information as to how your back or leg pain is affecting your ability to manage in everyday life. Please answer by checking ONE box in each section for the statement which best applies to you. We realise you may consider that two or more statements in any one section apply but please just shade out the spot that indicates the statement which most clearly describes your problem.

Section 1 – Pain intensity

- ☐ I have no pain at the moment
- ☐ The pain is very mild at the moment
- ☐ The pain is moderate at the moment
- ☐ The pain is fairly severe at the moment
- ☐ The pain is very severe at the moment
- ☐ The pain is the worst imaginable at the moment

Section 2 – Personal care (washing, dressing etc)

- ☐ I can look after myself normally without causing extra pain
- ☐ I can look after myself normally but it causes extra pain
- ☐ It is painful to look after myself and I am slow and careful
- ☐ I need some help but manage most of my personal care
- ☐ I need help every day in most aspects of self-care
- ☐ I do not get dressed, I wash with difficulty and stay in bed

Section 3 – Lifting

- ☐ I can lift heavy weights without extra pain
- ☐ I can lift heavy weights but it gives extra pain
- ☐ Pain prevents me from lifting heavy weights off the floor, but I can manage if they are conveniently placed eg. on a table
- ☐ Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned
- ☐ I can lift very light weights
- ☐ I cannot lift or carry anything at all

Section 4 – Walking*

- ☐ Pain does not prevent me walking any distance
- ☐ Pain prevents me from walking more than 1 mile
- ☐ Pain prevents me from walking more than 1/2 mile
- ☐ Pain prevents me from walking more than 100 yards
- ☐ I can only walk using a stick or crutches
- ☐ I am in bed most of the time

Section 5 – Sitting

- ☐ I can sit in any chair as long as I like
- ☐ I can only sit in my favourite chair as long as I like
- ☐ Pain prevents me sitting more than one hour
- ☐ Pain prevents me from sitting more than 30 minutes
- ☐ Pain prevents me from sitting more than 10 minutes
- ☐ Pain prevents me from sitting at all

Section 6 – Standing

- ☐ I can stand as long as I want without extra pain
- ☐ I can stand as long as I want but it gives me extra pain
- ☐ Pain prevents me from standing for more than 1 hour
- ☐ Pain prevents me from standing for more than 30 minutes
- ☐ Pain prevents me from standing for more than 10 minutes
- ☐ Pain prevents me from standing at all

Section 7 – Sleeping

- ☐ My sleep is never disturbed by pain
- ☐ My sleep is occasionally disturbed by pain
- ☐ Because of pain I have less than 6 hours sleep
- ☐ Because of pain I have less than 4 hours sleep
- ☐ Because of pain I have less than 2 hours sleep
- ☐ Pain prevents me from sleeping at all

Section 8 – Sex life (if applicable)

- ☐ My sex life is normal and causes no extra pain
- ☐ My sex life is normal but causes some extra pain
- ☐ My sex life is nearly normal but is very painful
- ☐ My sex life is severely restricted by pain
- ☐ My sex life is nearly absent because of pain
- ☐ Pain prevents any sex life at all

Section 9 – Social life

- ☐ My social life is normal and gives me no extra pain
- ☐ My social life is normal but increases the degree of pain
- ☐ Pain has no significant effect on my social life apart from limiting my more energetic interests eg, sport
- ☐ Pain has restricted my social life and I do not go out as often
- ☐ Pain has restricted my social life to my home
- ☐ I have no social life because of pain

Section 10 – Travelling

- ☐ I can travel anywhere without pain
- ☐ I can travel anywhere but it gives me extra pain
- ☐ Pain is bad but I manage journeys over two hours
- ☐ Pain restricts me to journeys of less than one hour
- ☐ Pain restricts me to short necessary journeys under 30 minutes
- ☐ Pain prevents me from travelling except to receive treatment

References

1. Fairbank JC, Pynsent PB. The Oswestry Disability Index. Spine 2000 Nov 15;25(22):2940-52; discussion 52.

APPENDIX 2. TEGNER ACTIVITY LEVEL SCALE

TEGNER ACTIVITY LEVEL SCALE

Please indicate in the spaces below the **HIGHEST** level of activity that you participate in **CURRENTLY**.

CURRENT: Level _____

Level 10	Competitive sports- soccer, football, rugby (national elite)
Level 9	Competitive sports- soccer, football, rugby (lower divisions), ice hockey, wrestling, gymnastics, basketball
Level 8	Competitive sports- racquetball or bandy, squash or badminton, track and field athletics (jumping, etc.), down-hill skiing
Level 7	Competitive sports- tennis, running, motorcars speedway, handball Recreational sports- soccer, football, rugby, bandy, ice hockey, basketball, squash, racquetball, running
Level 6	Recreational sports- tennis and badminton, handball, racquetball, down-hill skiing, jogging at least 5 times per week
Level 5	Work- heavy labor (construction, etc.) Competitive sports- cycling, cross-country skiing, Recreational sports- jogging on uneven ground at least twice weekly
Level 4	Work- moderately heavy labor (e.g. truck driving, etc.)
Level 3	Work- light labor (nursing, etc.)
Level 2	Work- light labor Walking on uneven ground possible, but impossible to back pack or hike
Level 1	Work- sedentary (secretarial, etc.)
Level 0	Sick leave or disability pension because of knee problems

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

SURGICAL HISTORY

Have you had any additional surgeries to your knee?

Yes____ No____

If Yes: _____

What procedure(s) were performed? _____

When was the surgery performed? _____

Who performed the surgery? _____

APPENDIX 3. KNEE OSTEOARTHRITIS OUTCOMES SCORE QUESTIONNAIRE

Knee injury and Osteoarthritis Outcome Score (KOOS), English version 1.K1.0

KOOS KNEE SURVEY

Today's date: ____/____/____ Date of birth: ____/____/____

Name: _____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.

Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms

These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

Stiffness

The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

Pain

P1. How often do you experience knee pain?

Never Monthly Weekly Daily Always

☐☐☐☐☐

What amount of knee pain have you experienced the last week during the following activities?

P2. Twisting/pivoting on your knee

None Mild Moderate Severe Extreme

☐☐☐☐☐

P3. Straightening knee fully

None Mild Moderate Severe Extreme

☐☐☐☐☐

P4. Bending knee fully

None Mild Moderate Severe Extreme

☐☐☐☐☐

P5. Walking on flat surface

None Mild Moderate Severe Extreme

☐☐☐☐☐

P6. Going up or down stairs

None Mild Moderate Severe Extreme

☐☐☐☐☐

P7. At night while in bed

None Mild Moderate Severe Extreme

☐☐☐☐☐

P8. Sitting or lying

None Mild Moderate Severe Extreme

☐☐☐☐☐

P9. Standing upright

None Mild Moderate Severe Extreme

☐☐☐☐☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A1. Descending stairs

None Mild Moderate Severe Extreme

☐☐☐☐☐

A2. Ascending stairs

None Mild Moderate Severe Extreme

☐☐☐☐☐

For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A4. Standing

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A5. Bending to floor/pick up an object

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A14. Sitting

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

A17. Light domestic duties (cooking, dusting, etc)

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the last week due to your knee.

SP1. Squatting

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

SP2. Running

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

SP3. Jumping

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

SP4. Twisting/pivoting on your injured knee

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

SP5. Kneeling

None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never ☐ Monthly ☐ Weekly ☐ Daily ☐ Constantly ☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all ☐ Mildly ☐ Moderately ☐ Severely ☐ Totally ☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all ☐ Mildly ☐ Moderately ☐ Severely ☐ Extremely ☐

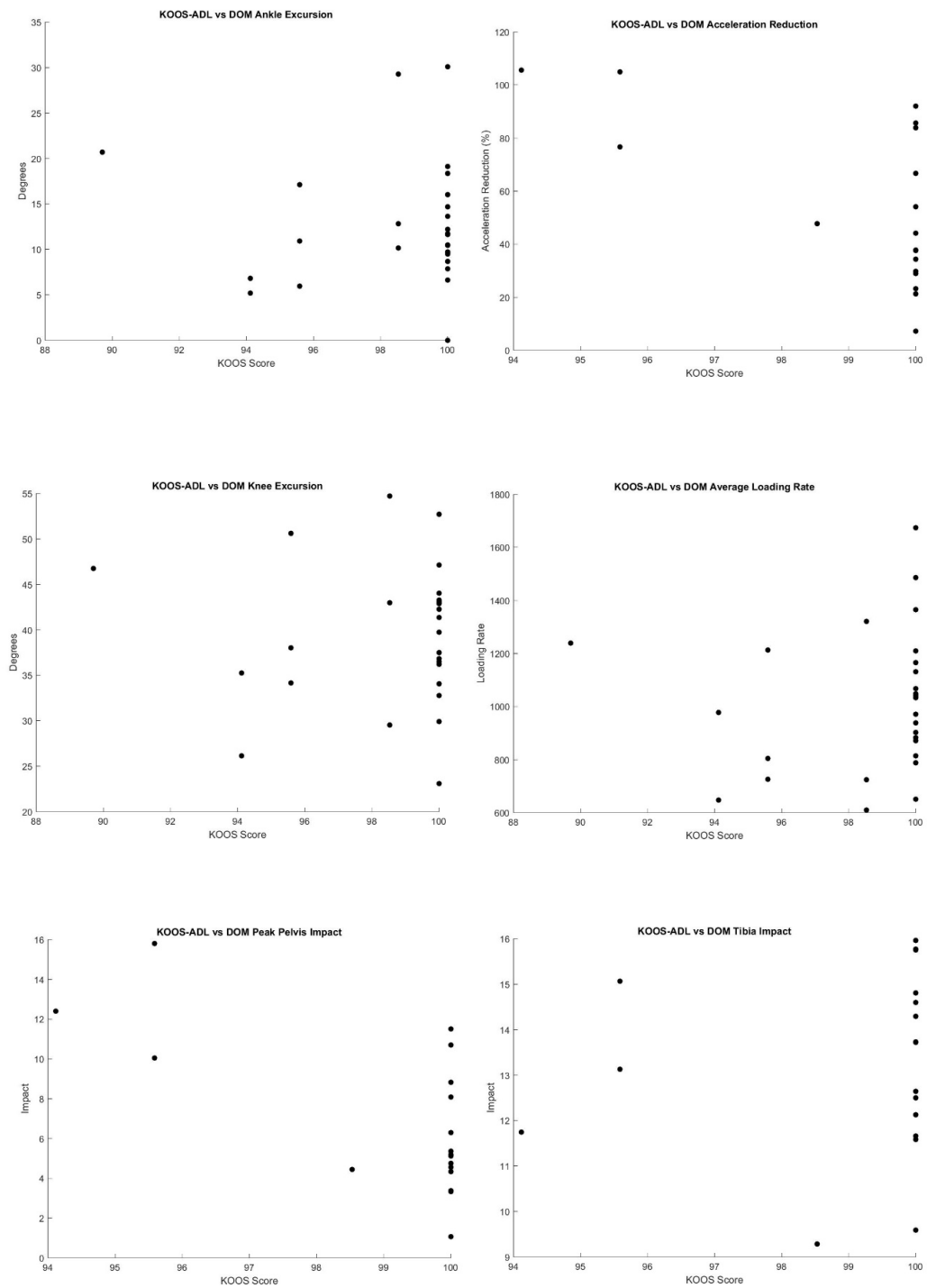
Q4. In general, how much difficulty do you have with your knee?

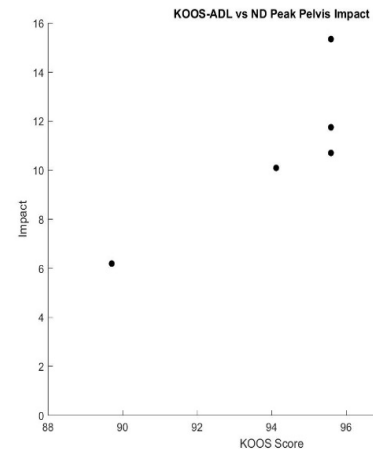
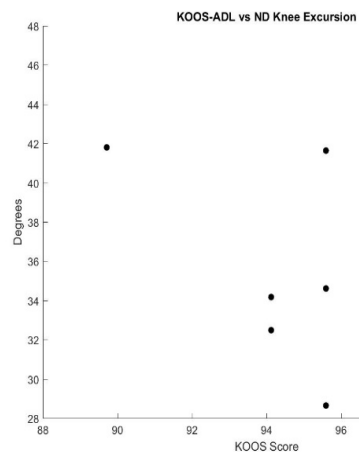
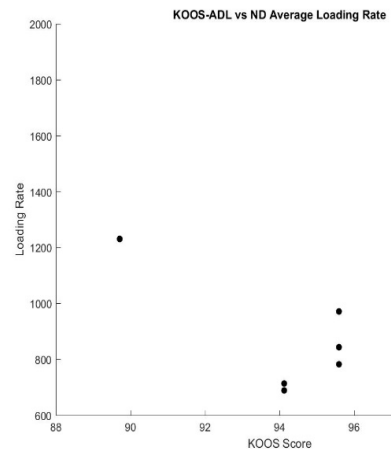
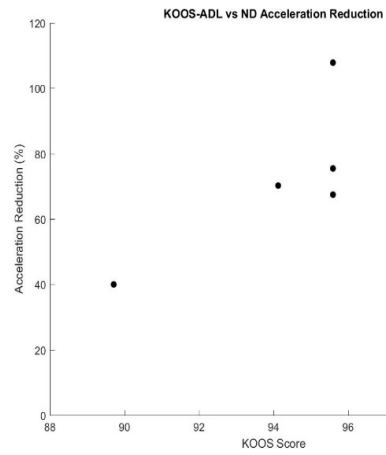
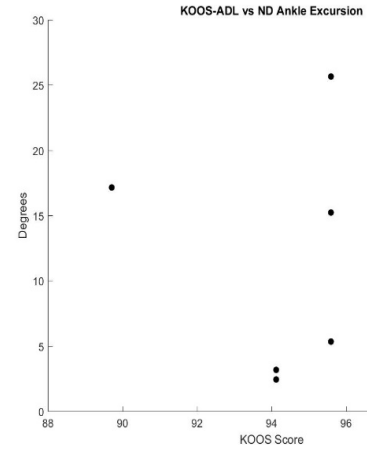
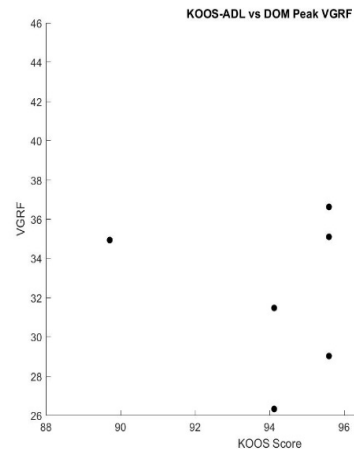
None ☐ Mild ☐ Moderate ☐ Severe ☐ Extreme ☐

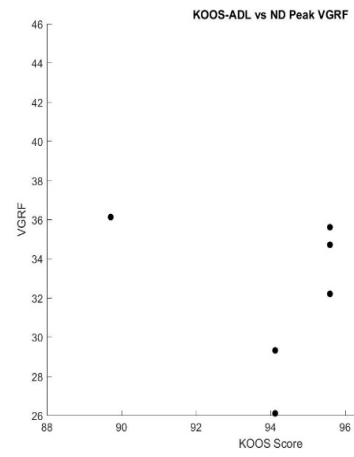
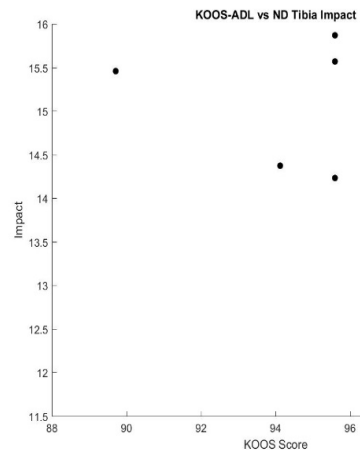
Thank you very much for completing all the questions in this questionnaire.

APPENDIX 4. CLINICAL RELATIONSHIP SCATTERPLOTS

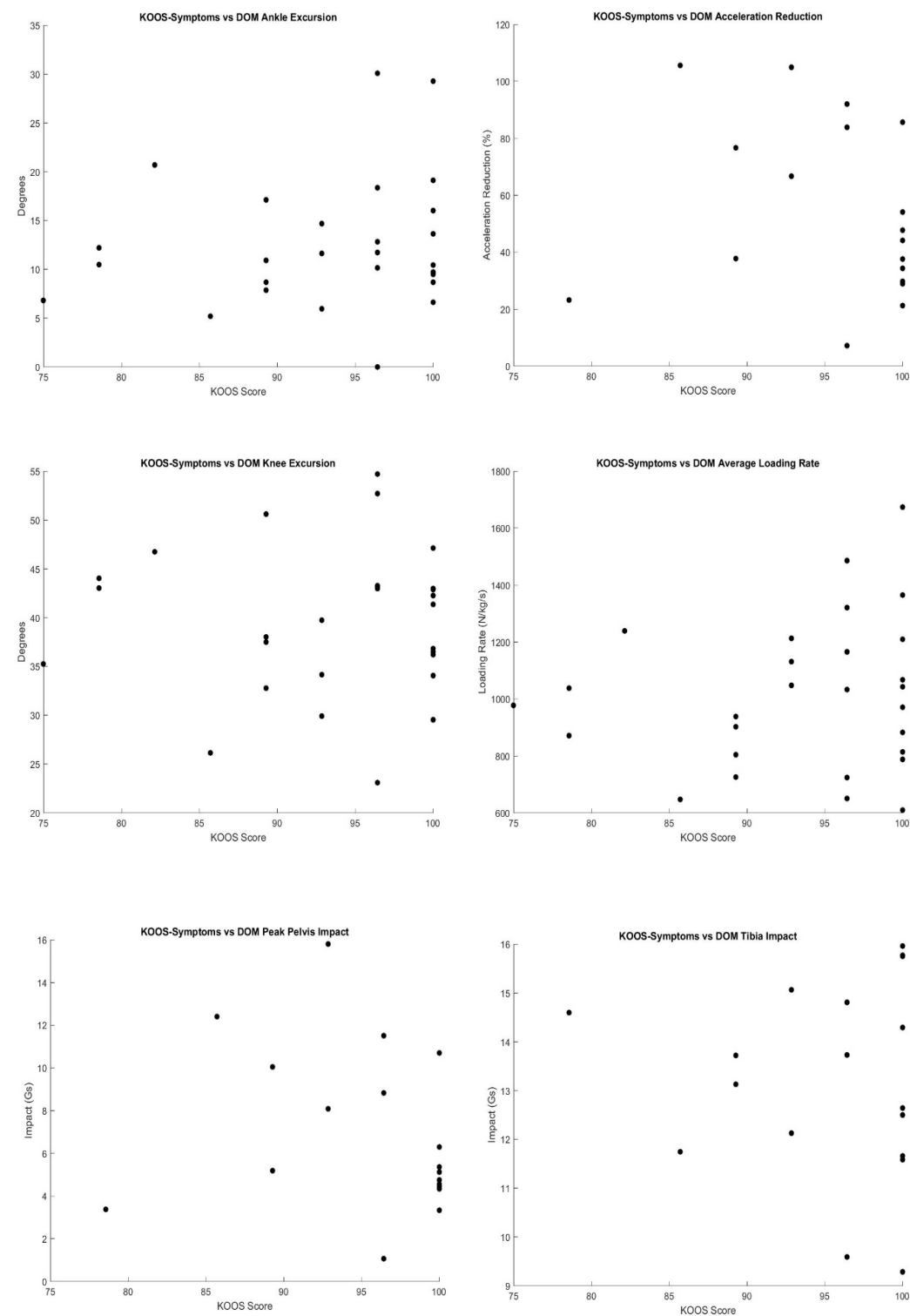
KOOS - ADL

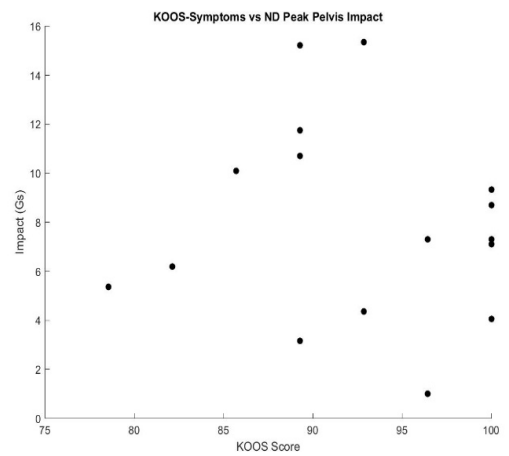
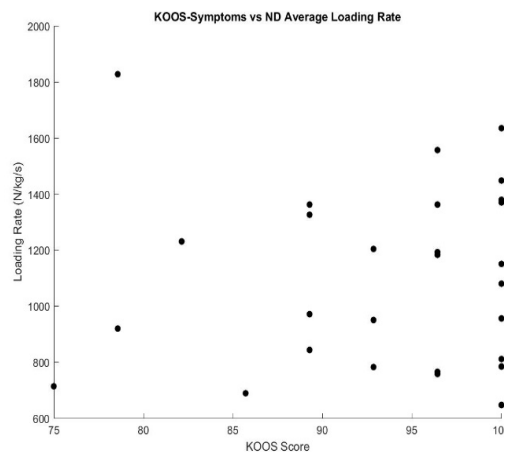
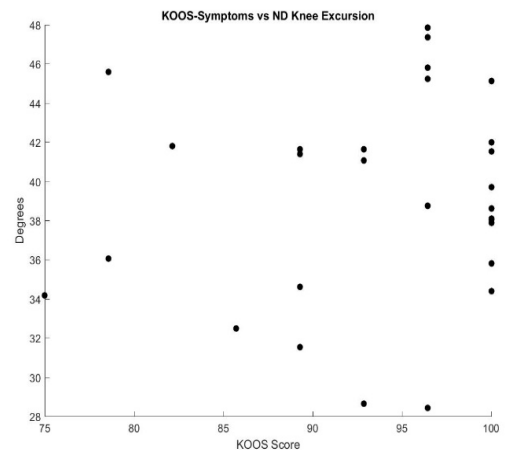
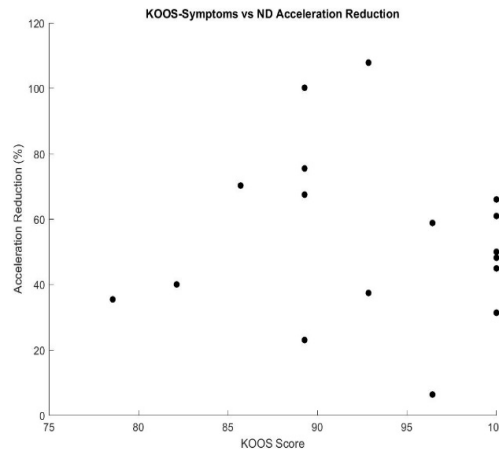
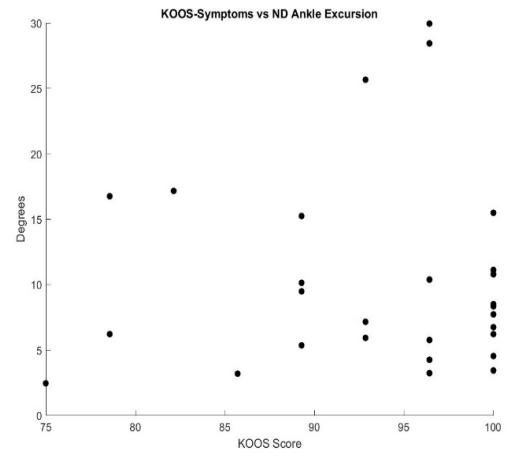
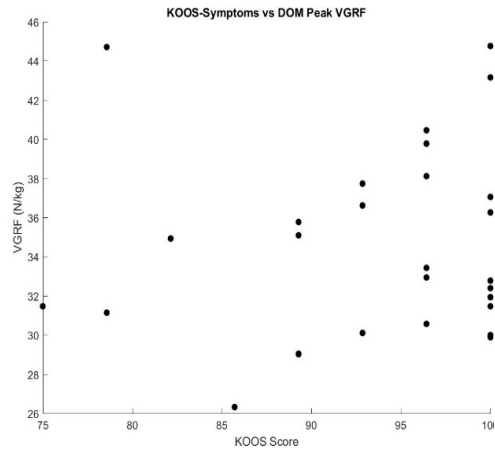


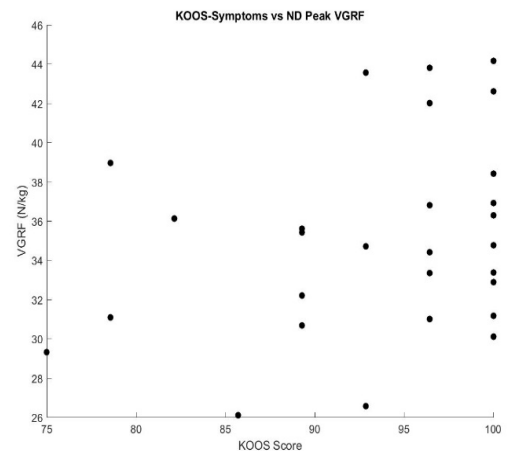
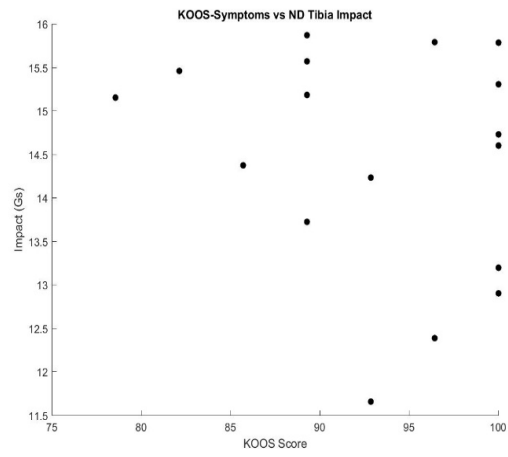




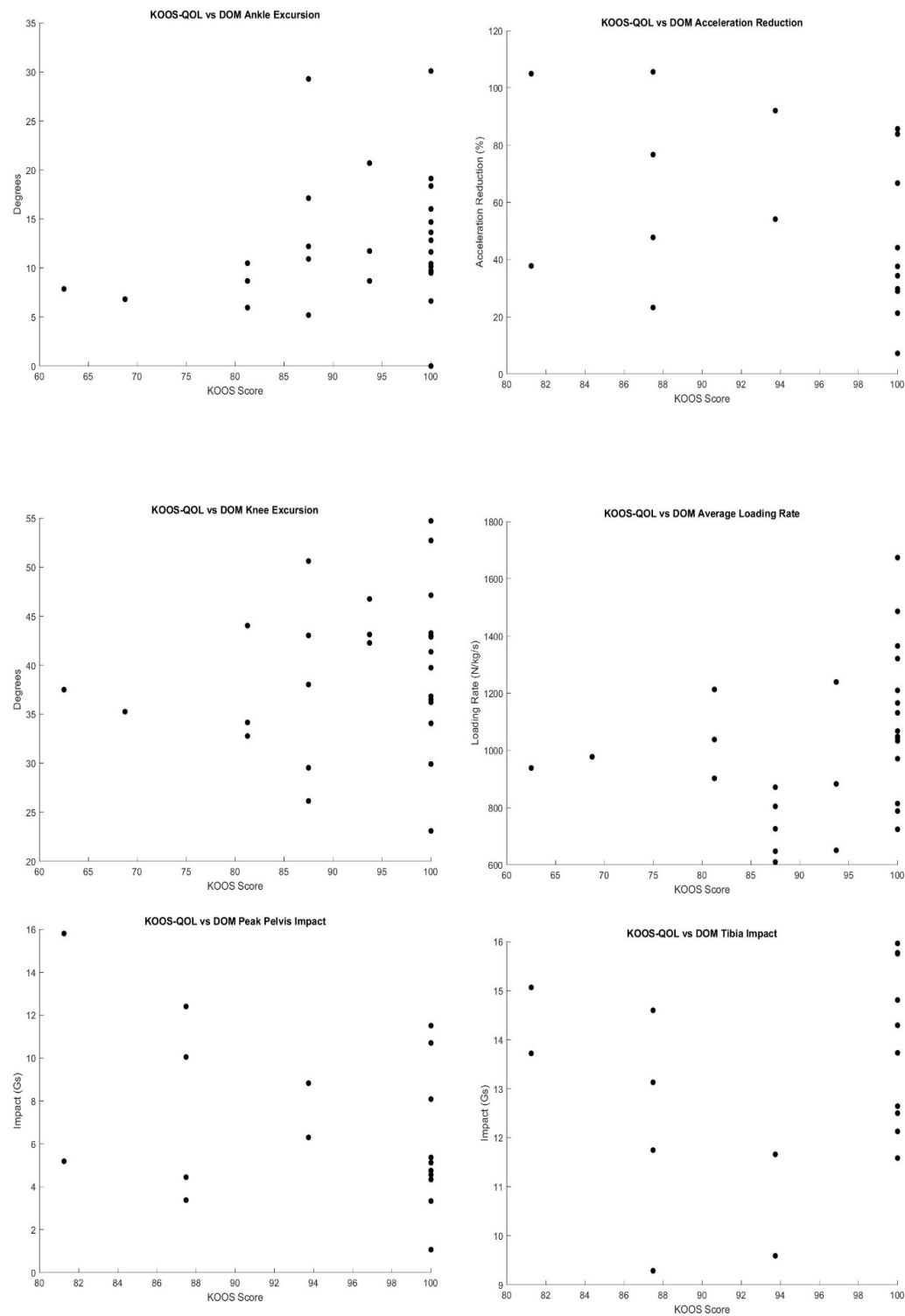
KOOS - Symptoms

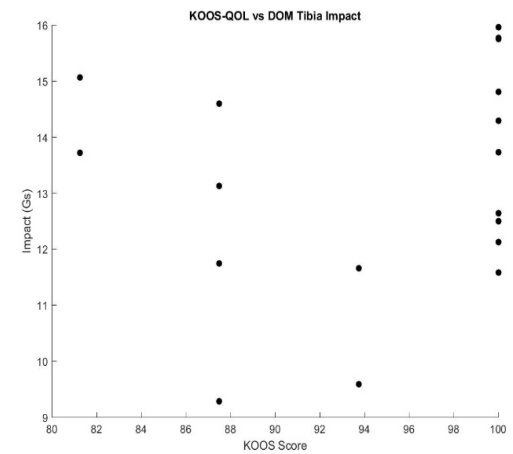
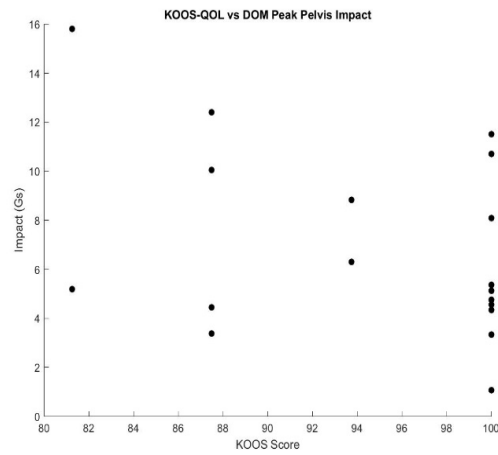
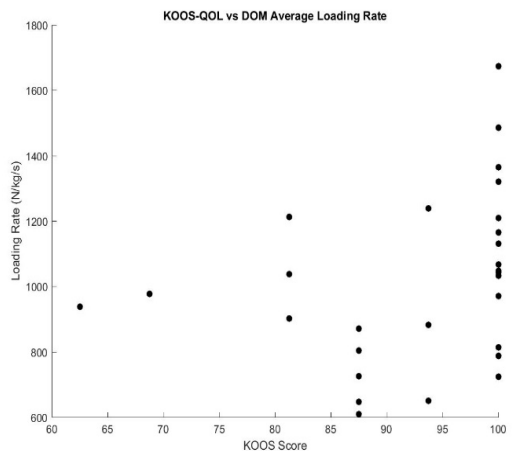
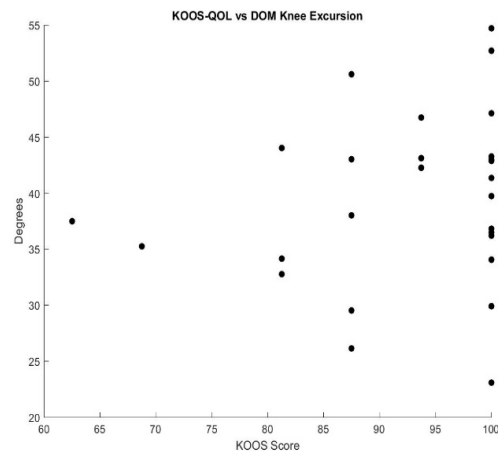
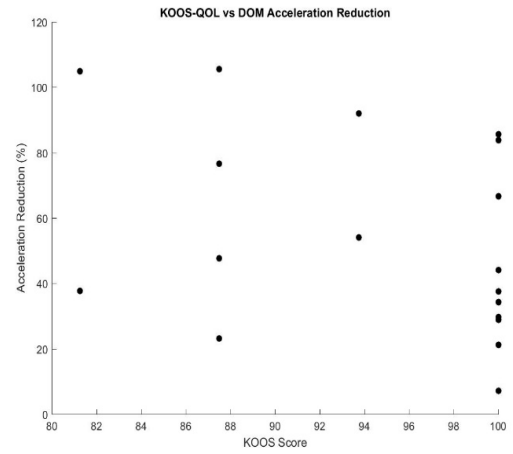
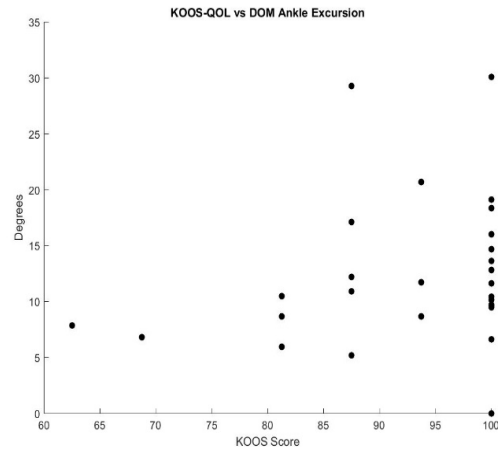


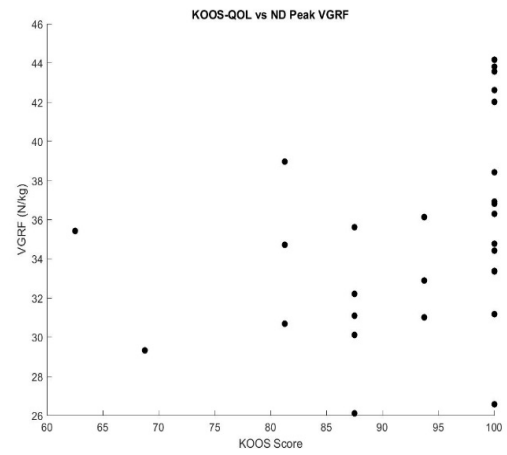
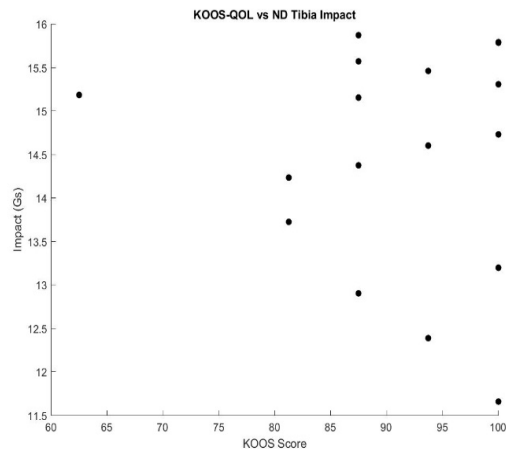




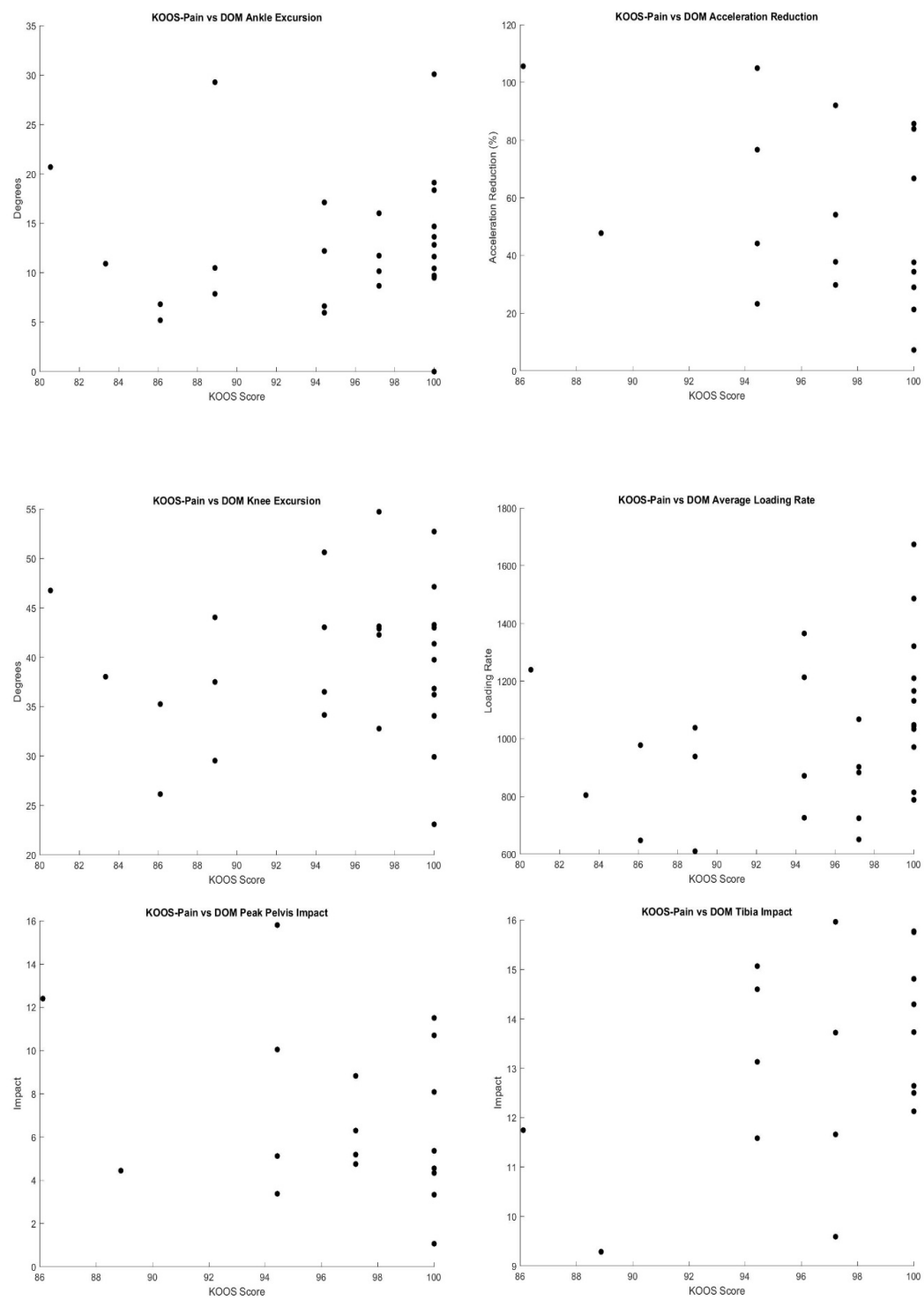
KOOS - QOL

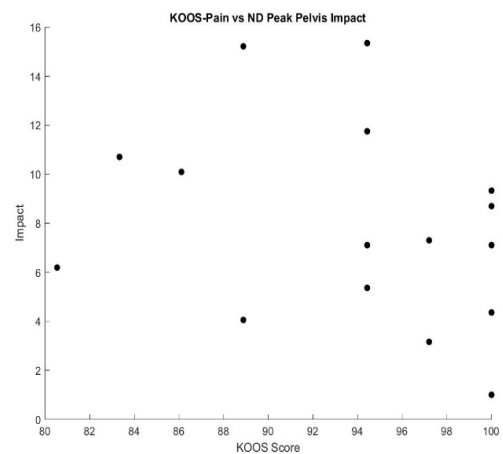
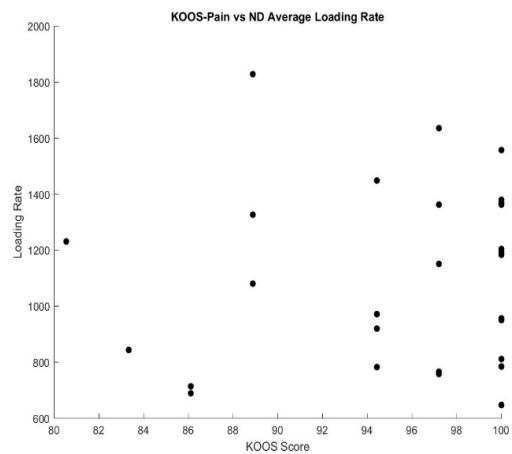
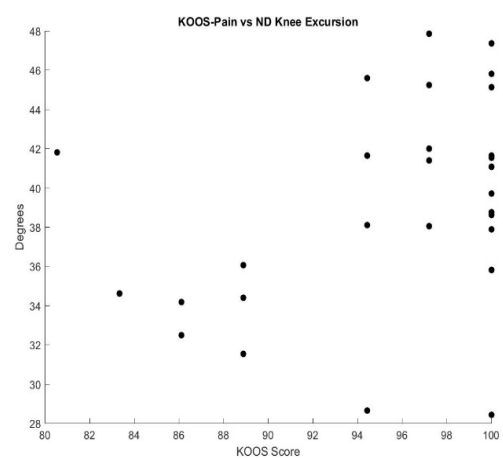
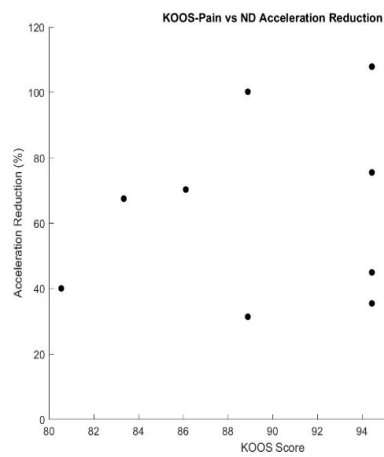
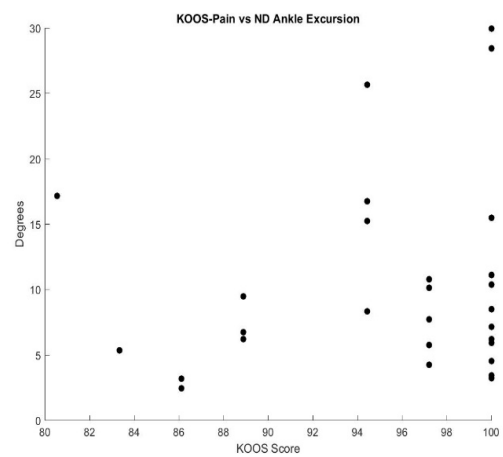
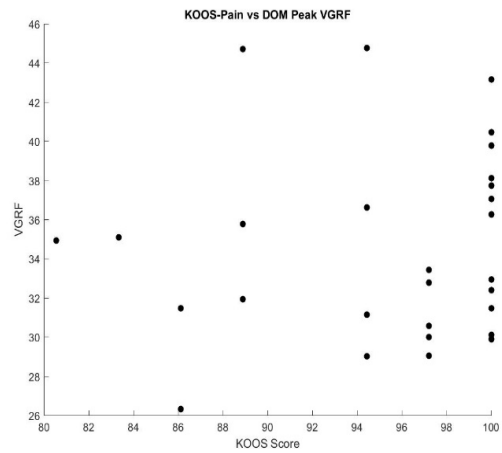


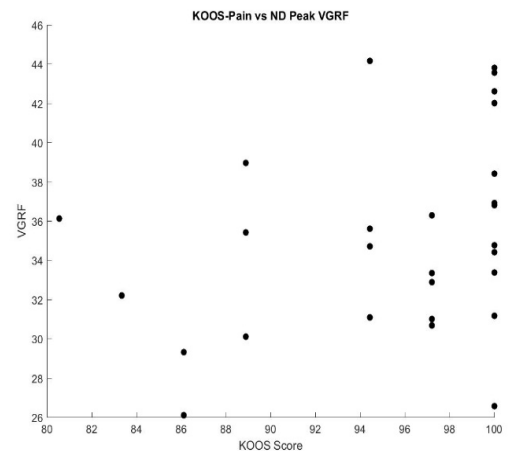
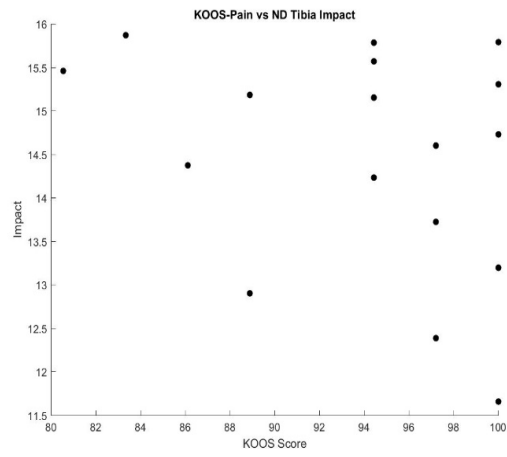




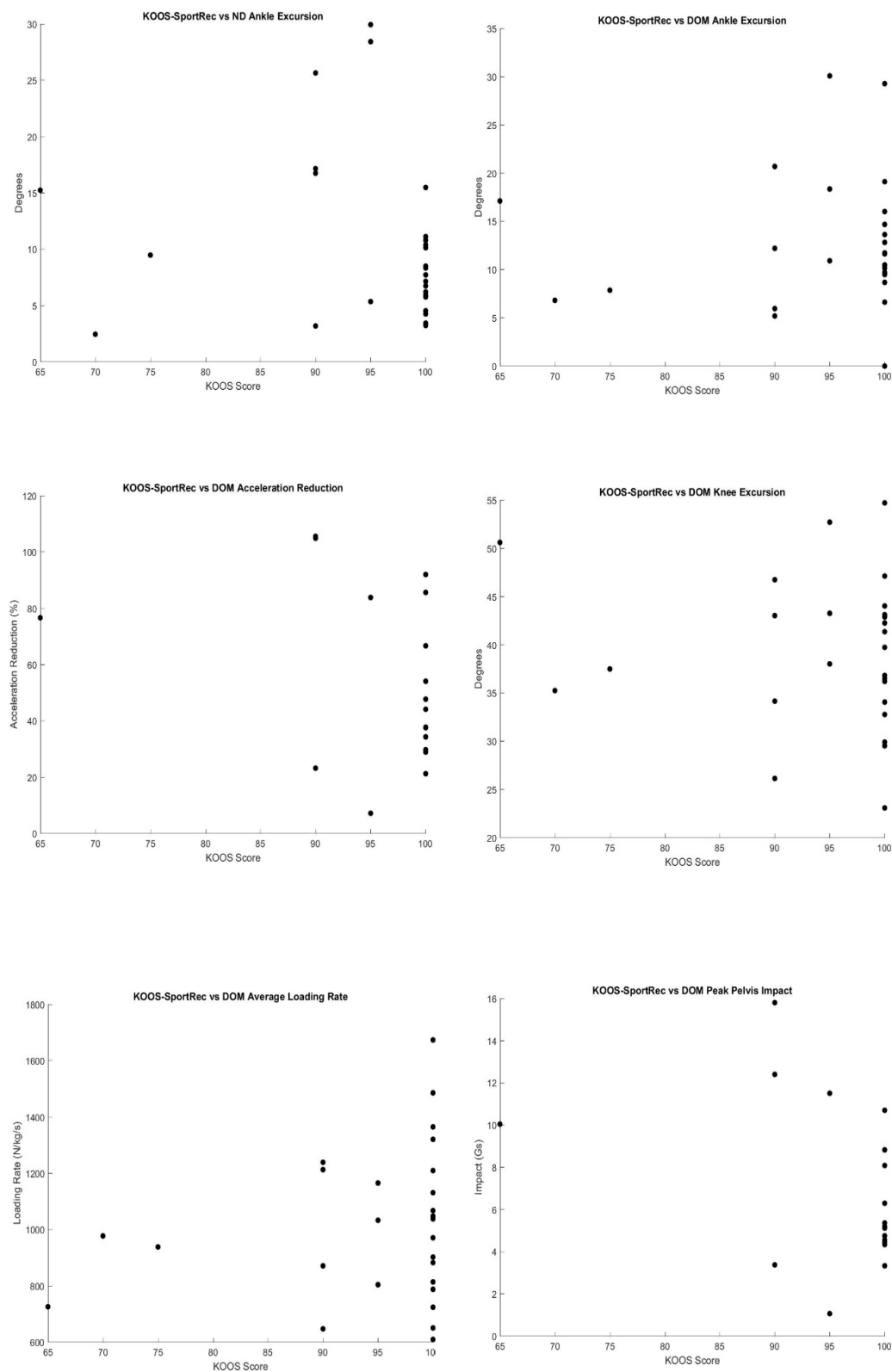
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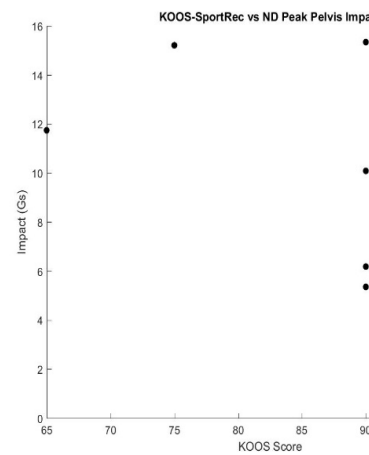
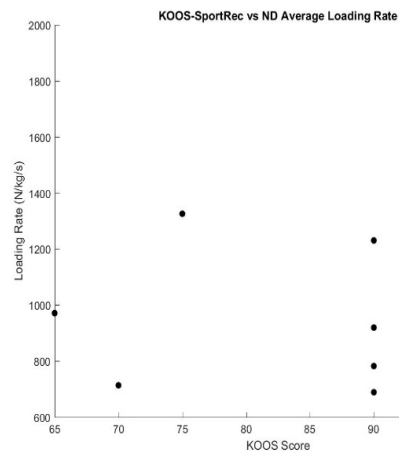
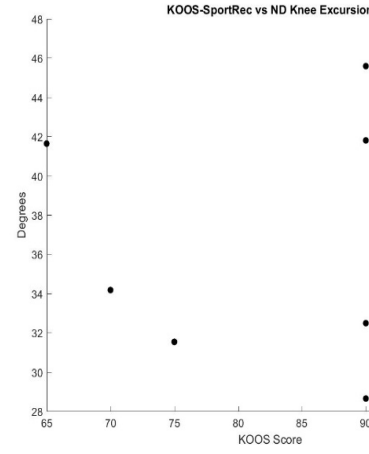
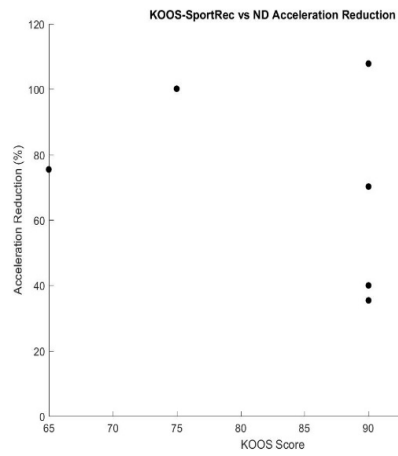
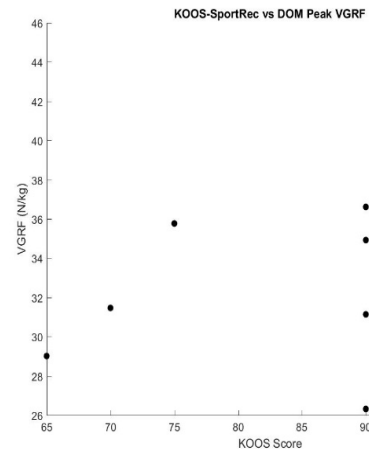
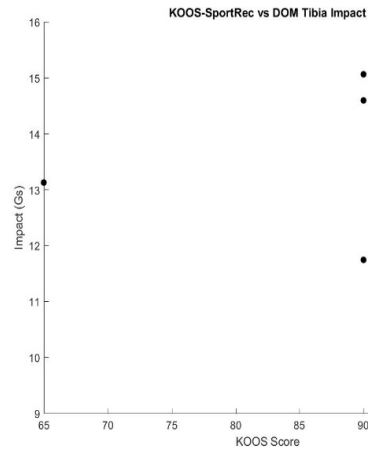


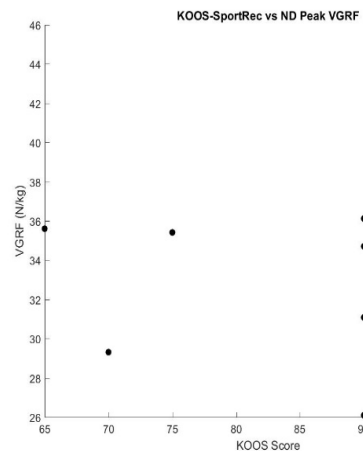
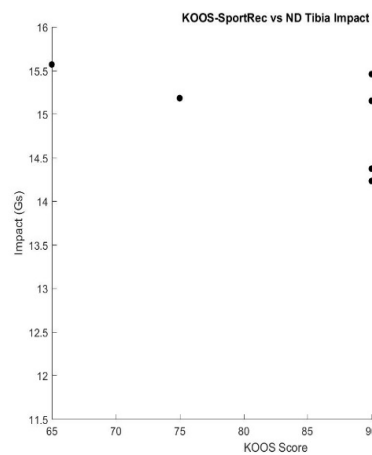




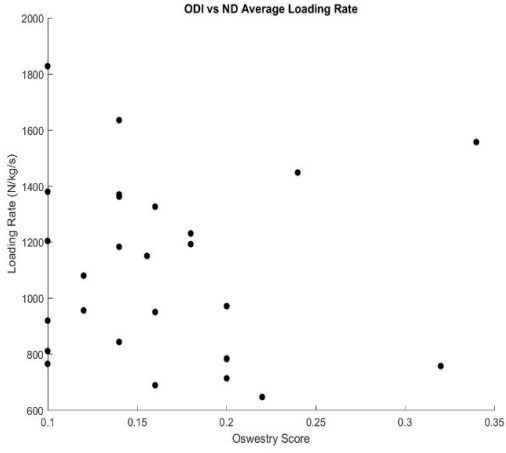
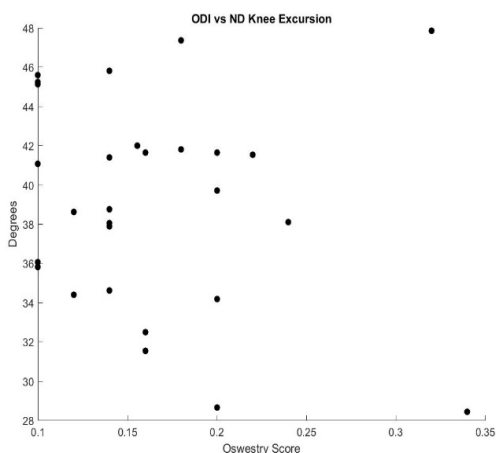
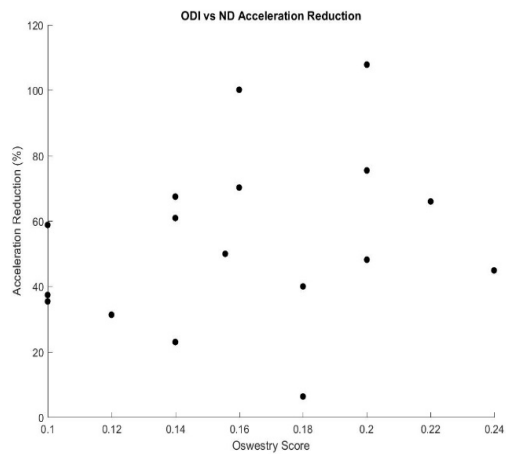
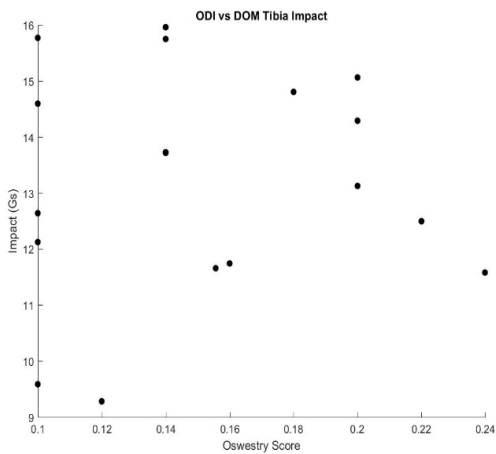
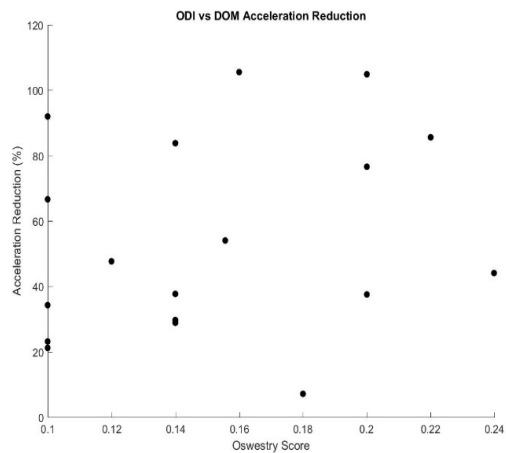
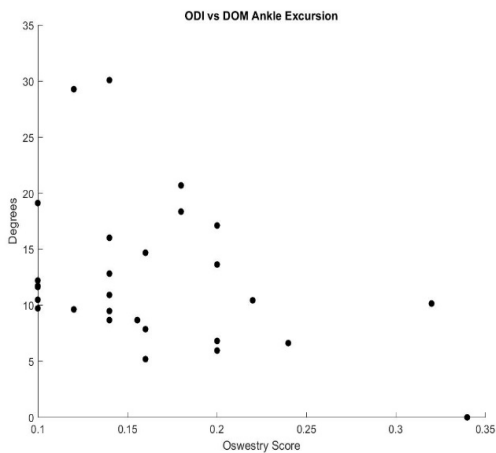
KOOS – Sport and Recreation

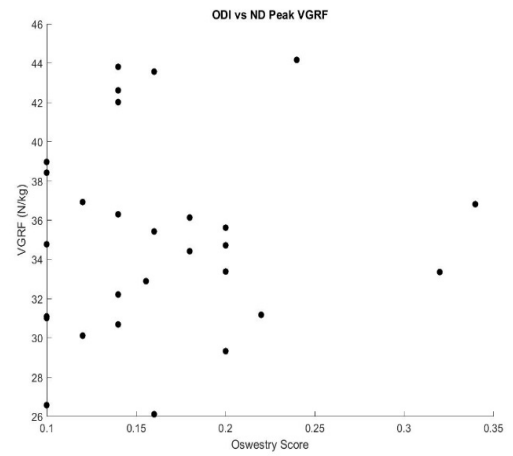
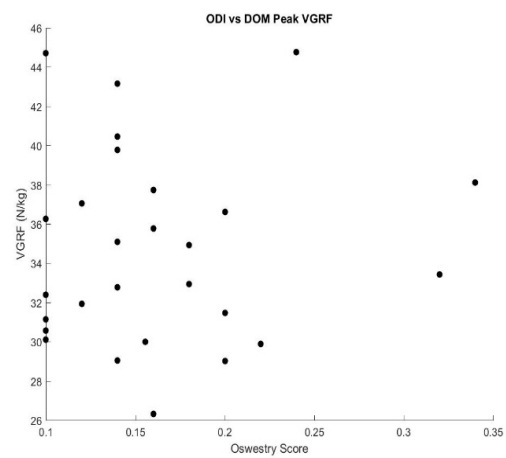
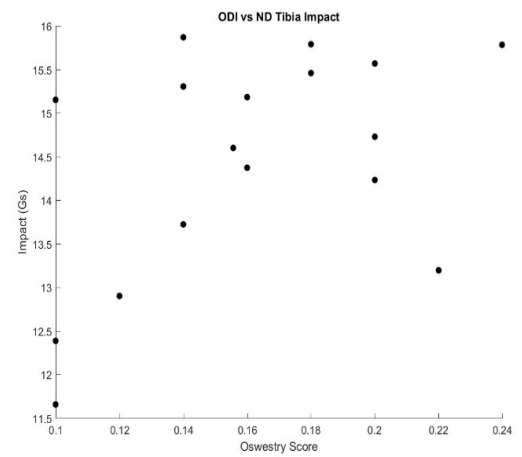
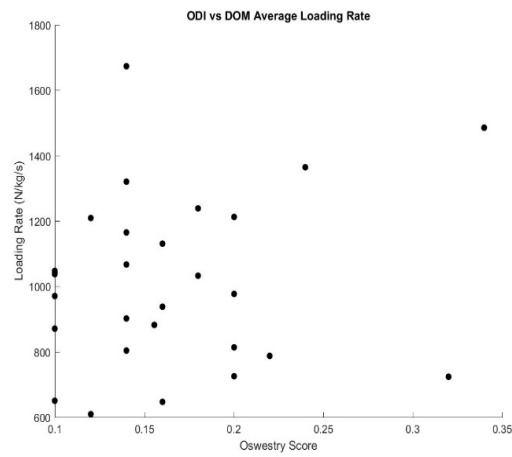
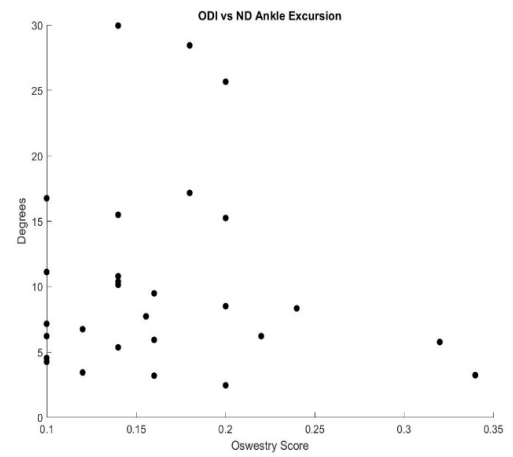
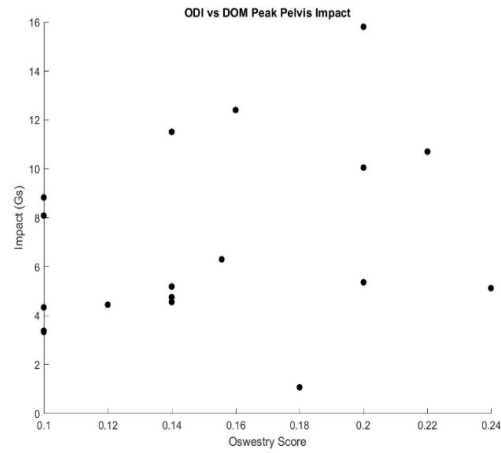


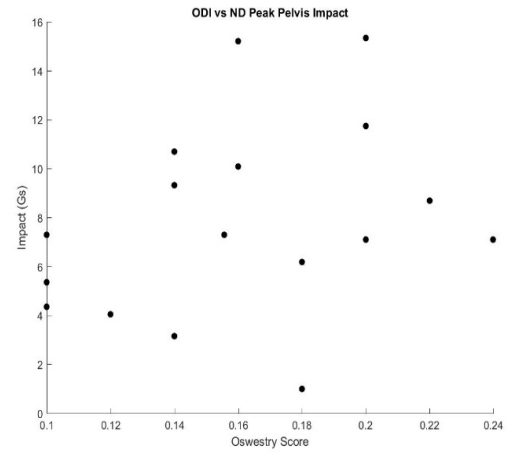
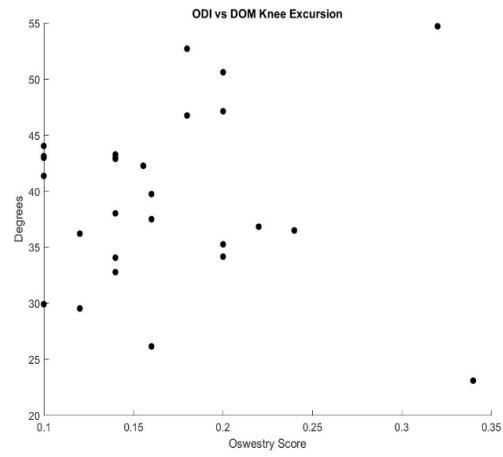




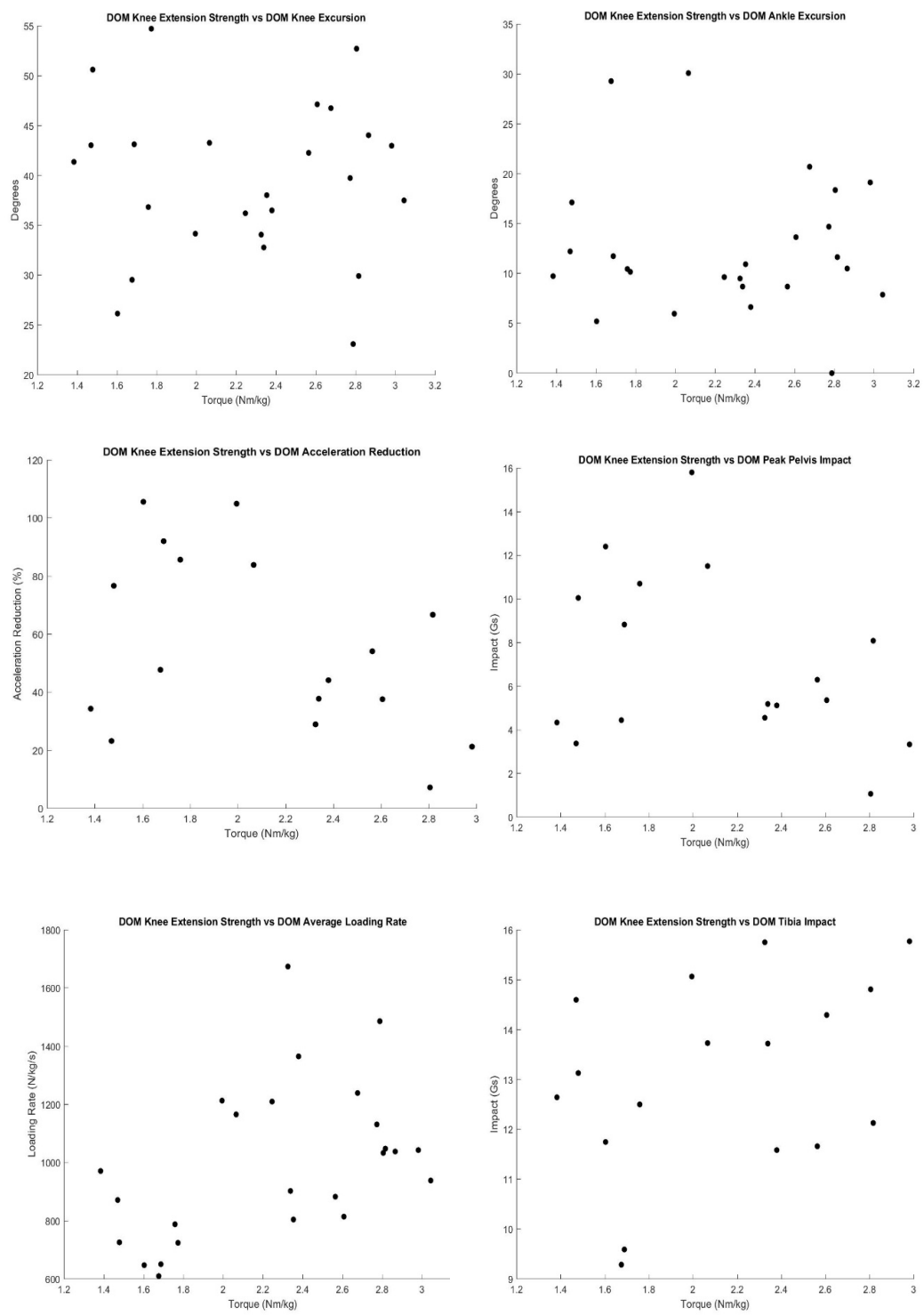
Oswestry Disability Index

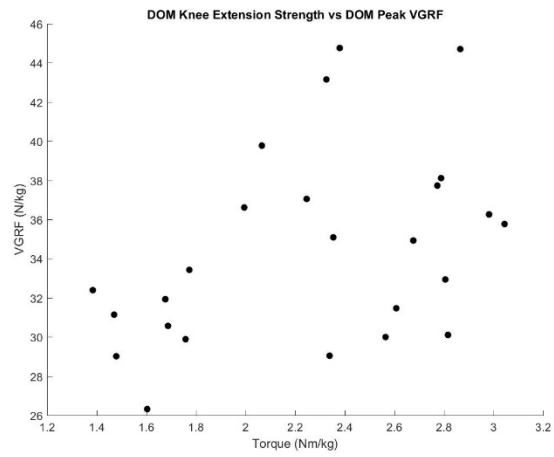




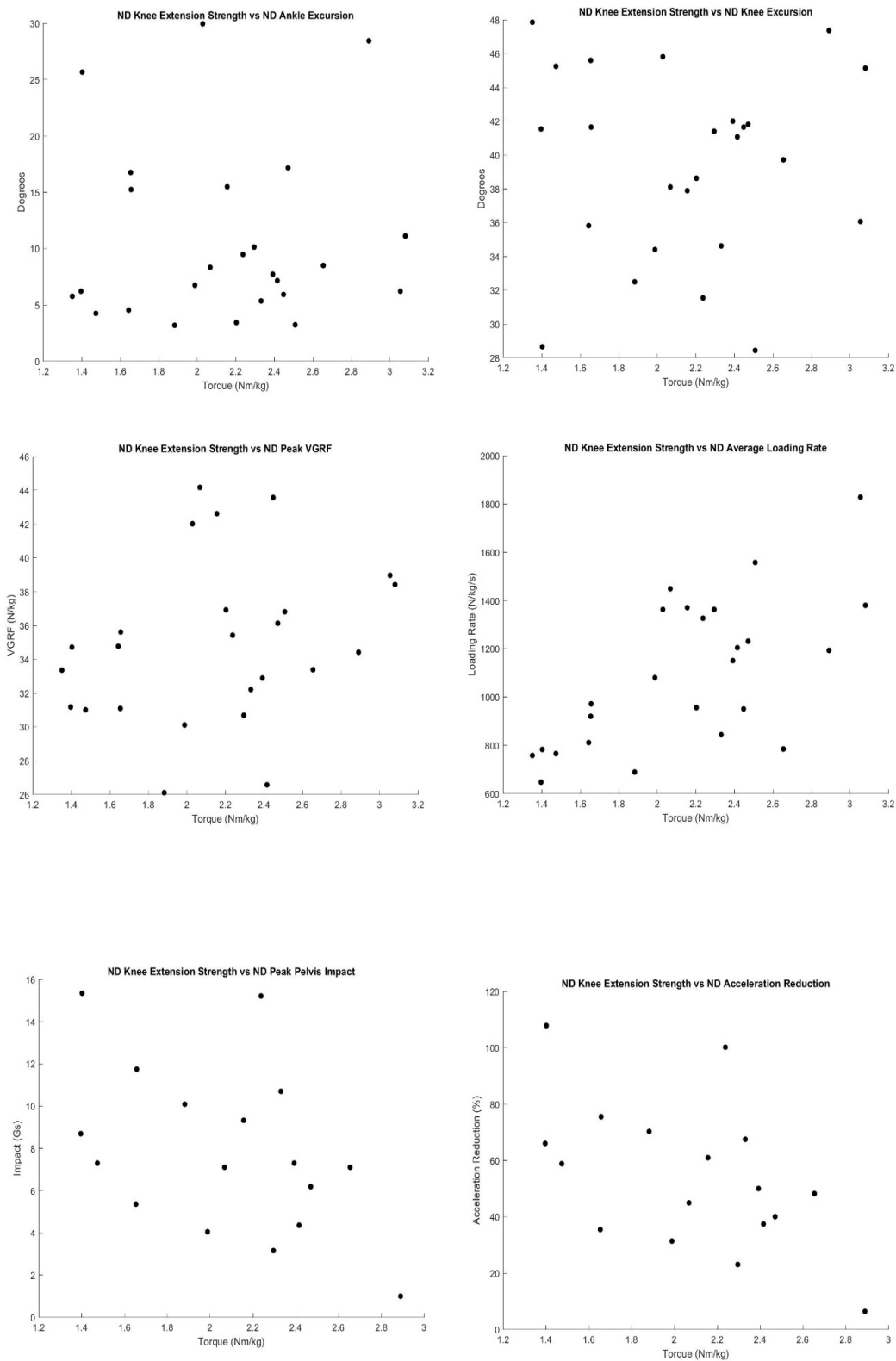


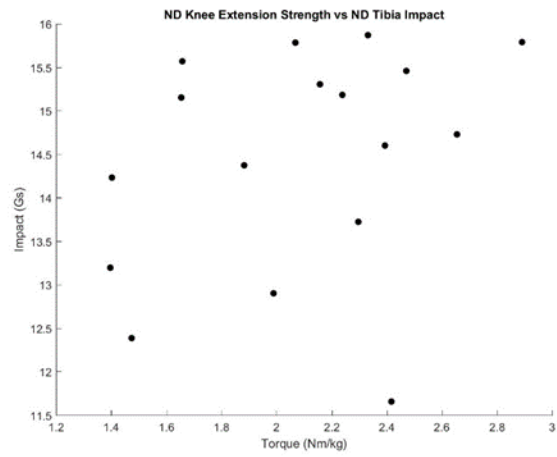
Dominant Knee Extension Strength



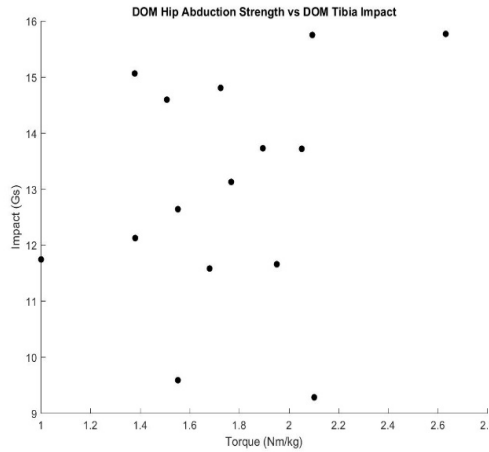
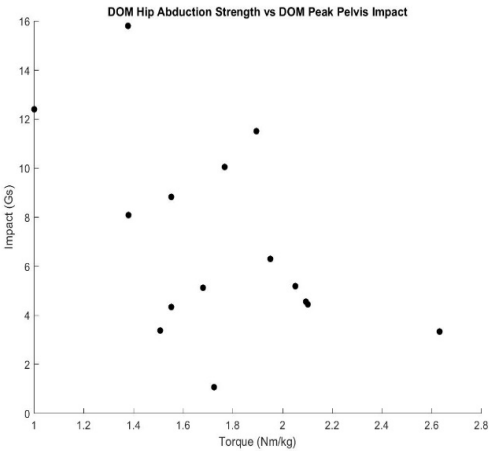
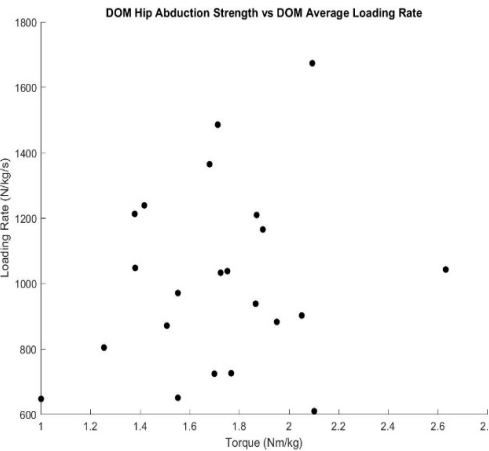
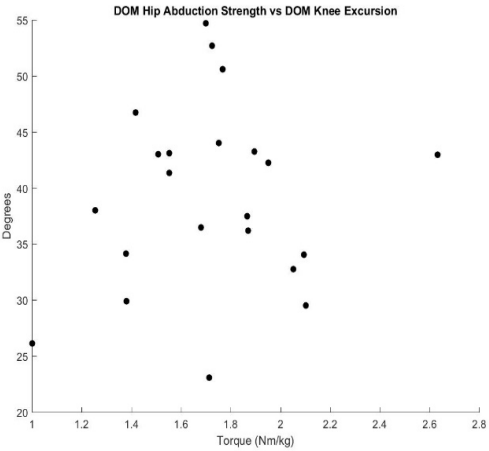
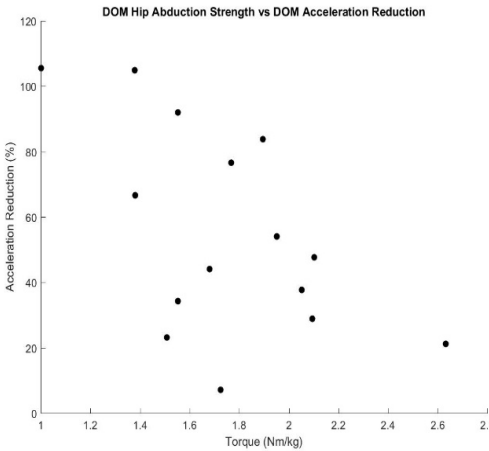
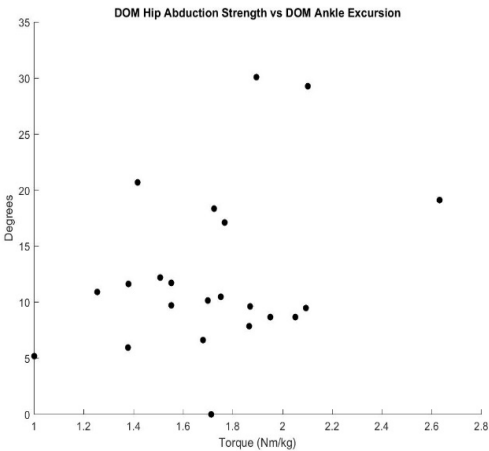


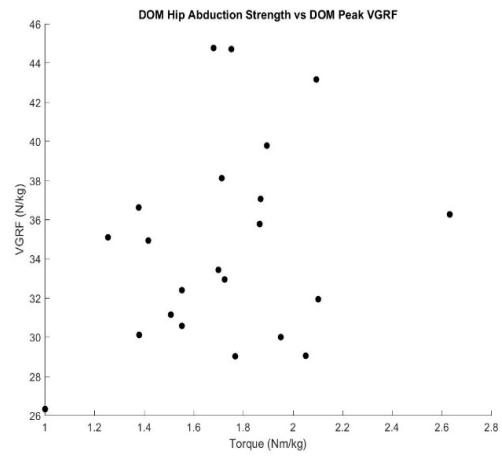
Non-Dominant Knee Extension Strength



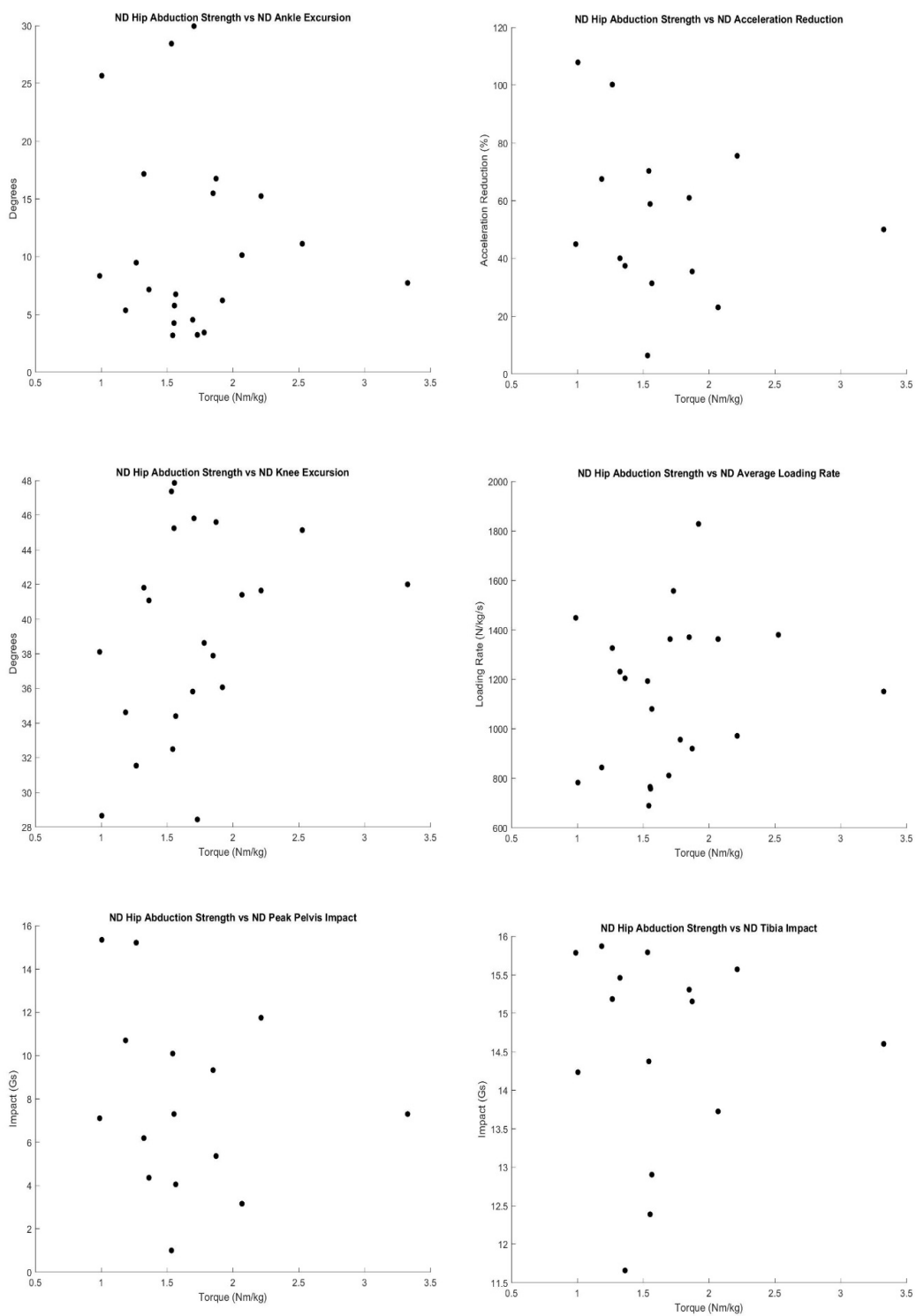


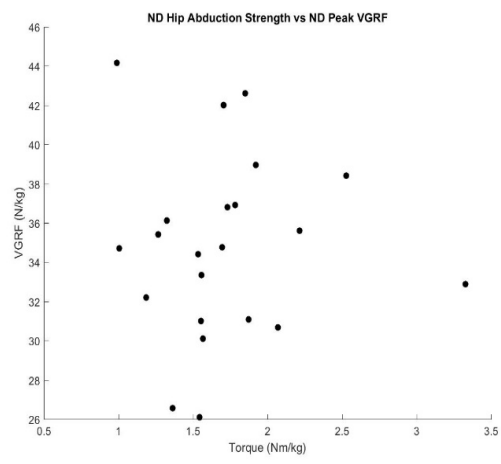
Dominant Hip Extension Strength



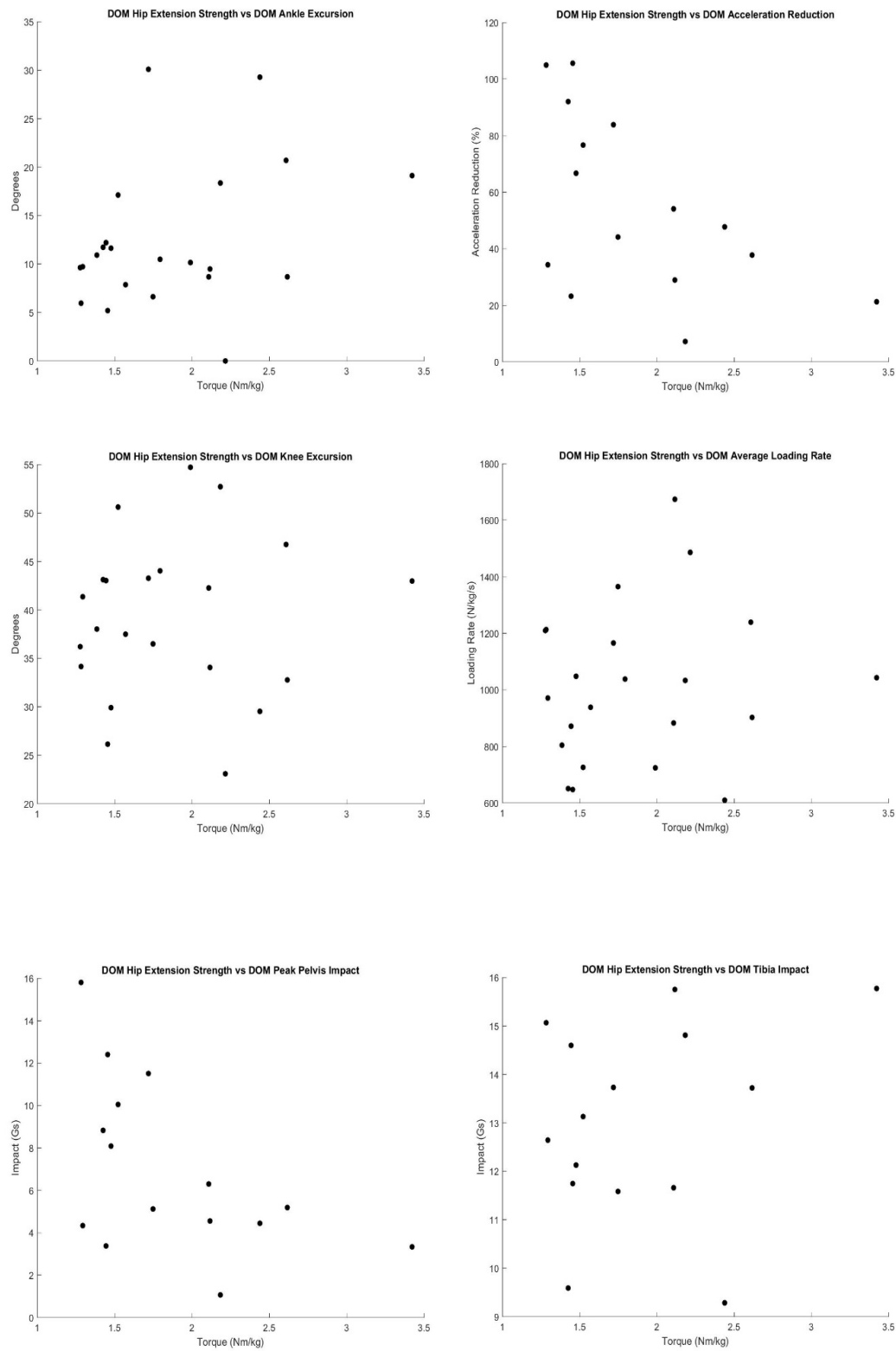


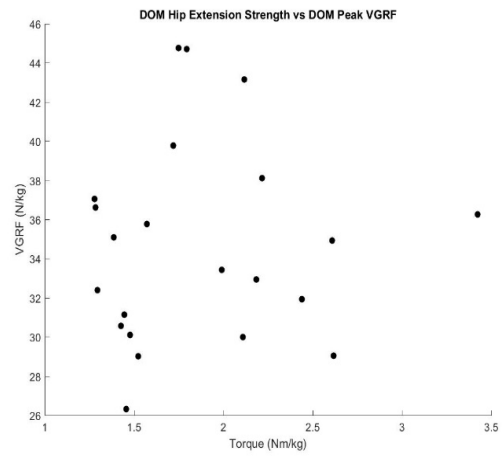
Non-Dominant Hip Abduction Strength



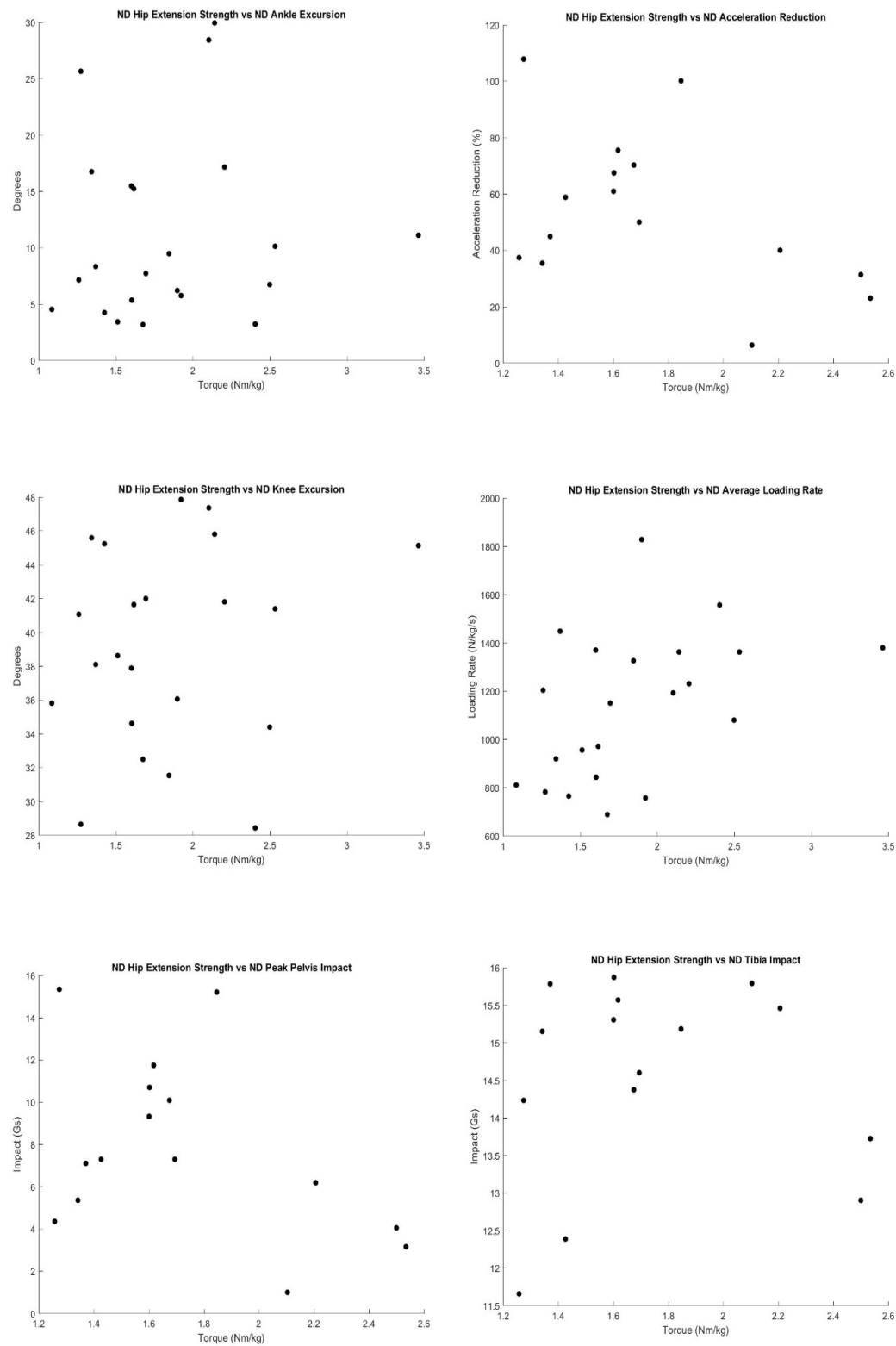


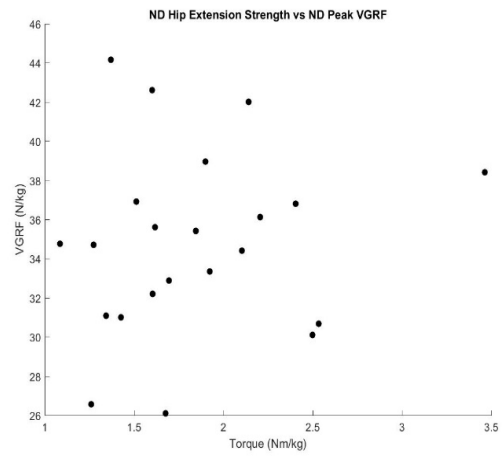
Dominant Hip Extension Strength





Non-Dominant Hip Extension Strength





REFERENCES

1. Andersson, G.B., *Epidemiological features of chronic low-back pain*. The lancet, 1999. **354**(9178): p. 581-585.
2. Andersson, G.B., *Epidemiology of low back pain*. Acta Orthopaedica Scandinavica, 1998. **69**(sup281): p. 28-31.
3. Luo, X., R. Pietrobon, S.X. Sun, G.G. Liu, and L.J.S. Hey, *Estimates and patterns of direct health care expenditures among individuals with back pain in the United States*. 2004. **29**(1): p. 79-86.
4. Rubin, D.I., *Epidemiology and risk factors for spine pain*. Neurologic clinics, 2007. **25**(2): p. 353-371.
5. Shekelle, P.G., M. Markovich, and R. Louie, *An epidemiologic study of episodes of back pain care*. Spine, 1995. **20**(15): p. 1668-1673.
6. d'Hemecourt, P.A., P.G. Gerbino II, and L.J. Micheli, *Back injuries in the young athlete*. Clinics in sports medicine, 2000. **19**(4): p. 663-679.
7. Hutchinson, M.R., *Low back pain in elite rhythmic gymnasts*. Med Sci Sports Exerc, 1999. **31**(11): p. 1686-8.
8. Bono, C.M., *Low-back pain in athletes*. JBJS, 2004. **86**(2): p. 382-396.
9. Nadler, S.F., K.D. Wu, T. Galski, and J.H. Feinberg, *Low back pain in college athletes: a prospective study correlating lower extremity overuse or acquired ligamentous laxity with low back pain*. Spine, 1998. **23**(7): p. 828-833.
10. Semon, R.L. and D. Spengler, *Significance of lumbar spondylolysis in college football players*. Spine, 1981. **6**(2): p. 172-174.
11. Greene, H.S., J. Cholewicki, M.T. Galloway, C.V. Nguyen, and A. Radebold, *A history of low back injury is a risk factor for recurrent back injuries in varsity athletes*. The American journal of sports medicine, 2001. **29**(6): p. 795-800.
12. Clark, L. and Z. Hu, *Low back pain, active component, US Armed Forces, 2010-2014*. MSMR, 2015. **22**(12): p. 8-11.
13. Hollingsworth, D.J., *The prevalence and impact of musculoskeletal injuries during a pre-deployment workup cycle: survey of a Marine Corps special operations company*. Journal of special operations medicine: a peer reviewed journal for SOF medical professionals, 2009. **9**(4): p. 11-15.
14. Abt, J.P., T.C. Sell, M.T. Lovalekar, K.A. Keenan, A.J. Bozich, J.S. Morgan, S.F. Kane, P.J. Benson, and S.M. Lephart, *Injury epidemiology of U.S. Army Special Operations forces*. Mil Med, 2014. **179**(10): p. 1106-12.
15. DeFraités, R., *Absolute and relative morbidity burdens attributable to various illnesses and injuries, US Armed Forces, 2009*. Medical Surveillance Monthly Report, 2010. **17**(4): p. 16-21.
16. Reynolds, K., L. Cosio-Lima, J. Creedon, R. Gregg, and T. Zigmont, *Injury occurrence and risk factors in construction engineers and combat artillery soldiers*. Military medicine, 2002. **167**(12): p. 971-977.
17. Lincoln, A.E., G.S. Smith, P.J. Amoroso, and N.S. Bell, *The natural history and risk factors of musculoskeletal conditions resulting in disability among US Army personnel*. Work, 2002. **18**(2): p. 99-113.

18. Nadler, S.F., G.A. Malanga, J.H. Feinberg, M. Prybicien, T.P. Stitik, and M. DePrince, *Relationship between hip muscle imbalance and occurrence of low back pain in collegiate athletes: a prospective study*. American journal of physical medicine & rehabilitation, 2001. **80**(8): p. 572-577.
19. Nadler, S.F., G.A. Malanga, L.A. Bartoli, J.H. Feinberg, M. Prybicien, and M. DePrince, *Hip muscle imbalance and low back pain in athletes: influence of core strengthening*. Medicine & Science in Sports & Exercise, 2002. **34**(1): p. 9-16.
20. Nourbakhsh, M.R. and A.M. Arab, *Relationship between mechanical factors and incidence of low back pain*. Journal of Orthopaedic & Sports Physical Therapy, 2002. **32**(9): p. 447-460.
21. Purcell, L. and L. Micheli, *Low back pain in young athletes*. Sports Health, 2009. **1**(3): p. 212-222.
22. Arendt-Nielsen, L., T. Graven-Nielsen, H. Sværre, and P. Svensson, *The influence of low back pain on muscle activity and coordination during gait: a clinical and experimental study*. Pain, 1996. **64**(2): p. 231-240.
23. Seay, J.F., R.E. Van Emmerik, and J. Hamill, *Low back pain status affects pelvis-trunk coordination and variability during walking and running*. Clin Biomech (Bristol, Avon), 2011. **26**(6): p. 572-8.
24. Cai, C. and P.W. Kong, *Low back and lower-limb muscle performance in male and female recreational runners with chronic low back pain*. Journal of Orthopaedic & Sports Physical Therapy, 2015. **45**(6): p. 436-443.
25. Nadler, S.F., G.A. Malanga, M. DePrince, T.P. Stitik, and J.H. Feinberg, *The relationship between lower extremity injury, low back pain, and hip muscle strength in male and female collegiate athletes*. Clinical Journal of Sport Medicine, 2000. **10**(2): p. 89-97.
26. Wolfe, F., *Determinants of WOMAC function, pain and stiffness scores: evidence for the role of low back pain, symptom counts, fatigue and depression in osteoarthritis, rheumatoid arthritis and fibromyalgia*. Rheumatology (Oxford, England), 1999. **38**(4): p. 355-361.
27. Wainner, R.S., J.M. Whitman, J.A. Cleland, and T.W. Flynn, *Regional interdependence: a musculoskeletal examination model whose time has come*. 2007, JOSPT, Inc. JOSPT, 1033 North Fairfax Street, Suite 304, Alexandria, VA 22134-1540.
28. Stupar, M., P. Côté, M.R. French, and G.A. Hawker, *The association between low back pain and osteoarthritis of the hip and knee: a population-based cohort study*. Journal of manipulative and physiological therapeutics, 2010. **33**(5): p. 349-354.
29. Deusinger, R.H., *Biomechanics in clinical practice*. Physical therapy, 1984. **64**(12): p. 1860-1868.
30. Granan, L.-P., V. Baste, L. Engebretsen, and M.C. Inacio, *Associations between inadequate knee function detected by KOOS and prospective graft failure in an anterior cruciate ligament-reconstructed knee*. Knee Surgery, Sports Traumatology, Arthroscopy, 2015. **23**(4): p. 1135-1140.
31. Roos, E.M. and L.S. Lohmander, *The Knee injury and Osteoarthritis Outcome Score (KOOS): from joint injury to osteoarthritis*. Health quality of life outcomes, 2003. **1**(1): p. 64.

32. Long, M.J., E. Papi, L.D. Duffell, and A.H. McGregor, *Predicting knee osteoarthritis risk in injured populations*. Clinical biomechanics, 2017. **47**: p. 87-95.
33. Kendall, K.D., C.A. Emery, J.P. Wiley, and R. Ferber, *The effect of the addition of hip strengthening exercises to a lumbopelvic exercise programme for the treatment of non-specific low back pain: A randomized controlled trial*. J Sci Med Sport, 2015. **18**(6): p. 626-31.
34. Hides, J.A., T. Oostenbroek, M.M. Franettovich Smith, and M.D. Mendis, *The effect of low back pain on trunk muscle size/function and hip strength in elite football (soccer) players*. Journal of sports sciences, 2016. **34**(24): p. 2303-2311.
35. Grindem, H., L. Snyder-Mackler, H. Moksnes, L. Engebretsen, and M.A. Risberg, *Simple decision rules can reduce reinjury risk by 84% after ACL reconstruction: the Delaware-Oslo ACL cohort study*. Br J Sports Med, 2016.
36. Palmieri-Smith, R.M. and A.C. Thomas, *A neuromuscular mechanism of posttraumatic osteoarthritis associated with ACL injury*. Exerc Sport Sci Rev, 2009. **37**(3): p. 147-53.
37. Powers, C.M., *The influence of abnormal hip mechanics on knee injury: a biomechanical perspective*. journal of orthopaedic & sports physical therapy, 2010. **40**(2): p. 42-51.
38. Nicholas, J.A., R.B. Grossman, and E.B. Hershman, *The importance of a simplified classification of motion in sports in relation to performance*. Orthopedic Clinics of North America, 1977. **8**(3): p. 499-532.
39. Hamill, J., M. Moses, and J. Seay, *Lower extremity joint stiffness in runners with low back pain*. Res Sports Med, 2009. **17**(4): p. 260-73.
40. Muller, R., T. Ertelt, and R. Blickhan, *Low back pain affects trunk as well as lower limb movements during walking and running*. J Biomech, 2015. **48**(6): p. 1009-14.
41. Müller, R., T. Ertelt, and R. Blickhan, *Low back pain affects trunk as well as lower limb movements during walking and running*. Journal of biomechanics, 2015. **48**(6): p. 1009-1014.
42. Keefe, F.J. and R.W. Hill, *An objective approach to quantifying pain behavior and gait patterns in low back pain patients*. Pain, 1985. **21**(2): p. 153-161.
43. Khodadadeh, S. and S.M. Eisenstein, *Gait analysis of patients with low back pain before and after surgery*. Spine, 1993. **18**(11): p. 1451-1455.
44. Voloshin, A. and J. Wosk, *An in vivo study of low back pain and shock absorption in the human locomotor system*. Journal of biomechanics, 1982. **15**(1): p. 21-27.
45. Dejnabadi, H., B.M. Jolles, and K. Aminian, *A new approach for quantitative analysis of inter-joint coordination during gait*. IEEE Transactions on Biomedical Engineering, 2008. **55**(2): p. 755-764.
46. Hutin, E., D. Pradon, F. Barbier, J.-M. Gracies, B. Bussel, and N. Roche, *Lower limb coordination patterns in hemiparetic gait: factors of knee flexion impairment*. Clinical biomechanics, 2011. **26**(3): p. 304-311.
47. Heiderscheit, B.C., J. Hamill, and R.E. van Emmerik, *Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain*. Journal of applied biomechanics, 2002. **18**(2): p. 110-121.

48. Miller, R.H., S.A. Meardon, T.R. Derrick, and J.C. Gillette, *Continuous relative phase variability during an exhaustive run in runners with a history of iliotibial band syndrome*. Journal of Applied Biomechanics, 2008. **24**(3): p. 262-270.
49. Chiu, S.-L., T.-W. Lu, and L.-S. Chou, *Altered inter-joint coordination during walking in patients with total hip arthroplasty*. Gait & posture, 2010. **32**(4): p. 656-660.
50. Worster, K., J. Valvano, and J.J. Carollo, *Sagittal plane coordination dynamics of typically developing gait*. Clinical Biomechanics, 2015. **30**(4): p. 366-372.
51. Lamothe, C., P.J. Beek, and O.G. Meijer, *Pelvis–thorax coordination in the transverse plane during gait*. Gait & posture, 2002. **16**(2): p. 101-114.
52. van Emmerik, R.E.A. and R. Wagenaar, *Effects of walking velocity on relative phase dynamics in the trunk in human walking*. Journal of biomechanics, 1996. **29**(9): p. 1175-1184.
53. Lund, J.P., R. Donga, C.G. Widmer, and C.S. Stohler, *The pain-adaptation model: a discussion of the relationship between chronic musculoskeletal pain and motor activity*. Canadian journal of physiology and pharmacology, 1991. **69**(5): p. 683-694.
54. Vogt, L., K. Pfeifer, and W. Banzer, *Neuromuscular control of walking with chronic low-back pain*. Manual therapy, 2003. **8**(1): p. 21-28.
55. Seay, J.F., R.E. Van Emmerik, and J. Hamill, *Low back pain status affects pelvis-trunk coordination and variability during walking and running*. Clin Biomech 2011. **26**(6): p. 572-8.
56. Seay, J.F., R.E. Van Emmerik, and J. Hamill, *Influence of low back pain status on pelvis-trunk coordination during walking and running*. Spine 2011. **36**(16): p. E1070-9.
57. Sung, P.S., J.T. Zippel, and P. Danial, *Gender differences in asymmetrical limb support patterns between subjects with and without recurrent low back pain*. Hum Mov Sci, 2017. **52**: p. 36-44.
58. Sung, P.S., *A kinematic analysis for shoulder and pelvis coordination during axial trunk rotation in subjects with and without recurrent low back pain*. Gait & posture, 2014. **40**(4): p. 493-498.
59. Sung, P.S., P. Danial, and D.C. Lee, *Comparison of the different kinematic patterns during lateral bending between subjects with and without recurrent low back pain*. Clinical Biomechanics, 2016. **38**: p. 50-55.
60. Dufek, J.S. and B.T. Bates, *Biomechanical factors associated with injury during landing in jump sports*. Sports medicine, 1991. **12**(5): p. 326-337.
61. Devita, P. and W.A. Skelly, *Effect of landing stiffness on joint kinetics and energetics in the lower extremity*. Med Sci Sports Exerc, 1992. **24**(1): p. 108-115.
62. Nagai, T., T.C. Sell, A.J. House, J.P. Abt, and S.M. Lephart, *Knee proprioception and strength and landing kinematics during a single-leg stop-jump task*. Journal of athletic training, 2013. **48**(1): p. 31-38.
63. Pollard, C.D., S.M. Sigward, and C.M. Powers, *Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments*. Clinical biomechanics, 2010. **25**(2): p. 142-146.
64. Hewett, T.E., G.D. Myer, K.R. Ford, R.S. Heidt Jr, A.J. Colosimo, S.G. McLean, A.J. Van den Bogert, M.V. Paterno, and P. Succop, *Biomechanical measures of*

- neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study*. The American journal of sports medicine, 2005. **33**(4): p. 492-501.
65. Lephart, S.M., J. Abt, C. Ferris, T. Sell, T. Nagai, J. Myers, and J. Irrgang, *Neuromuscular and biomechanical characteristic changes in high school athletes: a plyometric versus basic resistance program*. British journal of sports medicine, 2005. **39**(12): p. 932-938.
 66. Thomas, A.C., S.G. McLean, and R.M. Palmieri-Smith, *Quadriceps and hamstrings fatigue alters hip and knee mechanics*. J Appl Biomech, 2010. **26**(2): p. 159-70.
 67. Haddas, R., C.R. James, and T.L. Hooper, *Lower extremity fatigue, sex, and landing performance in a population with recurrent low back pain*. J Athl Train, 2015. **50**(4): p. 378-84.
 68. Zhou, H. and H. Hu, *Human motion tracking for rehabilitation—A survey*. Biomedical signal processing control, 2008. **3**(1): p. 1-18.
 69. Winter, D.A., *Biomechanics and motor control of human movement*. 2009: John Wiley & Sons.
 70. Robertson, G.E., G.E. Caldwell, J. Hamill, G. Kamen, and S. Whittlesey, *Research methods in biomechanics*. 2013: Human kinetics.
 71. Chappell, J.D., B. Yu, D.T. Kirkendall, and W.E. Garrett, *A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks*. The American journal of sports medicine, 2002. **30**(2): p. 261-267.
 72. Malinzak, R.A., S.M. Colby, D.T. Kirkendall, B. Yu, and W.E. Garrett, *A comparison of knee joint motion patterns between men and women in selected athletic tasks*. Clinical biomechanics, 2001. **16**(5): p. 438-445.
 73. Shull, P.B., W. Jirattigalachote, M.A. Hunt, M.R. Cutkosky, and S.L. Delp, *Quantified self and human movement: a review on the clinical impact of wearable sensing and feedback for gait analysis and intervention*. Gait & Posture, 2014. **40**(1): p. 11-19.
 74. Vitali, R., S. Cain, R. McGinnis, A. Zaferiou, L. Ojeda, S. Davidson, and N. Perkins, *Method for estimating three-dimensional knee rotations using two inertial measurement units: Validation with a coordinate measurement machine*. Sensors, 2017. **17**(9): p. 1970.
 75. Sigward, S.M., M.-S.M. Chan, P.E.J.G. Lin, and posture, *Characterizing knee loading asymmetry in individuals following anterior cruciate ligament reconstruction using inertial sensors*. 2016. **49**: p. 114-119.
 76. Matijevich, E.S., L.M. Branscombe, L.R. Scott, and K.E. Zelik, *Ground reaction force metrics are not strongly correlated with tibial bone load when running across speeds and slopes: Implications for science, sport and wearable tech*. PloS one, 2019. **14**(1).
 77. McGinnis, R.S., S.M. Cain, S.P. Davidson, R.V. Vitali, N.C. Perkins, and S.G.J.S.E. McLean, *Quantifying the effects of load carriage and fatigue under load on sacral kinematics during countermovement vertical jump with IMU-based method*. 2016. **19**(1): p. 21-34.

78. Derrick, T.R., D. Dereu, and S.P. McLean, *Impacts and kinematic adjustments during an exhaustive run*. Medicine & Science in Sports & Exercise, 2002. **34**(6): p. 998-1002.
79. Elvin, N.G., A.A. Elvin, S.P. Arnoczky, and M.R. Torry, *The correlation of segment accelerations and impact forces with knee angle in jump landing*. Journal of Applied Biomechanics, 2007. **23**(3): p. 203-212.
80. Pratt, K. and S. Sigward, *Inertial Sensor Angular Velocities Reflect Dynamic Knee Loading during Single Limb Loading in Individuals Following Anterior Cruciate Ligament Reconstruction*. Sensors, 2018. **18**(10): p. 3460.
81. Shorten, M.R. and D.S. Winslow, *Spectral analysis of impact shock during running*. International Journal of Sport Biomechanics, 1992. **8**(4): p. 288-304.
82. Abt, J.P., T.C. Sell, Y. Chu, M. Lovalekar, R.G. Burdett, and S.M. Lephart, *Running kinematics and shock absorption do not change after brief exhaustive running*. The Journal of Strength & Conditioning Research, 2011. **25**(6): p. 1479-1485.
83. Mercer, J.A., P. Devita, T.R. Derrick, and B.T. Bates, *Individual effects of stride length and frequency on shock attenuation during running*. Medicine & Science in Sports & Exercise, 2003. **35**(2): p. 307-313.
84. Lafortune, M.A., M.J. Lake, and E.M. Hennig, *Differential shock transmission response of the human body to impact severity and lower limb posture*. Journal of biomechanics, 1996. **29**(12): p. 1531-1537.
85. Hargrave, M.D., C.R. Carcia, B.M. Gansneder, and S.J. Shultz, *Subtalar pronation does not influence impact forces or rate of loading during a single-leg landing*. Journal of athletic training, 2003. **38**(1): p. 18.
86. Dai, B., W.E. Garrett, M.T. Gross, D.A. Padua, R.M. Queen, and B. Yu, *The effects of 2 landing techniques on knee kinematics, kinetics, and performance during stop-jump and side-cutting tasks*. The American journal of sports medicine, 2015. **43**(2): p. 466-474.
87. Norcross, M.F., J.T. Blackburn, B.M. Goerger, and D.A.J.C.B. Padua, *The association between lower extremity energy absorption and biomechanical factors related to anterior cruciate ligament injury*. 2010. **25**(10): p. 1031-1036.
88. Decker, M.J., M.R. Torry, D.J. Wyland, W.I. Sterett, and J.R. Steadman, *Gender differences in lower extremity kinematics, kinetics and energy absorption during landing*. Clinical biomechanics, 2003. **18**(7): p. 662-669.
89. Ericksen, H.M., P.A. Gribble, K.R. Pfile, and B.G.J.J.o.a.t. Pietrosimone, *Different modes of feedback and peak vertical ground reaction force during jump landing: a systematic review*. 2013. **48**(5): p. 685-695.
90. Dufek, J.S. and B.T. Bates, *The evaluation and prediction of impact forces during landings*. Medicine and Science in Sports and Exercise, 1990. **22**(3): p. 370-377.
91. Clark, N.C., C.J. Gumbrell, S. Rana, C.M. Traole, and M.C. Morrissey, *Intratester reliability and measurement error of the adapted crossover hop for distance*. Physical Therapy in Sport, 2002. **3**(3): p. 143-151.
92. Augustsson, J., R. Thomee, C. Linden, M. Folkesson, R. Tranberg, and J. Karlsson, *Single-leg hop testing following fatiguing exercise: reliability and biomechanical analysis*. Scandinavian journal of medicine & science in sports, 2006. **16**(2): p. 111-120.

93. James, C.R., J.S. Dufek, and B.T. Bates, *Effects of injury proneness and task difficulty on joint kinetic variability*. Medicine and science in sports and exercise, 2000. **32**(11): p. 1833-1844.
94. Nordin, A.D. and J.S. Dufek, *Load accommodation strategies and movement variability in single-leg landing*. Journal of applied biomechanics, 2017. **33**(4): p. 241-247.
95. Blackburn, J.T. and D.A. Padua, *Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity*. Journal of athletic training, 2009. **44**(2): p. 174-179.
96. Hanson, A.M., D.A. Padua, J. Troy Blackburn, W.E. Prentice, and C.J. Hirth, *Muscle activation during side-step cutting maneuvers in male and female soccer athletes*. Journal of athletic training, 2008. **43**(2): p. 133-143.
97. Noyes, F.R., S.D. Barber, and R.E. Mangine, *Abnormal lower limb symmetry determined by function hop tests after anterior cruciate ligament rupture*. Am J Sports Med, 1991. **19**(5): p. 513-518.
98. Wouda, F.J., M. Giuberti, G. Bellusci, E. Maartens, J. Reenalda, B.-J.F. Van Beijnum, and P.H.J.F.i.p. Veltink, *Estimation of vertical ground reaction forces and sagittal knee kinematics during running using three inertial sensors*. 2018. **9**: p. 218.
99. Fairbank, J.C. and P.B. Pynsent, *The Oswestry disability index*. Spine, 2000. **25**(22): p. 2940-2953.
100. Tegner, Y. and J. Lysholm, *Rating systems in the evaluation of knee ligament injuries*. Clinical orthopaedics and related research, 1985(198): p. 43-49.
101. Winter, D.A.r.i.b.e., *Biomechanics of human movement with applications to the study of human locomotion*. Critical reviews in biomedical engineering, 1984. **9**(4): p. 287-314.
102. Puddle, D.L. and P.S. Maulder, *Ground reaction forces and loading rates associated with parkour and traditional drop landing techniques*. Journal of sports science & medicine, 2013. **12**(1): p. 122.
103. Portney, L.G. and M.P. Watkins, *Foundations of clinical research: applications to practice*. Vol. 892. 2009: Pearson/Prentice Hall Upper Saddle River, NJ.
104. Manal, T.J. and L. Snyder-Mackler, *Practice guidelines for anterior cruciate ligament rehabilitation: a criterion-based rehabilitation progression*. Operative Techniques in Orthopaedics, 1996. **6**(3): p. 190-196.
105. Ortiz, A., S. Olson, E. Trudelle-Jackson, M. Rosario, and H.L. Venegas, *Landing mechanics during side hopping and crossover hopping maneuvers in noninjured women and women with anterior cruciate ligament reconstruction*. PM&R, 2011. **3**(1): p. 13-20.
106. Havens, K.L. and S.M. Sigward, *Whole body mechanics differ among running and cutting maneuvers in skilled athletes*. Gait & posture, 2015. **42**(3): p. 240-245.
107. Sigward, S.M. and C.M. Powers, *Loading characteristics of females exhibiting excessive valgus moments during cutting*. Clinical biomechanics, 2007. **22**(7): p. 827-833.

108. Myer, G.D., D.A. Chu, J.L. Brent, and T.E. Hewett, *Trunk and hip control neuromuscular training for the prevention of knee joint injury*. Clinics in sports medicine, 2008. **27**(3): p. 425-448.
109. McLean, S.G., X. Huang, and A.J. van den Bogert, *Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury*. Clinical Biomechanics, 2005. **20**(8): p. 863-870.
110. Havens, K.L. and S.M. Sigward, *Cutting mechanics: relation to performance and anterior cruciate ligament injury risk*. Medicine and science in sports and exercise, 2015. **47**(4): p. 818-824.
111. Woodard, C.M., M.K. James, and S.P. Messier, *Computational methods used in the determination of loading rate: Experimental and clinical implications*. Journal of Applied Biomechanics, 1999. **15**(4): p. 404-417.
112. Collins, J. and M. Whittle, *Impulsive forces during walking and their clinical implications*. Clinical Biomechanics, 1989. **4**(3): p. 179-187.
113. Slater, A., A. Campbell, A. Smith, and L.J.S.b. Straker, *Greater lower limb flexion in gymnastic landings is associated with reduced landing force: a repeated measures study*. 2015. **14**(1): p. 45-56.
114. Phillips, J.H., S.P.J.T.J.o.S. Flanagan, and C. Research, *Effect of ankle joint contact angle and ground contact time on depth jump performance*. 2015. **29**(11): p. 3143-3148.
115. Al-Obaidi, S.M., R.M. Nelson, S. Al-Awadhi, and N. Al-Shuwaie, *The role of anticipation and fear of pain in the persistence of avoidance behavior in patients with chronic low back pain*. Spine, 2000. **25**(9): p. 1126-1131.
116. Pflingsten, M., E. Leibing, W. Harter, B. Kröner-Herwig, D. Hempel, U. Kronshage, and J. Hildebrandt, *Fear-avoidance behavior and anticipation of pain in patients with chronic low back pain: a randomized controlled study*. J Pain medicine, 2001. **2**(4): p. 259-266.
117. O'Sullivan, P., *Diagnosis and classification of chronic low back pain disorders: maladaptive movement and motor control impairments as underlying mechanism*. Manual therapy, 2005. **10**(4): p. 242-255.
118. Seay, J.F., R.E. Van Emmerik, and J. Hamill, *Trunk bend and twist coordination is affected by low back pain status during running*. Eur J Sport Sci, 2014. **14**(6): p. 563-8.
119. Baliunas, A., D. Hurwitz, A. Ryals, A. Karrar, J. Case, J. Block, and T. Andriacchi, *Increased knee joint loads during walking are present in subjects with knee osteoarthritis*. Osteoarthritis and cartilage, 2002. **10**(7): p. 573-579.
120. Callahan, D.M., T.W. Tourville, J.R. Slauterbeck, P.A. Ades, J. Stevens-Lapsley, B.D. Beynnon, and M.J. Toth, *Reduced rate of knee extensor torque development in older adults with knee osteoarthritis is associated with intrinsic muscle contractile deficits*. Exp Gerontol, 2015. **72**: p. 16-21.
121. McGregor, A.H. and D.W. Hukins, *Lower limb involvement in spinal function and low back pain*. J Back Musculoskelet Rehabil, 2009. **22**(4): p. 219-22.
122. Granata, K., D. Padua, and S. Wilson, *Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks*. Journal of Electromyography and Kinesiology, 2002. **12**(2): p. 127-135.

123. Sueki, D.G., J.A. Cleland, and R.S. Wainner, *A regional interdependence model of musculoskeletal dysfunction: research, mechanisms, and clinical implications*. Journal of manual & manipulative therapy, 2013. **21**(2): p. 90-102.
124. Nicholas, J.A., *The importance of a simplified classification of motion in sports in relation to performance*. Orthop Clin North Am, 1977. **8**: p. 499-532.
125. Edwards, W.B., E.D. Ward, S.A. Meardon, and T.R. Derrick, *The use of external transducers for estimating bone strain at the distal tibia during impact activity*. Journal of biomechanical engineering, 2009. **131**(5): p. 051009.
126. Sigward, S.M., M.-S.M. Chan, and P.E. Lin, *Characterizing knee loading asymmetry in individuals following anterior cruciate ligament reconstruction using inertial sensors*. Gait & posture, 2016. **49**: p. 114-119.
127. Smith, D.G., D.M. Ehde, M.W. Legro, G.E. Reiber, M. del Aguila, and D.A. Boone, *Phantom limb, residual limb, and back pain after lower extremity amputations*. Clinical Orthopaedics, 1999. **361**: p. 29-38.
128. Ehde, D.M., D.G. Smith, J.M. Czerniecki, K.M. Campbell, D.M. Malchow, and L.R. Robinson, *Back pain as a secondary disability in persons with lower limb amputations*. Archives of physical medicine and rehabilitation, 2001. **82**(6): p. 731-734.
129. Nolan, L. and A. Lees, *The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees*. Prosthetics and orthotics international, 2000. **24**(2): p. 117-125.
130. Prince, F., P. Allard, R. Therrien, and B. McFadyen, *Running gait impulse asymmetries in below-knee amputees*. Prosthetics and Orthotics International, 1992. **16**(1): p. 19-24.
131. Fairbank, J., J. Couper, J. Davies, and J. O'brien, *The Oswestry low back pain disability questionnaire*. Physiotherapy, 1980. **66**(8): p. 271-273.
132. Paterno, M.V., L.C. Schmitt, K.R. Ford, M.J. Rauh, G.D. Myer, B. Huang, and T.E. Hewett, *Biomechanical measures during landing and postural stability predict second anterior cruciate ligament injury after anterior cruciate ligament reconstruction and return to sport*. The American journal of sports medicine, 2010. **38**(10): p. 1968-1978.
133. Bijur, P.E., W. Silver, and E.J. Gallagher, *Reliability of the visual analog scale for measurement of acute pain*. Academic emergency medicine, 2001. **8**(12): p. 1153-1157.
134. Lathrop-Lambach, R.L., J.L. Asay, S.T. Jamison, X. Pan, L.C. Schmitt, K. Blazek, R.A. Siston, T.P. Andriacchi, and A.M. Chaudhari, *Evidence for joint moment asymmetry in healthy populations during gait*. Gait & posture, 2014. **40**(4): p. 526-531.
135. Schmitt, L.C., M.V. Paterno, and T.E. Hewett, *The impact of quadriceps femoris strength asymmetry on functional performance at return to sport following anterior cruciate ligament reconstruction*. journal of orthopaedic & sports physical therapy, 2012. **42**(9): p. 750-759.
136. Garrick, J.G., *The frequency of injury, mechanism of injury, and epidemiology of ankle sprains*. The American journal of sports medicine, 1977. **5**(6): p. 241-242.

137. Leinonen, V., M. Kankaanpää, M. Luukkonen, O. Hänninen, O. Airaksinen, and S. Taimela, *Disc herniation-related back pain impairs feed-forward control of paraspinal muscles*. Spine, 2001. **26**(16): p. E367-E372.
138. Leinonen, V., S. Määttä, S. Taimela, A. Herno, M. Kankaanpää, J. Partanen, M. Kansanen, O. Hänninen, and O. Airaksinen, *Impaired lumbar movement perception in association with postural stability and motor-and somatosensory-evoked potentials in lumbar spinal stenosis*. Spine, 2002. **27**(9): p. 975-983.
139. Radebold, A., J. Cholewicki, G.K. Polzhofer, and H.S. Greene, *Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain*. Spine, 2001. **26**(7): p. 724-730.
140. Lephart, S.M., D.M. Pincivero, J.L. Giraido, and F.H. Fu, *The role of proprioception in the management and rehabilitation of athletic injuries*. The American journal of sports medicine, 1997. **25**(1): p. 130-137.
141. Kiernan, D., D.A. Hawkins, M.A. Manoukian, M. McKallip, L. Oelsner, C.F. Caskey, and C.L. Coolbaugh, *Accelerometer-based prediction of running injury in National Collegiate Athletic Association track athletes*. Journal of biomechanics, 2018. **73**: p. 201-209.
142. Neugebauer, J.M., K.H. Collins, and D.A. Hawkins, *Ground reaction force estimates from ActiGraph GT3X+ hip accelerations*. PloS one, 2014. **9**(6): p. e99023.
143. Heidari, B., *Knee osteoarthritis prevalence, risk factors, pathogenesis and features: Part I*. Caspian journal of internal medicine, 2011. **2**(2): p. 205.
144. Chu, C.R. and T.P. Andriacchi, *Dance between biology, mechanics, and structure: a systems-based approach to developing osteoarthritis prevention strategies*. Journal of Orthopaedic Research, 2015. **33**(7): p. 939-947.
145. Andriacchi, T.P. and A. Mündermann, *The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis*. Current opinion in rheumatology, 2006. **18**(5): p. 514-518.
146. Strowbridge, N., *Gender differences in the cause of low back pain in British soldiers*. Journal of the Royal Army Medical Corps, 2005. **151**(2): p. 69-72.
147. Leboeuf-Yde, C. and K.O.J.S. Kyvik, *At what age does low back pain become a common problem?: A study of 29,424 individuals aged 12-41 years*. Spine, 1998. **23**(2): p. 228-234.
148. Chenot, J.-F., A. Becker, C. Leonhardt, S. Keller, N. Donner-Banzhoff, J. Hildebrandt, H.-D. Basler, E. Baum, M.M. Kochen, and M. Pfingsten, *Sex differences in presentation, course, and management of low back pain in primary care*. The Clinical Journal of Pain, 2008. **24**(7): p. 578-584.
149. Blackwell, D.L., J.W. Lucas, and T.C. Clarke, *Summary health statistics for US adults: national health interview survey, 2012*. Vital and health statistics. Series 10, Data from the National Health Survey, 2014(260): p. 1-161.
150. Trompeter, K., D. Fett, and P. Platen, *Prevalence of Back Pain in Sports: A Systematic Review of the Literature*. Sports Medicine, 2017: p. 1-25.
151. Harber, P., L. Peña, P. Hsu, E. Billet, D. Greer, and K. Kim, *Personal history, training, and worksite as predictors of back pain of nurses*. American Journal Of Industrial Medicine, 1994. **25**(4): p. 519-526.

152. Roos, E.M. and L.S. Lohmander, *The Knee injury and Osteoarthritis Outcome Score (KOOS): from joint injury to osteoarthritis*. Health and quality of life outcomes, 2003. **1**(1): p. 64.
153. Lepley, L.K. and R.M. Palmieri-Smith, *Quadriceps Strength, Muscle Activation Failure, and Patient-Reported Function at the Time of Return to Activity in Patients Following Anterior Cruciate Ligament Reconstruction: A Cross-sectional Study*. J Orthop Sports Phys Ther, 2015. **45**(12): p. 1017-25.
154. Palmieri-Smith, R.M. and L.K. Lepley, *Quadriceps Strength Asymmetry After Anterior Cruciate Ligament Reconstruction Alters Knee Joint Biomechanics and Functional Performance at Time of Return to Activity*. Am J Sports Med, 2015. **43**(7): p. 1662-9.
155. Kosik, K.B., M. Terada, R. McCann, A. Thomas, N. Johnson, and P. Gribble, *Differences in temporal gait mechanics are associated with decreased perceived ankle joint health in individuals with chronic ankle instability*. Gait & posture, 2019. **70**: p. 403-407.
156. Hubbard, T.J., L.C. Kramer, C.R. Denegar, and J. Hertel, *Correlations among multiple measures of functional and mechanical instability in subjects with chronic ankle instability*. Journal of athletic training, 2007. **42**(3): p. 361.
157. Houston, M.N., J.M. Hoch, M.L. Gabriner, J.L. Kirby, and M.C. Hoch, *Clinical and laboratory measures associated with health-related quality of life in individuals with chronic ankle instability*. Physical therapy in sport, 2015. **16**(2): p. 169-175.
158. Roos, E.M., H.P. Roos, L.S. Lohmander, C. Ekdahl, and B.D. Beynnon, *Knee Injury and Osteoarthritis Outcome Score (KOOS)—development of a self-administered outcome measure*. Journal of Orthopaedic & Sports Physical Therapy, 1998. **28**(2): p. 88-96.
159. Collins, N.J., D. Misra, D.T. Felson, K.M. Crossley, and E.M. Roos, *Measures of knee function: International Knee Documentation Committee (IKDC) Subjective Knee Evaluation Form, Knee Injury and Osteoarthritis Outcome Score (KOOS), Knee Injury and Osteoarthritis Outcome Score Physical Function Short Form (KOOS-PS), Knee Outcome Survey Activities of Daily Living Scale (KOS-ADL), Lysholm Knee Scoring Scale, Oxford Knee Score (OKS), Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC), Activity Rating Scale (ARS), and Tegner Activity Score (TAS)*. Arthritis care & research, 2011. **63**(S11): p. S208-S228.
160. Pfingsten, M., E. Leibling, W. Harter, B. Kröner-Herwig, D. Hempel, U. Kronshage, and J. Hildebrandt, *Fear-avoidance behavior and anticipation of pain in patients with chronic low back pain: a randomized controlled study*. Pain medicine, 2001. **2**(4): p. 259-266.
161. Feuerstein, M. and P. Beattie, *Biobehavioral factors affecting pain and disability in low back pain: mechanisms and assessment*. Physical therapy, 1995. **75**(4): p. 267-280.

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Education

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2014-2016	University of Michigan Master of Science, Kinesiology Thesis: Underlying Factors of Neural Activity that Regulate Torque Development after Anterior Cruciate Ligament Reconstruction

Professional Positions

Undergraduate Research Assistant – Center for Human Performance, Biomechanics Laboratory, California State University Fullerton, 2013
Graduate Research Assistant – Bone Joint Injury Prevention and Rehabilitation Laboratory, University of Michigan, 2014-2016
Academic Mentor – Athletic Department, University of Michigan, 2014-2016
Independent Graduate Research Assistant – Rehabilitation Biomechanics Laboratory, University of Michigan, 2015
Summer Research Scholar, Michigan Institute of Clinical and Health Research, University of Michigan, 2015
Adjunct Faculty – Kinesiology and Health Sciences, Georgetown College, 2018-2019
Graduate Research Assistant, Sports Medicine Research Institute, University of Kentucky, 2016-Present

Scholastic and Professional Honors

Dean's List, 2010-2013
University of Michigan Kinesiology Travel Award, 2015
University of Kentucky Graduate School Travel Award, 2017
American Society of Biomechanics Student Travel Award 2017
American Association for the Advancement of Science and
Association of Public & Land-Grant Universities, CASE Workshop:
Catalyzing Advocacy for Science and Engineering – University Selectee 2018
University of Kentucky College of Health Sciences Academic Excellence Award, 2018
University of Kentucky Rehabilitation Sciences Travel Award, 2018, 2019
National Science Policy Network Annual Conference Travel Award, 2018
Center for Graduate and Professional Diversity Initiatives
Professional Development Award, 2018
American Society of Biomechanics, Student Grant in Aid, 2018

Publications

- Johnson, AK, Palmieri-Smith, RM, Lepley, LK. 2016. Contribution of Neuromuscular Factors to Quadriceps Asymmetry After Anterior Cruciate Ligament Reconstruction. *Journal of Athletic Training*. 53(4):347-354. 2018. DOI: 10.4085/1062-6050-463-16
- Winters, JD, Heebner, NR, Johnson, AK, Poploski, KM, Royer, SD, Nagai, T, Randall, C, Abt, JP, Lephart, SM. Advanced Military Training Effects on Performance Characteristics in Marine Corps Forces Special Operations Command Operators. 2019. *Journal of Strength and Conditioning Research*
- Quintana, C, Grimshaw, B, Rockwood, HE, Heebner, NR, Johnson, AK, Ryan, KD, Mattacola, CG. Differences in Head Accelerations and Physiological Demand between Live and Simulated Professional Horse Racing. 2019. *Journal of Comparative Exercise Physiology*