RELATIONSHIP BETWEEN VOLUNTARY QUADRICEPS ACTIVATION AND GAIT BIOMECHANICS IN PATIENTS WITH ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

Grace Elizabeth Jungclas

A thesis submitted to the faculty at the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Masters of Arts in the Department of Exercise & Sport Science in the College of Arts & Sciences.

Chapel Hill
2015

Approved by:
J. Troy Blackburn
Brian Pietrosimone
Derek Pamukoff
ABSTRACT
Grace Elizabeth Jungclas: Relationship between Voluntary Quadriceps Activation and Gait Biomechanics in Patients with Anterior Cruciate Ligament Reconstruction (Under the direction of J. Troy Blackburn)

CONTEXT: Patients with anterior cruciate ligament reconstruction (ACL-R) have quadriceps dysfunction yet the relationship to gait biomechanics remains unclear. OBJECTIVE: Examine the relationship between quadriceps function (CAR) and gait biomechanics in patients with ACL-R. DESIGN: Cross-sectional study. SETTING: Research laboratory. SUBJECTS: 31 physically active volunteers with ACL-R. INTERVENTION(S): CAR quantified quadriceps function. Lower extremity kinematics and kinetics were captured during gait with a Vicon Nexus interfaced with a force plate. A one-tailed Pearson r correlation was used for statistical analyses. MAIN OUTCOME MEASURE(S): Gait biomechanics were identified during stance phase. Forces were normalized to body weight and moments were normalized to the product of weight and height. RESULTS: Subjects quadriceps dysfunction displayed aberrant gait biomechanics. CONCLUSIONS: There is a relationship between quadriceps function and gait biomechanics linking altered gait and quadriceps dysfunction to the development of OA.
# TABLE OF CONTENTS

LIST OF FIGURES .................................................................................................................. vi

LIST OF TABLES ...................................................................................................................... vii

LIST OF ABBREVIATIONS ..................................................................................................... viii

CHAPTER 1: Introduction ....................................................................................................... 1
   Research Questions: ........................................................................................................... 3
   Variables: .......................................................................................................................... 4

CHAPTER 2: Literature Review ............................................................................................. 6
   Prevalance ........................................................................................................................ 6
   Financial Burden ............................................................................................................... 9
   Anatomy ............................................................................................................................ 9
   Quadriceps Function and Gait ......................................................................................... 12
      Part I: Quadriceps Function during Gait ................................................................. 12
      Part II: Alterations in Quadriceps Function following ACL Injury ....................... 13
      Part III: Kinematic and Kinetic Aberrations following ACL-R ................................ 15
      Part IV: How Alterations in Gait Influence Osteoarthritis ....................................... 16
   Summary ........................................................................................................................ 17

CHAPTER 3: Research Methods ......................................................................................... 18
LIST OF FIGURES

Figure 1: HUMAC NORM subject set up. ................................................................. 32
Figure 2: CAR Computer program with feedback lines. ........................................... 33
Figure 3: Relationship between CAR and Peak Internal Knee Extension Moment (N*m) .... 34
Figure 4: Relationship between CAR and Peak Knee Flexion Angle (°) ......................... 35
Figure 5: Relationship between CAR and Peak Knee Varus Angle (°) .......................... 36
Figure 6: Relationship between CAR and Peak Loading Rate of the vGRF (N/s) .............. 37
Figure 7: Relationship between CAR and Peak Knee Flexion Angle (°) ......................... 38
Figure 8: Relationship between CAR and Peak Internal Knee Extension Moment (N*m) .... 39
LIST OF TABLES

Table 1. Subject Demographics .............................................................................................................. 40

Table 2: Descriptive Statistics of CAR, Gait Kinematics & Kinetics ...................................................... 41

Table 3: Relationship between CAR and gait biomechanics ................................................................. 42
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACL</td>
<td>Anterior cruciate ligament</td>
</tr>
<tr>
<td>ACL-D</td>
<td>Anterior cruciate ligament deficiency</td>
</tr>
<tr>
<td>ACL-R</td>
<td>Anterior cruciate ligament reconstruction</td>
</tr>
<tr>
<td>ADL</td>
<td>Activity of daily living</td>
</tr>
<tr>
<td>AM</td>
<td>Anteromedial</td>
</tr>
<tr>
<td>AMI</td>
<td>Arthrogenic muscle inhibition</td>
</tr>
<tr>
<td>BPTB</td>
<td>Bone patellar tendon bone graft</td>
</tr>
<tr>
<td>BMI</td>
<td>Body mass index</td>
</tr>
<tr>
<td>BW</td>
<td>Body weight</td>
</tr>
<tr>
<td>CAR</td>
<td>Central activation ratio</td>
</tr>
<tr>
<td>CNS</td>
<td>Central nervous system</td>
</tr>
<tr>
<td>ECM</td>
<td>Extracellular matrix</td>
</tr>
<tr>
<td>HT</td>
<td>Hamstring tendon graft</td>
</tr>
<tr>
<td>IGC</td>
<td>Initial ground contact</td>
</tr>
<tr>
<td>OA</td>
<td>Osteoarthritis</td>
</tr>
<tr>
<td>MVIC</td>
<td>Maximal voluntary isometric contraction</td>
</tr>
<tr>
<td>PL</td>
<td>Posterolateral</td>
</tr>
<tr>
<td>pMVIC</td>
<td>Potential maximal voluntary isometric contraction</td>
</tr>
<tr>
<td>QOL</td>
<td>Quality of life</td>
</tr>
<tr>
<td>RFD</td>
<td>Rate of force development</td>
</tr>
<tr>
<td>ROM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>vGRF</td>
<td>Vertical ground reaction force</td>
</tr>
</tbody>
</table>
CHAPTER 1: Introduction

One in 3,500 individuals in the general population sustains an anterior cruciate ligament (ACL) rupture every year in the United States.\(^1\) The standard of treatment for this injury is surgical reconstruction (ACL-R), which occurs at a rate of 200,000 reconstructions per year.\(^2\) ACL-R represents a substantial financial cost, with each procedure costing approximately $50,000.\(^3,4\) However, this figure does not reflect costs attributable to patients with ACL-R who develop knee osteoarthritis (OA). Patients with ACL-R are 3.62 times more likely to develop OA compared to healthy controls with no injury.\(^5\) OA refers to the gradual degradation of articular cartilage,\(^5,7\) and is the most common cause of disability in the United States.\(^8\)\(^,\)\(^9\)\(^,\)\(^10\)

Furthermore, OA contributes to pain-induced sedentary behavior, which influences comorbidities such as cardiovascular disease and diabetes.\(^11\) Overall, the direct annual cost associated with knee osteoarthritis in the United States is approximately $51 billion.\(^12\) Even with surgical intervention, patients who suffer ACL injuries often develop OA as early as 5 years following injury.\(^6\)\(^,\)\(^13\)\(^,-\)\(^15\) More than half of these ACL injuries occur in individuals between the ages of 15 and 25 years, thus this demographic is of particular importance due to their potential to develop OA very early in life.\(^2\) ACL injury currently incurs an annual lifetime financial burden of $7.6 billion dollars in the United States, $2.78 billion of which is attributable specifically to post-traumatic OA.\(^3\)

Traumatic knee injuries, particularly those of the ACL, dramatically increase the risk of developing knee OA.\(^8\)\(^,\)\(^16\)\(^,-\)\(^18\) A growing body of literature suggests that this heightened risk of OA is linked to chronic quadriceps dysfunction. Quadriceps weakness is a common symptom
following ACL-R\textsuperscript{19-21} due to a neuromuscular phenomenon called arthrogenic muscle inhibition (AMI).\textsuperscript{22} AMI refers to the reflexive inhibition of voluntarily activation of a muscle, and is commonly measured via the central activation ratio (CAR).\textsuperscript{22,23} CAR represents the portion of the quadriceps motor neuron pool that can be activated voluntarily.\textsuperscript{24} Furthermore, AMI persists long after completion of traditional rehabilitation programs,\textsuperscript{19,20,25,26} and may lead to aberrant gait biomechanics that contribute to the development of OA.\textsuperscript{27} Proper quadriceps function is necessary for shock absorption during the early stance phase of gait as it eccentrically slows the limb upon contact. Therefore, diminished quadriceps function may contribute to a greater impulsive/high rate loading at the knee joint during gait.\textsuperscript{22,28-30} Impulse describes the relationship of force applied over time, and greater impulsive loading contributes to the development and progression of OA.\textsuperscript{31,32} Articular cartilage is viscoelastic and is sensitive to loading rate,\textsuperscript{33} and greater loading rates contribute to greater cartilage degeneration.\textsuperscript{31,32}

Several studies have shown that patients with ACL-R simultaneously display AMI\textsuperscript{19,20,25,26} and alterations in gait biomechanics.\textsuperscript{34-39} However, it is unclear how quadriceps weakness caused by AMI influences characteristics of gait. In patients with ACL-R, biomechanical alterations have been identified in the frontal and sagittal planes.\textsuperscript{34} Specifically, alterations in kinematics (peak knee flexion,\textsuperscript{35,36,38} extension,\textsuperscript{35,38} and varus\textsuperscript{37,38} angles) and kinetics (knee extension moment\textsuperscript{36,39} and vertical ground reaction\textsuperscript{37-39} force) have been identified which may influence OA development. For example, lesser quadriceps activation results in a lesser knee extensor moment during gait. These patients also display less knee flexion, or quadriceps avoidance gait.\textsuperscript{40} Decreased knee flexion excursion during gait\textsuperscript{22,41-44} reduces the time over which the ground reaction force is absorbed, thus increasing impulsive loading. If the ground reaction force is not adequately absorbed in the sagittal plane, the energy may be
dissipated by compensatory action in the frontal plane.\textsuperscript{45} Similarly, quadriceps deficits leading to varus compression of the medial joint ultimately contribute to the development and progression of OA.\textsuperscript{46} However, there are few studies that characterize the relationship between quadriceps dysfunction and gait biomechanics in patients with ACL-R.

Many studies have reported alterations in gait biomechanics in patients with OA that appear to be influenced by quadriceps dysfunction.\textsuperscript{42,43,45,47-49} Altered knee adduction moment,\textsuperscript{42,47-49} knee varus angle,\textsuperscript{45} knee flexion angle\textsuperscript{43,45} and vertical ground reaction force\textsuperscript{37-39} have been observed in patients with OA compared to healthy cohorts. It is clear that patients with ACL-R display changes in gait biomechanics similar to those noted in knee OA patients, and are at greater risk for developing knee OA.\textsuperscript{8,16-18} However it’s unclear if deficits in quadriceps activation contribute to these deviations following ACL-R. Further research is needed to confirm the influence of quadriceps dysfunction on gait abnormalities in patients with ACL-R. These data will clarify the role of quadriceps dysfunction in the development of knee OA, and may inform the development of preventive interventions for ACL-R patients such as gait training and enhanced methods for quadriceps strengthening.

The purpose of this cross-sectional study is to examine the relationship between quadriceps function (CAR) and gait biomechanics (peak knee flexion angle, internal knee extension moment, knee varus angle, and loading rate of the vertical ground reaction force) in patients with ACL-R. This study will address the following research questions.

\textbf{Research Questions:}

1. What is the relationship between CAR and peak knee flexion angle during the first 50\% of the stance phase of gait?

\textit{H}_{1}: A lower CAR will be associated with a smaller peak knee flexion angle.
2. What is the relationship between CAR and peak internal knee extension moment during the first 50% of the stance phase of gait?
   
   \( H_2 \): A lower CAR will be associated with a smaller peak internal knee extension moment.

3. What is the relationship between CAR and peak knee varus angle during the first 50% of the stance phase of gait?
   
   \( H_3 \): A lower CAR will be associated with a greater peak knee varus angle.

4. What is the relationship between CAR and peak loading rate of the vertical ground reaction force during the first 50% of the stance phase of gait?
   
   \( H_4 \): A lower CAR will be associated with a greater peak vertical ground reaction force loading rate.

**Variables:**

**Criterion Variable:**

- Central Activation Ratio: the ratio of torque (N*m) generated from a voluntary contraction to the torque generated by maximal electrical stimulation

**Predictor Variables:**

- Gait biomechanics during the first 50% of stance phase:
  
  **Peak Knee Flexion Angle:** maximum sagittal plane angle (°).
  
  **Peak Internal Knee Extensor Moment:** the maximum sagittal plane net moment at the knee joint directed in extension normalized to the product of weight and height (N*m).
  
  **Peak Knee Varus Angle:** maximum frontal plane angle (°).
**Loading Rate of the Peak Vertical Ground Reaction Force:** the ratio of the peak vGRF to the time from initial ground contact to peak vGRF normalized to body weight (BW/s).

**Operational Definitions:**

Initial Ground Contact (IGC): time at which the vGRF exceeds 20N

Toe-off: time which vGRF falls below 20N

Stance phase: the time between IGC and toe-off
CHAPTER 2: Literature Review

Prevalence

The prevalence of anterior cruciate ligament (ACL) injuries in the general population is one in 3,500.\textsuperscript{50} ACL reconstruction (ACL-R) occurs at a rate of 200,000 per year. More than half of all ACL injuries occur in individuals between the ages of 15 and 25 years.\textsuperscript{51} Following ACL-R, 50\% of subjects display radiographic evidence of tibiofemoral osteoarthritis (OA) after 12 years in women and 14 years in men.\textsuperscript{8} In a sample of male soccer players who were 14 years post-ACLR, 78\% had radiographic changes and 47\% had changes equivalent to grade 2 OA on the Kellgren and Lawrence scale.\textsuperscript{52} In a similar study, a sample of female soccer players 12 years post-ACLR 82\% had radiographic changes and 51\% had changes equivalent to grade 2 on the Kellgren and Lawrence scale.\textsuperscript{13} Subject age at initial ACL injury affects OA development, as OA can appear as early as 5 years following ACL injury in individuals over the age of 30, while OA onset occurs after roughly 12 years in individuals ages 17-30 years of age.\textsuperscript{15} However, in 127 studies investigating the prevalence and rate of OA after ACL-R, there is still considerable variability in the reported rate and prevalence which is likely attributable to differences in the injury, treatment of the injury, the individual, and the methods used to assess these characteristics. For instance, concomitant meniscal injury greatly affects the likelihood of OA development.\textsuperscript{52} Furthermore, patients treated with ACL-R versus conservative rehabilitation have different risks associated with OA; one study showed that the risk of developing OA was significantly greater in patients with ACL-R compared to a group treated with conservative (i.e. non-surgical) rehabilitation.\textsuperscript{53} However, these results may be skewed by the fact that subjects
were placed in a plaster cast for 3 weeks before beginning rehabilitation for only 2 months, a component which is not consistent with current clinical practice. However, a review of the literature provided conflicting results, stating that the risk of developing OA is similar regardless of whether a surgical or non-surgical approach is utilized. Age and characteristics of the individual such as bony alignment of the knee joint and obesity affect the rate of OA development. This degenerative joint disease decreases quality of life (QOL) and limits activities of daily living (ADL). OA is the most common form of arthritis, affecting 37% of the population, and is the most common cause of physical disability in the US. Furthermore, 55% of all OA is reported within the tibiofemoral joint. The medial compartment is most commonly affected in patients with tibiofemoral OA. Overall, knee prevalence reportedly ranges 10-90 percent 10-20 years post ACL-R.

OA is a long-term consequence for patients with ACL injury treated with reconstruction. ACL injury dramatically increases the risk of developing OA, as patients with ACL-R are 3.62 times more likely to develop OA compared to the contralateral control and 3.84 times more likely to develop moderate or severe OA on the Kellgren-Lawrence scale. The risk of developing knee OA does not differ between surgically and non-surgically managed cases in patients with ACL-R. Furthermore, a study of athletes using the contralateral limb as a control found that OA was present in both knees and OA in the ACL-R knee was not significantly greater in severity than the uninjured knee.

ACL injury leads to chronic alterations in joint loading patterns during gait causing degredation to the tibiofemoral articular cartilage and the menisci, thus increasing the risk of developing OA. This is one example of the how the risk of developing OA is related to injury. Anthropometric measurements such as body mass index (BMI) and age also contribute to the
risk of developing OA.\textsuperscript{16} Age is a risk factor for OA but is often overlooked in patients with ACL-R because the population is generally young.\textsuperscript{16} BMI is associated with knee OA, more specifically joint space narrowing, following ACL injury.\textsuperscript{16} More importantly, greater BMI in patients with ACL-R is a significant risk factor for meniscal injury and articular cartilage damage.\textsuperscript{58} A systematic review of case-control and cohort studies found that the second greatest risk factor for developing OA was previous knee trauma only following obesity when compared to healthy controls. However, the limitation of this study was that knee trauma was a very broad category and was not defined with inclusion criteria.\textsuperscript{17} Another systematic review of 31 studies revealed that the greatest risk factor for OA is an ACL injury compounded by a meniscus injury.\textsuperscript{59} Chondral damage to the tibiofemoral joint is positively correlated with the extent of a meniscus injury.\textsuperscript{16} Furthermore, the risk of OA is related to the chosen surgical intervention. Decisions about the timing of the surgical intervention and the graft choice can impact the risk of OA.\textsuperscript{60} ACL-R delay after initial ACL injury is a risk factor for tibiofemoral OA.\textsuperscript{61} Furthermore, OA developed in 52% of patients with ACL-R that waited 6 or more months for surgical intervention compared to 18% of patients who received ACL-R less than 6 months after injury.\textsuperscript{62} There has been little research about the impact of graft choice upon OA.\textsuperscript{16} The literature reveals controversy about the increased risk of OA based upon graft choice. One study found no difference in OA changes between groups of bone-patellar tendon-bone (BPTB) and hamstring tendon (HT) grafts.\textsuperscript{60,63} However, three studies found OA to be more prevalent with BPTB grafts compared to the HT grafts.\textsuperscript{61,63,64} The combination of a lack of meniscal pathology, the choice of a hamstring graft and timely surgical intervention may decrease the risk of OA, but no evidence supports the ability of ACL-R surgery to prevent development of OA.\textsuperscript{16}
Financial Burden

Rehabilitation as an intervention for a typical patient with an ACL injury incurs a lifetime cost of approximately $88,000 per person; whereas ACL-R saves $50,000 by only incurring a cost of approximately $38,000. A common cost-effective ratio of $50,000 has been documented to compare costs of interventions. Hamstring autograft is the least costly intervention at $5,375 per surgery, whereas patellar tendon autograft costs $5,580 per surgery, and an allograft costs $6,958 per surgery. However, ACL-R still results in an annual cost of $2.78 billion.

The direct annual cost of tibiofemoral OA in the US is approximately $51 billion. This estimate incorporates medical costs including all health care related resources (i.e. visits, procedures, testing, pharmaceuticals, adaptive aids), time loss pay, and costs associated with activities of daily living (i.e. inability to do chores, paid and unpaid help for household chores). The mean lifetime cost to society for physical therapy for a patient with ACL-R is $88,538. If ACL-R leads to OA and does not respond to treatment, the cost of a joint replacement is approximately $25,000 per surgery incurring an annual cost of $16 billion to the US. However, indirect costs are immeasurable and include lost wages and productivity, and disability. It was estimated that the cost of OA secondary to ACL-R increases 60% from mild to moderate and from moderate to end-stage OA. Also, OA contributes to comorbidities that are risk factors for cardiovascular diseases, including hypertension, high total cholesterol, and diabetes as compared to the healthy population.

Anatomy

The ACL is structurally composed of two bundles; the anteromedial (AM) and posterolateral (PL) bundles are named for the orientation of their insertions on the tibia.
the knee is extended the PL bundle is taught and the AM bundle is moderately lax. As the knee moves into flexion the AM bundle becomes taught while the PM bundle becomes lax. The primary function of the ACL is prevention of anterior translation of the tibia, but it also provides resistance to internal rotation of the tibia and tibiofemoral valgus. The ACL is dense with mechanoreceptors, thus it also functions as a neural organ. This provides sensory information about joint position, load, and movement to the spinal cord and supraspinal centers.

Articular cartilage refers to hyaline cartilage that is 2-4mm thick and, in the knee, coats the femoral condyles and the tibial plateaus. The biomechanical function of articular cartilage includes joint lubrication, decreasing friction, and transmitting joint loads. It is an avascular structure that is constructed of chondrocytes and an extracellular matrix (ECM). Chondrocytes generate and repair the ECM. The ECM consists of water, collagen, proteoglycans and other glycoproteins. The functional shock absorption of tensile loads within cartilage originates with collagen and proteoglycans. The structure of collagen, a triphle helix, functions to supports the matrix against shear and tensile stresses. Type II collagen embodies about 90% of collagen within the ECM and is knitted with proteoglycans. A proteoglycan is made of a protein core with monosaccharide attachments. A high density of negatively charged monosaccharide chains, commonly found within proteoglycans, causes a pressure gradient supporting diffusion of water into cartilage which is critical to maintain regeneration and repair of cartilage. The high osmotic pressure is responsible for up to 90% resistance during normal loading, i.e. a gait task. Furthermore, normal compressive loading causes a peripheralization of interstitial fluid and pressure. This fluid peripheralization places tension upon the peripheral proteoglycans and
collagen fibers within the ECM. Tensile loading can compromise the integrity of proteoglycans and collagen, thus inabling the regeneration and health of the cartilage.\textsuperscript{74}

Articular cartilage is viscoelastic and sensitive to the rate of loading. Normal load-bearing mechanics and fluid diffusion within the cartilage are altered during impulsive loading. When the rate of loading is increased, the articular cartilage does not have ample time to properly peripheralize fluid, causing the outermost cartilage to stiffen or become brittle.\textsuperscript{75} Brittle structures are more likely to fail.\textsuperscript{76} This is congruent with the stress strain model which explains that the magnitude, frequency, and rate of loading, or a combination thereof, can cause tissue damage.\textsuperscript{33}

Though cartilage does not possess nociceptors, the pain that accompanies the breakdown of articular cartilage can be debilitating. A literature review examined cartilage-related pain theories and proposed that the mechanism of this pain is multifaceted including the peripheral nervous system, inflammation, bony pathology, and central sensitization. The peripheral mechanism is initiated by information stemming from A\textsubscript{\(\delta\)} and C fibers located in the synovial capsule, periarticular ligaments, menisci, adjacent periosteum and subchondral bone. These fibers are stimulated by mechanical, thermal, or chemical noxious stimuli. Inflammation contributes to pain by sensitizing nociceptive primary afferent neurons. Furthermore, a lack of articular cartilage can cause harm to the underlying bone. Pain can result from any of the following conditions: osteophyte genesis, osseous remodeling, decreased vascularization and elevated interosseous pressure. The most controversial theory is the neuropathic mechanism. Central sensitization refers to when chronic nociception from the periphery to the central nervous system causes hypersensitivity to painful stimuli, thus lowering the threshold and increasing the excitability of nociceptors. This decreased threshold affects the ascending and descending
nociceptive pathways and ultimately can cause the sensation of pain which would have not
previously been triggered. However, this theory supports the contribution of temporal, spatial
and threshold changes associated with chronic pain.\textsuperscript{77}

\textbf{Quadriceps Function and Gait}

\textbf{Part I: Quadriceps Function during Gait}

The quadriceps primarily functions to produce concentric knee extension\textsuperscript{78,79} However, during the early part of the stance phase of gait the quadriceps act eccentrically to control knee
flexion, decelerate the limb, and absorb ground contact force.\textsuperscript{80} Full extension is the most stable
weight bearing position, but it is not desirable because the impact of body weight will be
transferred through the limb with minimal cushioning or support from muscles. Inadequate
quadriceps strength and/or activation manifest as lesser knee flexion and knee extension
moment. Reducing the demand upon the quadriceps by flexing the knee less preserves weight
bearing stability without substantial quadriceps activity.\textsuperscript{81} This knee extended position is present
in patients with ACL-R and OA\textsuperscript{34-36,38} which can be problematic for joint loading.

The quadriceps muscle group functions as a shock absorber in the sagittal plane.\textsuperscript{22}
Decreased knee flexion results in a “stiff” landing causing an increase in joint loading. Similar to
ground contact during walking gait, a drop landing task in healthy human knees with an effusion
injection caused decreased knee flexion and an increased peak vertical ground reaction force.\textsuperscript{82}
When knee flexion is decreased, impact force which is not absorbed in the sagittal plane must
be absorbed in the frontal and/or transverse planes.\textsuperscript{42,43} Patients with knee OA display lesser knee
flexion and greater knee varus during gait.\textsuperscript{42,43} The quadriceps provides resistance to knee varus
loading during controlled joint perturbations.\textsuperscript{78,83} Therefore, the quadriceps theoretically play a
role in controlling frontal plane knee loading during walking gait. As such, this mechanism may
be reduced with quadriceps dysfunction (i.e. following ACL-R) and result in greater knee varus motion. An increased varus moment yields greater loading of the medial compartment of the tibiofemoral joint where OA is most prevalent.\textsuperscript{22,41,43,47,74} However, there is no evidence within the ACL-R population to support this claim due to a lack of research.

The quadriceps’ attenuation of load serves as a protective neuromuscular mechanism for the knee by dissipating load over time. A rabbit study simulated impulsive joint loading prior to and after a Botox injection to the knee extensors. Paralyzing these muscles resulted in atrophy and cartilage changes in the paralytic limb that mimicked OA, while the normal limb did not.\textsuperscript{84} Another study had human subjects complete a gait task over a force plate after the femoral nerve was injected with lidocaine, mocking quadriceps dysfunction, which resulted in impulsive loading.\textsuperscript{85} When healthy human knees were injected with saline the peak vertical ground reaction force was increased compared to controls.\textsuperscript{82} This may be transitive to the ACL-R population, but research has not defined the relationship between quadriceps dysfunction and rate of vertical ground reaction force.

**Part II: Alterations in Quadriceps Function following ACL Injury**

Quadriceps weakness after ACL-R is due to arthrogenic muscle inhibition (AMI).\textsuperscript{19,22} AMI is a form of inhibition occuring in the central nervous system (CNS) that impedes voluntary activation of a muscle. After injury, AMI acts as a protective mechanism by limiting joint loading caused by muscle forces. However, this inhibition seems to persistent for several months and/or years,\textsuperscript{19-21} and may lead to aberrant gait biomechanics, cartilage degeneration, and development of OA. After an ACL injury, damage to the ligament also damages mechanoreceptors, resulting in partial de-afferentation. This altered afferent input to the CNS as well as changes in afferent signals due to joint effusion, laxity, pain, or combination thereof
signals joint damage, and the CNS responds via inhibition of the alpha motor neuron pool in the quadriceps musculature (i.e. AMI).\textsuperscript{22}

Quadriceps weakness is common after ACL-R.\textsuperscript{19-21} A literature review, including 39 studies in individuals 6 months post ACL-R, reported average deficits in voluntary quadriceps activation exceeding 20\% with a maximum of 40\%.\textsuperscript{19} One such study revealed patients with ACL-R have a 14.7\% deficit of voluntary quadriceps activation 2 years after reconstruction.\textsuperscript{20} Deficits in voluntary quadriceps activation contribute to muscle weakness and atrophy. Subjects 2-9 years post ACL-R have a 10\% less isokinetic strength when compared to the contralateral healthy limb. However, the time from injury to surgical intervention differed substantially across subjects in this study (6 weeks - 2 years) which could have affected the results.\textsuperscript{21}

Rehabilitation after ACL-R aims to address this quadriceps weakness, but AMI persists long after completion of traditional rehab programs.\textsuperscript{19,22,86,87} A significant negative correlation exists between AMI and quadriceps strength.\textsuperscript{87} The central activation ratio (CAR) reflects the percentage of the motor neuron pool that can be activated voluntarily.\textsuperscript{26} The existing literature varies in reports of quadriceps activation deficits, ranging from 8-45\%.\textsuperscript{22} In subjects 6 or more months post-ACL-R, quadriceps deficits of 10\% or more were found in a majority of studies.\textsuperscript{19} Healthy subjects can voluntarily recruit approximately 96\% of the motor units available within the quadriceps\textsuperscript{88} and a systematic review\textsuperscript{26} found similar results. This review found a decreased CAR in patients with ACL-R (86.5\%), as compared a control (98.3\%). Eight males with AMI after knee injury displayed significantly more quadriceps inhibition in the injured leg (45.6) as compared to the contralateral limb (18.6\%) both before and after conservative rehabilitation, 28.5\% and 10.4\% respectively. After rehabilitation the quadriceps inhibition of the injured leg remained significantly reduced from baseline. A significant difference was also found in the
maximal isometric strength when the injured leg was compared to the uninjured leg before and after rehabilitation (40.5% and 45.5%, respectively).\textsuperscript{87}

**Part III: Kinematic and Kinetic Aberrations following ACL-R**

Quadriceps weakness in patients with ACL-R is theorized to cause aberrant gait biomechanics. During gait, kinematic\textsuperscript{35-38} and kinetic\textsuperscript{22,25,36-39} alterations in the frontal and sagittal planes\textsuperscript{34} are present in patients with ACL-R. Peak knee flexion angle is reduced 6 weeks post ACL-R compared to 12 months post ACL-R\textsuperscript{35} and compared to healthy individuals.\textsuperscript{36,39} Consequently, patients 6 weeks post ACL-R had greater peak knee extension values than the control, but at 12 months after ACL-R there was no difference.\textsuperscript{35} The varus angle was found to be significantly greater in patients post ACL-R when compared to matched controls.\textsuperscript{38} Greater varus motion and loading increase loading of the medial compartment of the tibiofemoral joint which is the most frequently involved in patients with OA. This increase causes a shift in the contact area of the joint, loading portions of the articular cartilage and menisci unfamiliar with stress.\textsuperscript{89} Regarding gait kinetics, the internal knee extension moment is lesser in patients with ACL-R compared to healthy individuals.\textsuperscript{25,36,90} Patients with OA also display a decreased knee extension moment, likely resulting from lesser quadriceps activation which is liked to impulsive loading.\textsuperscript{22,41-44} A group of healthy subjects performed a gait task in which the knee was restricted to 25 degrees of knee flexion, 10 degrees of knee flexion and no restriction. An increased loading rate of the vGRF was noted with the decreased knee flexion condition.\textsuperscript{91} The loading rate of the vGRF is greater in patients with ACL-R when compared to a control or the contralateral limb, during a horizontal hop test.\textsuperscript{92} Similarly, an increased peak vertical loading rate is associated with the gait of patients with OA.\textsuperscript{93} This is defined as the difference in time from initial ground contact until peak vGRF. An increased loading rate yields greater stress upon the articular joint
surfaces and menisci. Patients with OA display this increased loading rate due in part to limited knee flexion.\textsuperscript{42-44} One study showed that subtle changes in gait biomechanics after ACL-R alters the area in which load is distributed; thus, previously unloaded areas receive load leading to premature thinning of cartilage.\textsuperscript{89}

\textbf{Part IV: How Alterations in Gait Influence Osteoarthritis}

Alterations in gait biomechanics potentially lead to OA via two mechanisms: greater loading rate and the shift of contact area.\textsuperscript{22} When the quadriceps is weak it does not absorb impact forces as effectively, thus a greater force is transmitted to the tibiofemoral joint at a greater rate.\textsuperscript{22} Strength training programs have been recommended for patients with OA as they slow the rate of quadriceps strength decline compared to range of motion (ROM) exercises. However, neither intervention prevents the decline in strength.\textsuperscript{94} Quadriceps weakness or dysfunction has been demonstrated prospectively to increase the risk of developing OA.\textsuperscript{86} The extent of joint degradation is dependent upon the impulsive load or the rate of force applied over time.\textsuperscript{32,95} When load magnitude was controlled, rabbits that were loaded impulsively had more significant knee OA changes in the form of cartilagenous fissure, thickening of subchonral bone, and traumatized cartilage. Though this study was specific to patellar rather than tibiofemoral cartilage, the results suggest that loading rate is an important contributor to the degeneration of cartilage.\textsuperscript{32} A similar study focused on the tibiofemoral joint of rabbits and agreed that joint degredation is not load dependent, but rather loading rate dependent.\textsuperscript{95} Ultimately impulsive joint loading contributes to the development of OA.\textsuperscript{32,95}

Joint force distribution is higher in the medial compartment which further augments the development of OA.\textsuperscript{21,79} A shift in joint contact area to a thinner cartilagenous portion can accelerate OA. Alterations in gait biomechanics, whether they are due to pain, inflammation, or
trauma, cause damage when joint loading is altered because it could lead to fissuring or affect the nutrient and waste exchange of cartilage. Reduced knee flexion angles in patients with ACL-R were shown to shift load distribution upon the articular cartilage. An increased varus moment yields loading of the medial compartment of the tibiofemoral joint where OA is most prevalent.

Summary

ACL-R patients are at greater risk for OA through altered biomechanical function during gait. We believe these alterations in gait are due to quad dysfunction stemming from ACL-R. Therefore, the purpose of this study is to examine the relationship between quadriceps function (CAR) and gait biomechanics (peak knee flexion, internal knee extension moment, knee varus angle, and loading rate of the vertical ground reaction force) in patients with ACL-R.
CHAPTER 3: Research Methods

Design and Procedures

This study is part of a larger investigation evaluating the potential contributions of a series of neuromechanical factors to the risks of secondary injury/graft-rupture and osteoarthritis following ACL-R. The overall experimental design involved two testing sessions. The first session consisted of a series of self-report surveys, a collection of a blood sample, a gait biomechanics assessment (utilized in this study), a landing biomechanics assessment, a lower extremity range of motion assessment, and a lower extremity strength assessment. The second session consisted of a quadriceps function assessment (utilized in this study), cortical neuron excitability, spinal neuron excitability assessment, ultrasound imaging of the quadriceps and hamstrings, and a hamstring stiffness assessment.

This study utilized a cross-sectional design which examined the relationship between quadriceps function and gait biomechanics in individuals who have undergone ACL-R (n = 31). Subjects reported to the Sports Medicine and Neuromuscular Research Laboratory on the UNC-CH campus for two testing sessions that lasted approximately 2½ hours each separated by approximately 1 week. In the first session, subjects completed the Gait Biomechanics assessment described below. In the second session, subjects completed the Quadriceps Function assessment described below. Upon reporting to the laboratory, subjects first performed a 5-minute “warm up” on a stationary cycle ergometer at a self-selected pace followed by the aforementioned assessments. Testing was performed on both limbs (ACLR and healthy/uninjured).
Subjects

Thirty-one physically active (at least 30 minutes of physical activity 3x/week) individuals who have undergone unilateral ACL-R using either a patellar tendon or hamstring tendon graft were recruited (Table 1). Subjects were required to be at least 6 months post ACL-R and have no history of ACL graft rupture or revision surgery, neurological disorder (stroke, multiple sclerosis, amyotrophic lateral sclerosis/Lou Gehrig’s Disease, diabetic neuropathy, epilepsy, traumatic brain injury resulting in a loss of consciousness, concussion within the 6 months prior to participation, cranial neural surgery and balance disorders), injury to either leg within 6 months prior to participation, osteoarthritis or current symptoms related to knee osteoarthritis (e.g. pain, swelling, stiffness), current pregnancy, pacemaker or other implantable electronic device, history of cardiac arrhythmia, psychiatric disorder, cancer in the brain or thigh musculature, any cardiac condition, or any implanted metal objects. Power analysis indicated that 22 subjects would provide a priori statistical power of 0.80 to evaluate the relationship between quadriceps strength and gait biomechanics (knee flexion angle and knee extensor moment) in patients with ACL-R. However, subjects in our study were evaluated upon the relationship between CAR and gait biomechanics (peak knee flexion angle, peak knee extensor moment, peak knee varus angle and loading rate of the vertical ground reaction force. Therefore, we will include 9 additional subjects to increase subject size and ensure adequate power.

Session 1: Gait Biomechanics Assessment

Using double-sided tape, 25 reflective markers were placed bilaterally on the acromion process, body of the sternum, anterior superior iliac spine, greater trochanter, anterior thigh, lateral and medial femoral epicondyles, anterior shank, lateral and medial malleoli, calcaneus, first metatarsal head, fifth metatarsal head and L4-L5 joint space. Gait kinematics and kinetics
were sampled at 120 Hz and 1,200 Hz, respectively, using a 7-camera optical motion capture system (Vicon Nexus) interfaced with three force plates (Bertec Corp, Columbus, Ohio) using Vicon Nexus v1.4.1 motion capture software (Vicon Motion Systems). Subjects walked forward along a 3m (~10 ft) walkway at a comfortable, self-selected “fast” speed while biomechanical data were collected. The force plates were staggered such that one gait trial provided kinematic and kinetic data for both limbs. At least 5 practice trials were performed to determine the average preferred speed and ensure subjects could consistently strike the force plate mounted in the walkway with the test limb without noticeably altering their gait (i.e. “aiming” for the force plate). Gait speed was monitored via an infrared timing system to ensure each trial was within ±5% of the preferred speed. Subjects performed 5 valid trials from which gait biomechanical variables were averaged for statistical analysis.

**Session 2: Quadriceps Function Assessment**

Quadriceps function was quantified via the central activation ratio (CAR) during maximal isometric knee extension. An electrical stimulus was superimposed to the maximal voluntary contraction which activated all available motor units. CAR was then calculated as the ratio of peak voluntary torque to peak torque resulting from the electrical stimulus, and represented the percentage of the motor unit pool that was activated voluntarily (Figure 3 in Appendix A).

Subjects were seated on a dynamometer (HUMAC NORM, CSMi, Stoughton, MA) with the knee in 90° of flexion. Straps were used to secure the torso, thigh, and leg to the device. The moveable arm of the dynamometer was fixed in place making the contraction mode isometric. Two 7x13cm Dura Stick II © adhesive stimulating electrodes (Chattanooga Group, Hixon, TN) were placed on the anterior thigh over the quadriceps muscle (Figure 1 in Appendix A). Subjects
were asked to extend the knee maximally and as quickly as possible against the dynamometer. Subjects performed 3 trials and the average value was recorded as the maximal voluntary isometric contraction (MVIC). The subject was then introduced to the electrical stimulus by being asked to produce 25%, 50%, and 75% of their perceived maximal contraction, followed by the same percentage of stimulus, serving as both a warm up and an acclimatization mechanism. This electrical stimulus consisted of a 10 pulse train, pulse duration of 0.6ms, delivered at a frequency of 100Hz, and an intensity of 125V. The electrical stimulator (Grass Telefactor model SK48) was isolated from fluctuations in the building’s electrical power supply via a stimulus isolation unit (Grass Telefactor model SIU5). The subject was then asked to produce a maximal quadriceps contraction which was followed by the stimulus. A computer screen in the subject’s view (Figure 1 in Appendix A) displayed a target consisting of horizontal lines corresponding with the MVIC torque and 10% above the MVIC torque (Figure 2 in Appendix A). Subjects were asked attempt to reach the upper margin of the target range, though the lower margin represented their MVIC. Once subjects exceeded their MVIC force the computer software (LabVIEW) automatically generated an output signal when the torque decreased by 1N/m that caused the stimulator to generate the stimulus. This approach ensured that a maximal contraction occurred and permitted an automated stimulus. The subjects performed at least one trial per limb.

Data Reduction

Joint torque and moment data during gait trials were normalized to the product of weight and height (N*m). The MVIC was calculated as the maximum torque value 60ms before the stimulus was given and was averaged if more than one trial was completed per subject. The potential MVIC (pMVIC) was calculated as the maximum torque value after the stimulus was
given and was averaged if more than one trial was completed per subject. CAR was calculated as the ratio of MVIC to pMVIC. The kinematic data were low pass filtered at 10 Hz and kinetic data at 75 Hz. Grood and Suntay angles were calculated as motion of the shank measured relative to the thigh. All kinematic and kinetic variables were identified during the first 50% of the stance phase, defined as the interval between initial ground contact and toe off (vertical ground reaction force >20N and <20N, respectively). Peak knee flexion angle was calculated as the maximal sagittal plane angle. Peak knee varus angle was calculated as the maximum frontal plane angle. Peak internal knee extension moment was calculated as the minimum sagittal plane moment and normalized to the product of subject weight (N) and height (m). The peak vGRF was calculated as the maximum vGRF, and the loading rate of the peak vertical ground reaction force was calculated by dividing the peak vGRF by the time between IGC and peak vGRF (N/s) and normalized to body weight.

**Statistical Analyses**

One-tailed Pearson product-moment correlation analyses and Shapiro-Wilk tests of normality were used to evaluate the relationships between CAR and peak knee flexion angle, internal knee extension moment, peak knee varus angle, and loading rate of the vertical ground reaction force (α = 0.05). Due to the fact that gait biomechanics are influenced by gait speed, partial correlations were computed after controlling for the variance attributable to gait speed. using IBM SPSS Statistics for Windows, Version 22.0.
CHAPTER 4: Results

Shapiro-Wilk test of normality revealed data for our subjects (n=31) were normal for peak knee flexion angle (p=0.999), peak knee varus (p=0.257), and peak knee extension moment (p=0.334), but not normal for CAR (p=0.000) and loading rate of the vGRF (p=0.011). Box plots revealed subjects 1, 19, and 38 to be outliers within the CAR data (0.73, 0.53, and 0.72) and subject 14 within the loading rate of the vGRF data (13.74). When these subjects were excluded, data became normal for CAR and across all variables (n=25). Subject characteristics and descriptive statistics are provided in Table 1. The means of kinetic and kinematic data, as well as CAR are presented in Table 2. One-tailed Pearson product-moment correlations revealed no significant findings between CAR and peak knee extension moment (p=0.159, r=-0.212; Figure 3), peak knee flexion angle (p=0.144, r=0.227; Figure 4), peak knee varus angle (p=0.092, r=0.281; Figure 5), as well as the loading rate of the vGRF (p=0.187, r=0.190; Figure 6). When the time from injury was controlled there were no significant findings between CAR and peak knee flexion angle (p=0.175; r=0.209), peak knee varus angle (p=0.187; r=0.199), loading rate of the vGRF (p=0.186; r=0.200), or peak knee extension moment (p=0.226; r=-0.169). When the time from surgery was controlled there were no significant findings between CAR and peak knee flexion angle (p=0.156; r=0.232), peak knee varus angle (p=0.212; r=0.184), loading rate of the vGRF (p=0.219; r=0.179), or peak knee extension moment (p=0.208; r=-0.188). When subjects without PT grafts were excluded (n=17) there were no significant findings between CAR and
peak knee flexion angle (p=0.442; r=-0.040), peak knee varus angle (p=0.441; r=0.041), loading rate of the vGRF (p=0.276; r=0.161), or peak knee extension moment (p=0.188; r=-0.238). When subjects with a CAR>0.95 were excluded (n=16), statistical significance was observed between CAR and peak knee flexion angle (p=0.015, r=0.561; Figure 7). When subjects without PT grafts and CAR>0.95 were excluded (n=11), statistical significance was observed between CAR and peak knee extension moment (p=0.022, r=-0.647; Figure 8).

Time from injury and time since surgery did not affect the data as it did not change the significance of the findings. However, after excluding subjects with a CAR <0.95, a moderate positive relationship was observed between CAR and peak knee flexion angle. The observed power for this correlation was 0.76, and an additional 3 subjects would be needed to achieve power of 0.80 with an alpha level of 0.05. Furthermore, subjects who received a PT graft and displayed a CAR <0.95, a moderate negative relationship was observed between CAR and the peak knee extension moment.
CHAPTER 5: Discussion

Summary

The primary findings of this study were that a lower quadriceps CAR is associated with a smaller peak knee flexion angle and a smaller peak internal knee extension moment during the loading phase of walking gait. No relationships were observed between CAR and peak knee varus angle or loading rate of the vGRF. Our findings indicate that quadriceps dysfunction influences gait biomechanics linked to OA development given the relationships between CAR, peak extension moment, and peak flexion angle. This suggests that improving quadriceps function may improve gait biomechanics in a manner that would reduce the risk of OA.

CAR & Peak Knee Flexion Angle

The observed power of the correlation between peak knee flexion angle and CAR was 0.76, and post hoc sample size calculations indicated that 3 additional subjects would be required to obtain power of 0.80 with an alpha level of 0.05. CAR explained only 31% of the variance in the peak knee flexion angle in our study. For example, the hamstrings could influence knee flexion angle as they co-contract with the quadriceps, in a healthy knee, during the loading phase of gait. Multiple studies have shown that patients with ACL-D, when compared with healthy controls, have greater activation of the hamstrings prior to heel strike. If also present following ACL-R, this heightened hamstring co-contraction could confound our findings, as greater hamstring activity may decrease the peak knee flexion angle. Berchuck et al. compared patients with ACL-R to healthy controls and found a lesser knee flexion angle occurring at the time of maximum knee flexion moment. However, this is an instantaneous measure. As such,
knee flexion excursion might be a better variable to examine the total motion regulated by the quadriceps. Knee flexion excursion (i.e. the range between maximum and minimum knee flexion angles during stance phase\textsuperscript{108}) reveals the extent of the motion occurring in the sagittal plane of the knee during the loading phase of gait. It has been reported that individuals with OA\textsuperscript{22,41-44,109-111} and ACL-R\textsuperscript{110} display less knee flexion excursion during gait compared to matched, healthy controls. Overall, this highlights the potential of the hamstrings to decrease the peak knee flexion angle and implicates the need to address the co-contraction in the clinical setting after knee injury. Clinicians should prioritize hamstring activation and co-contraction goals during post ACL-R rehabilitation.

**CAR & Peak Internal Knee Extension Moment**

A lower quadriceps CAR was correlated with a smaller peak internal knee extension moment. The observed power of the correlation between peak internal knee extension moment and CAR was 0.70, and post hoc sample size calculations indicated that one additional subject would be required to obtain power of 0.80 with an alpha level of 0.05. CAR explained only 41\% of the variance in the peak internal knee extension moment in our study. Patients with ACL-R average 20-40\% deficits in voluntary quadriceps activation,\textsuperscript{26} more commonly known as athrogenic muscle inhibition (AMI), or the inability to voluntarily activate muscle fibers needed for muscular contraction. AMI is commonly measured via the central activation ratio (CAR),\textsuperscript{22,23} which represents the portion of the quadriceps motor neuron pool that can be activated voluntarily.\textsuperscript{24} The internal knee extension moment reflects, in part, the torque produced by the quadriceps. Lesser internal knee extension moment during the stance phase of gait has been reported in individuals with ACL-R compared to matched healthy controls.\textsuperscript{25,36,90} Individuals with knee OA also display a lesser knee extension moment compared to controls, which is linked
to impulsive loading.\textsuperscript{42,43} Impulse describes the relationship of force applied over time in which greater impulse could mean more force applied over a shorter period of time, or the same force applied over a long period of time. Greater impulsive loading contributes to the development and progression of OA.\textsuperscript{31,32} Lewek et al.\textsuperscript{36} divided ACL-R subjects into weak and strong groups, as defined by a quadriceps index (involved volitional force/uninvolved volitional force x100) of 80\% or less and 90\% or more, respectively. Only the weak ACL-R group displayed a lesser internal knee extension moment compared to the healthy subjects\textsuperscript{36} CAR explained only 41\% of the variance in the peak internal knee extension moment in our study. Similar to the findings of Lewek et al.,\textsuperscript{36} the unexplained variance may be attributable to a limited range of CAR in the sample. While the CAR of our subjects ranged from 0.53 to 0.97, all but 10 of our subjects displayed a CAR greater than 0.90. Also, other factors (i.e. co-contraction of the hamstrings and static structures) may have contributed to the magnitude of the knee extension moment.\textsuperscript{100,101} Kellis et al.\textsuperscript{100} described the total knee joint moment as a result of both the agonist (quadriceps) and the antagonist (hamstrings) muscles. Also, ligaments within the knee provide static support and limit the knee during extension.\textsuperscript{66,67} The contribution of the hamstrings and ligaments to the net knee joint moment may explain the low correlation coefficient between quadriceps CAR and the internal knee extension moment. Arnason et al.\textsuperscript{102} reported greater activation of the hamstrings bilaterally in individuals with ACL-R compared to controls during concentric and eccentric exercises. Individuals with ACL-D display heightened hamstring activity during gait to promote joint stability in the absence of the native ACL.\textsuperscript{103} This same phenomenon is potentially present following ACL-R, which would reduce the influence of quadriceps CAR on the net knee moment. Though CAR is a good measure the efficiency of quadriceps activation,\textsuperscript{23,26} it may not reflect activity of the quadriceps during gait. For example, the primary function of the
quadriiceps during stance phase is eccentric contraction, but CAR does not necessarily reflect the manner in which an individual activates the quadriceps during gait. Clinicians must be cautious of using maximal quadriceps contraction force as a marker for return to play and rehabilitation progression, as our subjects still displayed deficits in both quadriceps activation and with a lesser knee extension moment during gait.

**CAR & Peak Knee Varus Angle**

Knee varus is a frontal plane motion in which the primary mechanisms of dynamic control, co-contraction of the quadriceps and the hamstrings only support 14% of a pure external varus moment. Even though the quadriceps and hamstrings are the primary dynamic restraints to a varus force, they have a limited capacity for controlling frontal plane motion and loading. As such, the lack of a strong relationship between quadriceps function and peak varus angle is not surprising. Hurwitz et al. reported that medial compartment OA was a predictor of peak knee varus moment, potentially suggesting that OA exacerbates knee varus motion as cartilage gradually degenerates rather than excessive knee varus motion leading to OA. Greater varus motion increases loading of the medial compartment of the tibiofemoral joint, possibly linking increased varus post-injury to the perpetuation of OA. Individuals with ACL-R can develop OA as early as 5 years following injury. However, our subjects were approximately 2 years post ACL-R. As such, cartilage thickness in the medial compartment and varus motion may have been within normal limits rather than excessive, which could explain why frontal plane motion was not associated with quadriceps function. The activation ratio of the vastus lateralis and medialis may be a better indicator of quadriceps function and knee varus. A greater ratio of lateral to medial contraction of the quadriceps may contribute to greater knee varus. Alterations in activation of the vastus lateralis compared to the medialis have been found in patients with
ACL-D\textsuperscript{104-106} and OA.\textsuperscript{93} It is important that these potential alterations be addressed by rehabilitation due to the increased risk of developing medial compartment OA. Furthermore, future research is necessary to evaluate the influence of quadriceps and hamstrings co-contraction on frontal plane knee motion and the development and perpetuation of knee OA following ACLR.

**CAR & Peak vGRF linear loading rate**

CAR quantifies the portion of the quadriceps motor neuron pool that can be activated voluntarily\textsuperscript{24} and represents the level of AMI.\textsuperscript{22,23} The quadriceps is not maximally active during gait.\textsuperscript{107} As such, CAR may have a limited influence on loading rates and other gait biomechanics variables during gait due to the fact that it is derived from maximal contraction. Other indices of quadriceps function such as power or rate of force development (RFD) at submaximal levels may provide a better reference for quadriceps activity during gait. Maffiuetti et al.\textsuperscript{108} suggested that RFD, compared to CAR, could be a better predictor of functional disabilities, such as walking. The load acceptance stage of gait is relatively short, taking only a portion of a second, making the function of the quadriceps to attenuate joint loading of the utmost importance. As such, a measure such as RFD or power that evaluates how rapidly force can be produced may be a better indicator of quadriceps function than CAR. High-velocity training focused on improving the speed at which weight is moved may be an effective intervention for improving quadriceps function relative to walking gait.\textsuperscript{109} Sayers et al\textsuperscript{110} found that individuals with knee OA completing a high-velocity training protocol had increased knee extension power but did not differ in strength when compared to a low-velocity training. This finding is important as clinical guidelines for return to play use strength values, not power or the maximal voluntary capacity of the quadriceps (i.e. CAR). Additionally, the vGRF loading rate is not specific to the knee joint.
Mundermann et al.\textsuperscript{43} observed greater loading rates in individuals with OA compared to healthy matched controls that was accompanied by greater joint loading at the ankle, hip, and knee of 64\%, 59\%, and 56\%, respectively. As such, the loading rates in our subjects were potentially influenced by kinematic and kinetic factors at the hip and knee that masked the influence of quadriceps dysfunction.

\textbf{Limitations}

The main limitation of this study is the cross-sectional design. We were unable to ensure that quadriceps deficits were not present pre-surgery or injury. Second, a bigger sample size is needed to observe substantial power to correlate peak knee flexion angle and CAR, as our observed power of 0.52 merely approached statistical significance (p = 0.052). Also, our findings are limited by the function of the quadriceps during gait, an eccentric, submaximal contraction compared to CAR, which is a concentric and maximal contraction. Furthermore, the CAR of our subjects may have been too high to observe a relationship with gait biomechanics. Lastly, we did not evaluate activity of the hamstrings during gait, thus we cannot determine the role of co-activation on our results.

Further research is warranted to define many unknown relationships and find the best measure of quadriceps function. First, the relationship of the hamstrings after ACL-R on gait biomechanics should be further explored. Second, the relationship between knee flexion excursion (i.e. the range between maximum and minimum knee flexion angles during stance phase\textsuperscript{111}) and gait biomechanics in ACL-R should be evaluated, as it has been reported to be lesser in individuals with OA\textsuperscript{22,41-44,112-114} and ACL-R.\textsuperscript{113} Third, the activation ratio of the vastus lateralis relative to the vastus medialis may shed light on the role of the quadriceps in controlling frontal plane knee motion during walking gait. A greater ratio of lateral to medial contraction of
the quadriceps may contribute to greater knee varus. Alterations in activation of the vastus lateralis compared to the medialis have been found in patients with ACL-D\textsuperscript{104-106} and OA.\textsuperscript{93} Fourth, the rate of force development (RFD) at submaximal levels may provide a better reference for quadriceps activity during gait.\textsuperscript{108} Fifth, evaluating the influence of quadriceps and hamstrings co-contraction on frontal plane knee motion and the development and perpetuation of knee OA following ACLR is warranted.

Conclusions

Our findings suggest that quadriceps dysfunction assessed via CAR is associated with smaller peak knee flexion angle and peak internal knee extensor moment. Clinical implications include the need to establish better return to play criteria. All of our subjects had been cleared to return to activity, yet they still displayed quadriceps deficits that influenced gait biomechanics that have been linked to development of knee OA. These findings also suggest that continued rehabilitation following return to play after ACL-R may be warranted to improve quadriceps function and reduce the risk of developing knee OA. Further research using other variables (RFD, knee flexion excursion) is needed to confirm the influence of quadriceps dysfunction on these gait abnormalities in individuals with ACL-R. Also, the role of hamstring co-contraction of the on gait biomechanics in individuals with ACL-R needs to be further examined.
APPENDIX A: FIGURES

Figure 1: HUMAC NORM subject set up. Two 7x13 Dura Stick II © adhesive stimulating electrodes connect a stimulus isolation unit and an electrical stimulator to the subject.
Figure 2: CAR Computer program with feedback lines.
Figure 3: Relationship between CAR and Peak Internal Knee Extension Moment (N\*m)
Figure 4: Relationship between CAR and Peak Knee Flexion Angle (°)

Relationship between CAR and Peak Knee Flexion Angle

- Peak Knee Flexion Angle (°)
- CAR

![Graph showing the relationship between CAR and Peak Knee Flexion Angle](image-url)
Figure 5: Relationship between CAR and Peak Knee Varus Angle (°)

Relationship between CAR and the Peak Knee Varus Angle

-5 -4 -3 -2 -1 0 1 2 3 4 5 6

Peak Knee Varus Angle (°)

0.8 0.82 0.84 0.86 0.88 0.9 0.92 0.94 0.96 0.98 1

CAR
Figure 6: Relationship between CAR and Peak Loading Rate of the vGRF (N/s)
Figure 7: Relationship between CAR and Peak Knee Flexion Angle (°)
Figure 8: Relationship between CAR and Peak Internal Knee Extension Moment (N*m)

Relationship between CAR and Peak Internal Knee Extension Moment

- CAR
- Peak Internal Knee Extension Moment (N*m)
Table 1. Subject Demographics

<table>
<thead>
<tr>
<th></th>
<th>All Subjects</th>
<th>n=25 (6 males, 19 females)</th>
<th>Subjects with a CAR&lt;0.95</th>
<th>n=16 (5 males, 11 females)</th>
<th>Subjects with PT, CAR&lt;0.95</th>
<th>n=11 (2 males, 9 females)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PT Graft</td>
<td>18</td>
<td></td>
<td>11</td>
<td></td>
<td>11</td>
<td></td>
</tr>
<tr>
<td>HT Graft</td>
<td>7</td>
<td></td>
<td>4</td>
<td></td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Meniscus Injury</td>
<td>11</td>
<td></td>
<td>7</td>
<td></td>
<td>4</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
<td>22±3</td>
<td></td>
<td>23±3</td>
<td></td>
<td>22±3</td>
<td></td>
</tr>
<tr>
<td>Height (m)</td>
<td>166.6±10.5</td>
<td></td>
<td>168.0±11.7</td>
<td></td>
<td>165.1±10.1</td>
<td></td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>67.7±13.3</td>
<td></td>
<td>68.9±15.9</td>
<td></td>
<td>64.6±9.76</td>
<td></td>
</tr>
<tr>
<td>Time from Injury (months)</td>
<td>58±45</td>
<td></td>
<td>64±40</td>
<td></td>
<td>73±40</td>
<td></td>
</tr>
<tr>
<td>Time from Surgery (months)</td>
<td>57±45</td>
<td></td>
<td>63±38</td>
<td></td>
<td>69±38</td>
<td></td>
</tr>
</tbody>
</table>

Values are mean±SD
Table 2: Descriptive Statistics of CAR, Gait Kinematics & Kinetics

All Subjects (n=25)

<table>
<thead>
<tr>
<th></th>
<th>Peak Loading Rate of vGRF (N/s)</th>
<th>Peak Knee Flexion Angle (°)</th>
<th>Peak Internal Knee Extensor Moment (N*m)</th>
<th>Peak Knee Varus Angle (°)</th>
<th>Gait Speed (m/s)</th>
<th>CAR</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>7.25±1.39</td>
<td>12.74±5.82</td>
<td>-0.04±0.02</td>
<td>0.25±2.57</td>
<td>1.11±0.11</td>
<td>0.92±0.05</td>
</tr>
</tbody>
</table>

Subjects with a CAR<0.95 (n=16)

<table>
<thead>
<tr>
<th></th>
<th>Peak Loading Rate of vGRF (N/s)</th>
<th>Peak Knee Flexion Angle (°)</th>
<th>Peak Internal Knee Extensor Moment (N*m)</th>
<th>Peak Knee Varus Angle (°)</th>
<th>Gait Speed (m/s)</th>
<th>CAR</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>7.31±1.45</td>
<td>12.74±6.4*</td>
<td>-0.04±0.02</td>
<td>-0.05±2.57</td>
<td>1.15±0.11</td>
<td>0.90±0.03</td>
</tr>
</tbody>
</table>

Subjects with PT graft and CAR<0.95 (n=11)

<table>
<thead>
<tr>
<th></th>
<th>Peak Loading Rate of vGRF (N/s)</th>
<th>Peak Knee Flexion Angle (°)</th>
<th>Peak Internal Knee Extensor Moment (N*m)</th>
<th>Peak Knee Varus Angle (°)</th>
<th>Gait Speed (m/s)</th>
<th>CAR</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>7.33±1.55</td>
<td>12.92±6.8</td>
<td>-0.04±0.02</td>
<td>0.61±2.35</td>
<td>1.13±0.12</td>
<td>0.90±0.02</td>
</tr>
</tbody>
</table>

Values are mean±SD
*p<0.05
Table 3: Relationship between CAR and gait biomechanics

<table>
<thead>
<tr>
<th></th>
<th><strong>All Subjects (n=25)</strong></th>
<th>r</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Loading Rate of vGRF (N/s)</td>
<td>0.190</td>
<td>0.187</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Flexion Angle (°)</td>
<td>0.227</td>
<td>0.144</td>
<td></td>
</tr>
<tr>
<td>Peak Internal Knee Extensor Moment (N*m)</td>
<td>-0.212</td>
<td>0.159</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Varus Angle (°)</td>
<td>0.281</td>
<td>0.092</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th><strong>Subjects with a CAR&lt;0.95 (n=16)</strong></th>
<th>r</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Loading Rate of vGRF (N/s)</td>
<td>0.049</td>
<td>0.431</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Flexion Angle (°)</td>
<td>0.561</td>
<td>0.015*</td>
<td></td>
</tr>
<tr>
<td>Peak Internal Knee Extensor Moment (N*m)</td>
<td>-0.316</td>
<td>0.125</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Varus Angle (°)</td>
<td>0.307</td>
<td>0.133</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th><strong>Subjects with PT graft and CAR&lt;0.95 (n=11)</strong></th>
<th>r</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Loading Rate of vGRF (N/s)</td>
<td>0.037</td>
<td>0.460</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Flexion Angle (°)</td>
<td>0.388</td>
<td>0.134</td>
<td></td>
</tr>
<tr>
<td>Peak Internal Knee Extensor Moment (N*m)</td>
<td>-0.547</td>
<td>0.022*</td>
<td></td>
</tr>
<tr>
<td>Peak Knee Varus Angle (°)</td>
<td>-0.067</td>
<td>0.427</td>
<td></td>
</tr>
</tbody>
</table>

*Values are one-tailed partial correlation coefficients (r) controlled for gait speed
*p<0.05
REFERENCES


20. Urbach D, Nebelung W, Becker R, Awiszus F. Effects of reconstruction of the anterior cruciate ligament on voluntary activation of quadriceps femoris a prospective twitch


