The Impact of Anterior Cruciate Ligament Reconstructive Surgery on Neuromotor Function

Cortney Noel Armitano

Old Dominion University, cnarmitano@gmail.com

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THE IMPACT OF ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTIVE SURGERY ON NEUROMOTOR FUNCTION

by

Cortney Noel Armitano
B.S. May 2010, Campbell University
M.S. May 2013, University of Rhode Island

A Dissertation Submitted to the Faculty of Old Dominion University in Partial Fulfillment of the Requirements for the Degree of

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Approved by:

Daniel M. Russell (Co-Director)
Steven Morrison (Co-Director)
Johanna M. Hoch (Member)
Hunter D. Bennett (Member)
ABSTRACT

THE IMPACT OF ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTIVE SURGERY ON NEUROMOTOR FUNCTION

Cortney Noel Armitano
Old Dominion University, 2019
Co-Directors: Dr. Daniel M. Russell
Dr. Steven Morrison

The purpose of this dissertation was to examine the systemic neuromechanical implications in individuals who have had an ACL reconstruction (ACLR) compared to healthy controls. The specific aims addressed were to: 1) examine differences in inter-limb coordination during walking at different speeds, 2) examine differences in trunk, neck and head acceleration during gait, and 3) investigate whether the reaction time responses assessed during stepping are negatively affected by ACLR.

The findings of study 1 revealed that maximal coordination stability was achieved when walking at the person’s preferred gait speed. However, individuals with a previous ACLR exhibited reduced coordination stability between the knees, indicative of decreased inter-limb coupling. Further, individuals within the ACLR group who deviated the most from anti-phase coordination during walking also demonstrated lower coordination stability. These findings could contribute to the secondary issues related to ACL damage.

Study two examined differences in upper body accelerations during gait, revealing that the ACLR group had a diminished capacity to attenuate gait-related oscillations from the trunk to the head. Further, the vertical acceleration signals for the ACLR individuals were more complex, indicating that they had a reduced ability to optimally accelerations during walking. These results demonstrate the impact of ACL damage is not localized but is more systemic and can negatively impact postural control.
The third study assessed how ACLR would impact of general neuromotor function and stepping reaction times. The findings revealed that ACLR individuals had slower reaction times during stepping compared to healthy controls. In contrast to the slowing of reaction time (under postural conditions), there were no changes across any other neuromotor/mechanical measures. This result indicates that the ACLR group had a reduced ability to respond to unexpected stimuli.

Overall, the results of this investigation suggest that ACL damage has a wide-spread impact as it not simply localized to the injured knee. The collective results from these studies show changes in movement strategy prioritization in those with an ACLR. These novel findings provide an alternate perspective and may change the ways in which clinicians and healthcare providers assess individuals who have had ACL reconstructive surgery.
ACKNOWLEDGEMENTS

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“A tree with strong roots laughs at storms” - Malay Proverb

Finally, I would not be the person I am today without the love and support of my family. To my parents, you have always encouraged me to be a person who dreams big, who doesn’t lose sight of how far I have come, and who does it their own way regardless. Thank you for always believing I could do whatever I set my mind to. To my siblings, no matter where this
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CHAPTER I

GENERAL INTRODUCTION

1.1. INTRODUCTION

In the United States there are more than 7.6 million individuals participating in high school sports (Rosenthal et al. 2014). At the collegiate level, the National Collegiate Athletic Association (NCAA) reports there are roughly 470,000 athletes across the country competing at Division I, II, and II level sports with a 70% increase over the past 20 years (Irick 2014). With the enactment of Title IX, there are more females participating in sports than ever before (Acosta and Carpenter 2014). The increase in collegiate female athletes over the past couple of decades has been roughly 200% where over the same timeframe male participation has increased roughly 25% (Irick 2014). With the rise of sport participation at the high school and collegiate levels there has also been an increase in the number of anterior cruciate ligament (ACL) injuries (Agel et al. 2016). The ACL is the primary stabilizing ligament in the knee joint, therefore reconstruction is often recommended to regain knee stability (Kaplan 2011). Despite high rates of successful surgical and rehabilitative outcomes, 1 of 17 individuals will incur a second ACL injury within two years following the initial ACL reconstruction (ACLR) (Wright et al. 2007). In addition, within 10-15 years 60% of ACL reconstructed individuals will have developed post-traumatic osteoarthritis (Roos et al. 1995; Lohmander et al. 2004; Lohmander et al. 2007; Øiestad et al. 2010). While these numbers seem rather large, there is still a good portion of ACL reconstructed people who don’t have these residual effects of the surgery. Because of this, it is believed that there are mechanical factors contributing to the secondary conditions as a result of having an ACLR.

Previous research has focused on trying to understand the factors contributing to residual risk factors associated with reconstructive surgery. When looking at the mechanical factors, however, there is no clear answer to why there are some who are at an increased rate of re-injury and why the majority experience joint degeneration. Measures of strength, proprioception and
sensation, balance, and spatiotemporal parameters all appear to return to normal function following reconstructive surgery (Hart et al. 2010a; Kaur et al. 2016). Kinetically and kinematically, there is no discernable difference in knee function compared to either the contralateral limb or healthy controls. While walking, individuals with an ACLR appear to have some increased joint moments in the knee and hip, however these changes tend to disappear within a couple of years post-surgery (Bulgheroni et al. 1997; Bush-Joseph et al. 2001a; Karimi et al. 2013). The absence of a natural ACL may limit a person’s ability to adapt to the inevitable variability innate to everyday walking, thus placing them at an increased risk of degeneration (Moraiti et al. 2010). Some researchers have adopted non-traditional biomechanical measures to assess ACL reconstructed individuals and have examined movement variability. These researchers have observed is ACL reconstructed gait exhibits more variable and unpredictable knee flexion-extension movement patterns (Moraiti et al. 2009; Moraiti et al. 2010). Further, ACL reconstructed individuals seem to adapt altered gait strategies, illustrating a loss of coordination around the knee joint during gait (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011). Decker and colleagues suggest that the changes in gait variability in ACL reconstructed individuals is due to a lack of proprioceptive information that would be typically transmitted with an intact ACL (Decker et al. 2011). Without this sensory feedback, ACL reconstructed individuals may not be able to make the appropriate neuromotor adjustments necessary during gait. The absence of a natural ACL may disrupt the system in a more systemic manner than has been previously considered. The impact of ACLR may manifest in three different ways throughout the body: 1) in the acceleration dynamics of the upper body during gait, 2) in the coupling strength between limbs during gait, and 3) the ability of the neuromotor system to respond to stimuli. Having a more comprehensive grasp on how ACLR impacts the body as a whole is imperative for understanding underlying mechanical changes post-surgery and allow us to develop strategies that reduce the residual symptoms of ACLR.

1.2. BACKGROUND

The ACL is the most commonly injured ligament in the knee joint, accounting for roughly 200,000 injuries per year in the United States (Prodromos et al. 2007; Spindler and Wright 2008). The anterior cruciate ligament (ACL) is a thick structure of connective tissue that
acts as the primary stabilizing ligament of the knee. Structurally, the ACL is comprised of two bundles, the anteromedial and the posterolateral bundles (Duthon et al. 2006). The ACL originates from the lateral femoral condyle and inserts on the tibial intercondylar spine (Girgis et al. 1975; Kopf et al. 2009). Based on the structural design, the ACL stabilizes the knee joint by resisting tibial translation and rotational stress on the knee (Butler et al. 1980; Zantop et al. 2007). Mechanoreceptors within the ACL provide proprioceptive information that may have an influence on gait and postural adaptations (Adachi et al. 2002). When activated, these receptors elicit a motor response of the muscles surrounding the knee joint (Duthon et al. 2006). This activation is referred to as the ACL reflex (Krogsgaard et al. 2002; Duthon et al. 2006). The ACL is not only considered the primary stabilizing ligament of the knee, it also provides afferent feedback necessary for normal knee function during gait (Fremerey et al. 2000; Krogsgaard et al. 2002). Injury to the ACL (i.e. ACL tear) often results in destabilization of the knee joint, and there are a couple of strategies to manage ACL injury (Kaplan 2011). Ultimately, reconstruction of the ACL is recommended to regain stability of the knee joint and allow individuals to return to an active lifestyle (Kvist 2004; Monk et al. 2014).

The aim of ACL reconstructive surgery is to maximize knee joint stability and functionality to allow individuals to return to pre-injury activity levels (Anderson et al. 2016). The outcomes of this surgery and subsequent rehabilitation are highly successful, indicated by the rate of return to high levels of athletic activity (Deehan et al. 2000). Despite the success of surgery and the return to sports, there are some significant consequences of this surgery that observably impact the quality of life of ACL reconstructed individuals. The overall cost of an ACLR is $38,121, in the US this adds up to $2.78 billion per year from ACL reconstructive surgery and subsequent treatment (Mather III et al. 2013). People who have had reconstructive surgery report a decreased quality of life and, despite high rates of successful outcomes, only 63% of individuals return to their pre-injury level of sport participation (Ardern et al. 2011). The rate of re-injury to that reconstructed ACL or injury to the contralateral ACL increases tremendously after reconstruction. The risk of initially injuring an ACL occurs 1 in 80 people, the rate of re-injury or injury to the contralateral ACL post-reconstruction is 1 in 17 within the first two years after surgery (Wright et al. 2007; Paterno et al. 2010). In addition, up to 13% of individuals with an ACLR will develop symptoms of post-traumatic osteoarthritis (PTOA) 10-15
years after surgery, with the number increasing to roughly 50% if the injury was not isolated to just the ACL (Øiestad et al. 2010). The annual cost of the development of PTOA after ACLR is $2.78 billion in the US alone (Mather III et al. 2013). Persons with an ACLR have shown a 10-fold increased risk of developing PTOA compared to those who have never sustained a knee injury (Roos et al. 1998; Gillquist and Messner 1999). Besides the immediate financial burden, the increased risk of subsequent ACL injury and the toll PTOA takes on younger adults and the associated costs of this secondary sequelae warrants further investigation. Because the consequences of ACLR do not impact every person who undergoes ACL reconstructive surgery, researchers have deduced that mechanical factors are what influence re-injury rate or the likelihood of developing PTOA. Therefore, research has investigated possible neuromotor or mechanical issues that lead the post-reconstruction risk factors. This has led to investigations into changes in neural pathways, strength, proprioception and sensation, balance, spatiotemporal measures, kinematics, and kinetics as plausible influencing factors in the aftereffects of having an ACLR.

1.2.1. Strength

Injury to the ACL can lead to quadriceps femoris muscle weakness. This weakness is believed to be due to a loss of feedback from the ACL that leads to a suppression in motor unit recruitment of the quadriceps (Williams et al. 2005). Following ACL-reconstruction there is often apparent quadriceps muscle atrophy in the injured limb (Palmieri-Smith et al. 2008). The progression of an ACL rehabilitation program is to have minimal differences in strength when compared bilaterally before they are allowed to return fully to athletic activities (Shelbourne et al. 1992). Typically this rehabilitation process takes between 6 months and a year. There is some controversy across the literature of whether ACL reconstructed individuals regain their quadriceps strength within the first years following surgery. Some research suggests quadriceps weakness persists up to a year after reconstruction (Bush-Joseph et al. 2001b; Lewek et al. 2002). Quadriceps weakness is associated with decreased knee flexion angles as well as knee extension moments (Bush-Joseph et al. 2001b). It is believed that individuals with an ACLR walk with less quadriceps contraction to avoid tibial translation, a phenomena known as “quadriceps avoidance” (Berchuck et al. 1990; Andriacchi and Birac 1993). In addition, greater hamstring contractions
have been observed as well, which is believed to be an effort to pull the tibia backwards and
again avoid translating the tibia during gait (Solomonow et al. 1987). Theoretically, these
adaptations, if not corrected, would lead to altered gait patterns and could reduce the ability to
attenuate forces about the knee joint during gait as well as decrease functional performance that
increase risk of a second ACL tear (Hart et al. 2010a; Schmitt et al. 2015). However, some
studies have observed restored quadriceps strength within the year post-surgery (Timoney et al.
1993; Roewer et al. 2011). At one year post-reconstruction there are those who restore at least a
90% of their quadriceps strength (10% bilateral difference is normal), and those who are barely
at 80% of their bilateral quadriceps strength (Lewek et al. 2002). Since quadriceps weakness has
been shown to contribute to altered movement patterns, it is important that quadriceps strength is
maintained to help ensure the health of the knee joint (Herzog et al. 2003). That being said, long-
term assessments have shown that quadriceps strength remains consistent with that of the
contralateral limb and healthy controls at two to even six years post-reconstruction (Mattacola et

1.2.2. Proprioception

Proprioception about the knee provides information about the movement and position of
the knee joint. Proprioceptive feedback provides the CNS with information it needs to make
adjustments to the environment. Loss of proprioception could result in instability of the joint and
poor position sense that could impact joint movement and/or joint moments. Within the first
three months post-reconstruction, there are decreases in proprioception (Fremerey et al. 1998).
By six months, proprioception is restored at full flexion and extension (Fremerey et al. 2000;
Reider et al. 2003). Three years post-reconstructive surgery there are no observable differences
in proprioception throughout any point in knee joint motion (Barrett 1991). While reconstruction
and rehabilitation do improve knee joint position sense, proprioceptive training has been shown
to improve proprioception as well as quadriceps and hamstring strength and is therefore an
important component of a rehabilitation program (Cooper et al. 2005).
1.2.3. Balance

The synchrony between balance, muscle activation, knee joint movement sensation, and joint position allows for dynamic knee stability and function during gait. The use of balance and proprioceptive training programs have been successfully utilized to reduce the risk of initial ACL injury. Immediately following reconstructive surgery there are balance, proprioception, and neuromuscular deficits present in the surgical limb as well as the contralateral healthy limb (Hoffman et al. 1999; Henriksson et al. 2001; Hiemstra et al. 2007; de Fontenay et al. 2015). Balance and proprioceptive training not only improves the control of bodily awareness, it can improve the response time to external forces as well as muscle strength (Hewett et al. 2002; Liu-Ambrose et al. 2003). Within the first couple of months post-reconstruction there continue to be differences in balance and lower limb postural control (Shiraishi et al. 1996; Chmielewski et al. 2002). However, there are those who have found no difference in balance when compared to healthy controls within the first 6 months following surgery (Hoffman et al. 1999; Henriksson et al. 2001; Moussa et al. 2009; Howells et al. 2011). Overall, for any balance and proprioceptive deviations there may have been, the tendency is for these to improve to be comparable to healthy controls by 1 year post-operation (Mattacola et al. 2002; Shi et al. 2010; Angoules et al. 2011; Anderson et al. 2016).

1.2.4. Spatiotemporal Parameters of Gait

Walking overground is the most commonly performed daily locomotor activity. While injury to the ACL happens most often in a sport-intensive environment, residual effects (i.e. progressive joint degeneration) occur from changes in loading patterns of less intensive activities of daily living (i.e. walking) (Chaudhari et al. 2008). Therefore, one of the first places to start looking at mechanical changes from ACLR would be to examine spatiotemporal characteristics of gait. Within the first couple of month’s to a year post-reconstruction ACL reconstructed individuals have similar stride length, step length, step time, cadence, and walking velocity as controls with healthy, intact ACLs (Devita et al. 1997; Gao and Zheng 2010). These spatiotemporal variables remain similar eight years after surgery, indicating other mechanical factors may have a greater impact on ACL reconstructed gait (Erhart-Hledik et al. 2017).
1.2.5. Kinematics of Gait

The use of kinematic analysis allows for the examination of motion at different body segments. With this information the movement of the segments and joint can be reconstructed and allows for changes in movement patterns to be recognized. For this reason several researchers have utilized kinematic measures to reconstruct joint and segment movements to try and identify changes in movement patterns (Chaudhari et al. 2008). Within the first year following surgery individuals who have had an ACLR have similar peak knee adduction angles as well as similar internal and external tibial rotation when compared to healthy controls (Bulgheroni et al. 1997; Georgoulis et al. 2003; Gao and Zheng 2010; Webster and Feller 2011; Wang et al. 2013b; Czamara et al. 2015). It has been observed that within the first year ACL reconstructed individuals do tend to walk with greater knee flexion and with decreased flexion of the hip and ankle (Ferber et al. 2002; Ferber et al. 2003; Gao and Zheng 2010; Karimi et al. 2013; Shabani et al. 2015). However, when looking more than a year post-reconstruction the majority of studies report no kinematic differences in the walking patterns of ACL reconstructed individuals compared to healthy controls (Bulgheroni et al. 1997; Georgoulis et al. 2003; Webster and Feller 2011; Hall et al. 2012; Webster and Feller 2012; Noehren et al. 2014; Patterson et al. 2014; Kaur et al. 2016). In summary, while there are some kinematic differences in gait during the first couple of months, these differences subside by the end of one year post-surgery and there are no discernable differences in gait kinematics in ACL reconstructed individuals.

1.2.6. Kinetics of Gait

Changes in loading patterns within the knee joint following the surgery and/or for compensatory strategies have been suggested to influence joint moments and torques that predispose ACL reconstructed individuals to secondary issues (Berchuck et al. 1990; Herzog et al. 2003; Hart et al. 2010a). There is some controversy across the literature when it comes to assessing kinetic measures of ACL reconstructed persons during gait. There are studies which report greater adduction moments and extension moments in the ACL reconstructed individuals during gait within the first year after surgery (Ferber et al. 2002; Ferber et al. 2003; Butler et al. 2009; Karimi et al. 2013), while others found no difference from healthy controls (Bulgheroni et
al. 1997; Webster et al. 2005; Karimi et al. 2013). Increased joint moments within a year post-surgery has been proposed to attribute to the onset of OA (Hart et al. 2010a). Within the first couple of years post-operation, however, knee adduction and extension moments tend to decrease, with strong evidence indicating joint moments comparable to healthy controls (Bush-Joseph et al. 2001a; Hall et al. 2012; Noehren et al. 2013; Zabala et al. 2013; Kaur et al. 2016). Despite the correction in joint forces within only a couple of years following surgery, the damage may have already be done and OA inevitable. We do see the impact of the surgery affecting other limbs, for instance there tends to be increased hip extension moments observed at least two years following surgery (Devita et al. 1998; Osternig et al. 2000; Hall et al. 2012). The increased moments at the hip may be a compensatory strategy to reduce tibial shear (Osternig et al. 2000).

While the majority of abnormal forces on the knee joint subside within a year after surgery, the impact this have may be more profound and should be considered in future ACL research.

1.3. PIÈCE DE RÉSISTANCE

The question still remains as to what factors, mechanical or other, are possibly contributing to the secondary injuries and degeneration experienced by people who have their ACL reconstructed? Researchers have used non-linear measures to quantify the coordination of movement patterns rather than traditional time-based analyses. What these scholars have observed is more irregular movement patterns of the legs during gait for ACLR persons patterns (Moraiti et al. 2009; Moraiti et al. 2010). In addition, the ACLR individuals tend to adopt altered walking strategies that manifest as a reduction of coordination about the knee joint (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011). This suggests that the lack of sensory feedback provided by a natural ACL these individuals develop changes in movement variability compared to healthy controls. In order for us to navigate successfully throughout our days, our central nervous system (CNS) relies on sensory input from across the body to make appropriate system-wide adjustments. The absence of a natural ACL may have greater neuromotor implications than has been investigated to date and these changes may manifest across systems. There are three main neuromechanical factors that may be impacted by an ACLR that have yet to be investigated: 1) in the coordination between legs during gait, 2) in the acceleration dynamics of the upper body during gait, and 3) the ability of the neuromotor system to respond to stimuli. Having a more comprehensive grasp on how ACLR impacts the body as a whole is imperative.
for understanding underlying mechanical changes post-surgery and allow us to develop strategies that reduce the residual symptoms of ACLR.

1.3.1. Coordination Strategies of Gait

With the absence of afferent feedback the ACL provides we would expect to see changes in the strategies utilized to for joint coordination and knee stabilization, resulting in changes in movement patterns. If this is the case, then traditional gait analysis and strength measures may not provide an analysis robust enough to detect changes in coordination strategies post reconstruction. Changes in coordination stability at the knee may not have an isolated impact to the knee joint either, but may also impact on the postural system. Analyses of relative phase (RP) have been utilized to help understand the impact of phase deviation and the stability of underlying coordination between two oscillators. Previous studies that have examined coordination of walking in individuals with an ACLR or chronic ankle instability and have typically observed a decrease in the standard deviation of relative phase within the limb, suggesting an increase in the coordination stability (Hamill et al. 1999; Heiderscheit et al. 2002; Miller et al. 2008; Drewes et al. 2009; Hamill et al. 2012; Herb et al. 2014; Yen et al. 2017). However, asymmetries or differences between legs have been shown to influence the coordination pattern and its stability. It is of interest to examine inter-limb coordination to understand whether ACLR affects coordination between the legs while walking. Insight into gait strategies may be gained by examining the degree of stability and coordination of ACL reconstructed walking patterns.

1.3.2. Upper Body Accelerations

Mechanoreceptors found within a natural ACL provide the CNS with feedback necessary to execute gait and postural adaptations. The loss of a natural ACL may limit feedback from the knee necessary to maintain postural control as well as normal gait patterns (Fremerey et al. 2000). While kinetic and kinematic measurements of the hip suggest ACL reconstructed individuals return to “normal” parameters, they tend to demonstrate changes in trunk kinematics during gait. These changes include increased trunk lean, ipsilateral trunk lean, and decreased
trunk stability (Noehren et al. 2014). In addition, individuals who have had an ACLR have demonstrated increased stiffness of their trunk that may have implications on postural control (Boggess et al. 2018). These results suggest compensatory strategies are not limited to the lower extremity and the changes seen at the trunk may affect the postural control system entirely. Postural control is considered a complex motor skill requiring the control of balance and orientation against acting forces (Shumway-Cook and Woollacott; Horak 2006). The central nervous system (CNS) makes rapid and overt adjustments to maintain postural control. One of the largest factors that may compromise postural stability is that 2/3 of our body weight is a large distance away from the ground, making the human body an intrinsically unstable platform (Winter 1995). For this reason, the trunk plays an active role in maintaining stability during gait under dynamic conditions (Winter 1995; Woollacott and Tang 1997). One of the most critical functions of the trunk is minimizing the impact of gait-related oscillations on the head. To do so, the trunk acts as a filter, dampening oscillations that travel up the body to the head. This is evidenced by lower amplitude of acceleration patterns at the head compared to accelerations at the low back (Menz et al. 2003; Kavanagh et al. 2004; Kavanagh et al. 2005b). Somatosensory information in combination with the neuromuscular and structural control of the trunk work together to maintain head control. Alterations to somatosensory information or neuromuscular function, however, may impact one’s ability to maintain postural control and thus the ability to stabilize the head.

1.3.3. Responsiveness of the System

The process of adapting to a stimulus requires processing the stimulus, determining the most effective response, and sending a signal to initiate and perform the desired response that allows us to navigate daily life (Grabiner and Enoka 1995). This allows us to adapt and correct ourselves to avoid loss of balance and/or injury. Measures of reaction time have revealed alterations in sensorimotor function following injury. Individuals with chronic ankle instability (CAI) exhibit delayed responsiveness of the peroneal muscles during sudden inversion perturbations (Hoch and Mckeon 2014; Thompson et al. 2017). Research has shown delays in closed-loop as well as open-loop neuromuscular control in persons with CAI. A common method of measuring the closed loop reflexive response in individuals with CAI has been to use
unanticipated ankle perturbation techniques. While the general accord is that there are delays in muscle responsiveness in CAI, there is research that has observed no differences (Gutierrez et al. 2009). This may be due to the perturbation techniques used which captures a plethora of somatosensory responses that may mask any neuromuscular deficits specifically at the ankle (Kavanagh et al. 2012). Kavanagh and colleagues (2012) looked at the voluntary responsiveness of the peroneus longus and tibialis anterior. To avoid stimuli that may impact the reaction time, the participants were given a visual or audio stimulus from which they were to perform the movement task. The researchers observed that individuals with CAI have a slower response time of the peroneus longus when moving the ankle to eversion. These results indicate that the neuromuscular function of the ankle is hindered after injury, and that the reaction time as well as motor time of the muscle is delayed in CAI (Kavanagh et al. 2012). While much of studies on ACL reconstructed individuals has focused on the mechanical impact of this surgery, it is possible that ACLR may impact neuromuscular function (i.e. reaction time), similar to CAI on the ankle.

Reaction time can be used to determine whether the impact of the injury is peripheral or central in nature. Because the damage is to a specific limb, the assumption would be that this injury would only exhibit peripheral changes. This theory is supported when comparing the peroneal muscle reaction time of the contralateral uninjured side in persons with CAI, there are no observable differences compared with healthy controls (Johnson and Johnson 1993; Löfvenberg et al. 1995). However, postural compensatory strategies have been observed on both the ankle with CAI and the contralateral healthy ankle (Beckman and Buchanan 1995). While changes in the postural system may be due to the changes in the peripheral nervous system, the changes as a result of injury could impact the performance of the central nervous system. One of the limitations in research looking at reaction time of chronic ankle instability is that the task is most often a simple reaction time task. Choice reaction time accentuates the decision time component as well as performing the correct task and the time it takes to perform the task (Lord and Fitzpatrick 2001). Overall, choice reaction time provides a window into the processing time of the central nervous system. There has been no direct assessment of whether an ACLR impacts reaction time during a postural task. It is expected that changes in responsiveness to simple and choice reaction time will reveal the neuromotor impact of reconstructive surgery. In addition,
examination of reaction time involving movement of a more dynamic motor responses (i.e. taking a step) has shown to identify individuals at risk of falling (Lord and Fitzpatrick 2001). While there are no observed changes in balance measures in persons with an ACLR, there may be still neurophysiological deficits in these individuals. A postural stepping reaction time (i.e. slower stepping response time) requires both central and peripheral neurophysiological factors to initiate and control taking a step (Patla et al. 1993). There has been no assessment of whether neurophysiological factors, central or peripheral, changes as a result of ACL reconstructive surgery.

1.4. STATEMENT OF THE PROBLEM

The majority of studies investigating individuals who have had an ACL reconstructive surgery focus on the effects specific to the knee joint and surrounding segments. The general findings of these studies are that there are conflicting results with regards to measures of strength, proprioception and sensation, balance, spatiotemporal measures, kinematics, and kinetics and whether or not individuals with an ACLR are similar to healthy controls across these variables. The only consistent difference is increased forces within the knee joint and even in the hip joint within the first year following surgery. While the forces seem to correct themselves, the damage from increased forces and compensatory factors may have already been done. In addition, these immediate mechanical changes following surgery have shown to impact the hip and even the trunk motions during gait. When looking at movement variability measures, ACLR individuals demonstrate irregular movement patterns that illustrate a change in coordination around the knee during gait (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011). Without the sensory feedback provided by a natural ACL, it is believed that ACL reconstructed individuals may not be able to fully gain function after their knee has been repaired (Decker et al. 2011). Perhaps there are greater systemic implications resulting from the absence of a natural ACL that may have greater neuromotor implications than has been investigated to date and these changes may manifest across systems. Pervious investigations are limited in that they have focused on traditional biomechanical assessments as well as have only examined the impact on the legs. We propose there is a more intrinsic impact ACLR has on movement and traditional
biomechanical measures may not be robust enough to elucidate neuromechanical factors that may be effected by ACL reconstructive surgery.

1.5. GENERAL PURPOSE OF THE STUDY

The general purpose of this study is to examine neuromechanical factors in ACLR persons compared to healthy controls and determine the implications of this surgery has on several systems. It is believed that the impact of ACLR may manifest in three different ways throughout the body: 1) in the acceleration dynamics of the upper body during gait, 2) in the coupling strength between limbs during gait, and 3) the ability of the neuromotor system to respond to stimuli. These measures will provide a more comprehensive examination of how the loss of sensory information provided by the ACL impacts movement dynamics. By having a deeper understanding of the neuromechanical implications following ACLR will allow for the development of more poignant strategies for reducing the residual symptoms post-ACLR.

1.6. SPECIFIC AIMS AND HYPOTHESES

Experiment One

The aim of this study was to examine differences in inter-limb coordination during walking at different speeds for individuals with a reconstructed ACL compared to age-matched controls.

Specially, it is hypothesized that;

1. Preferred walking speed will be the most stable walking condition for persons with ACLR and the age-matched healthy controls.

2. Persons with an ACLR will demonstrate a reduced coordination stability between legs (indicated by reduced coupling strength between legs) compared with age-matched controls.
Experiment Two

The aim of this study is to examine differences in the anterior-posterior, medio-lateral and vertical acceleration from the lower trunk, neck and head for individuals who have had an ACLR compared to healthy individuals of similar age.

Specifically, it is hypothesized that;

1. Individuals with a history of ACLR will demonstrate differences in the vertical and anterior-posterior accelerations for the trunk, neck and/or head regions compared to age-matched controls which will reflect a diminished ability to compensate for gait oscillations.
2. The diminished capacity to attenuate gait related oscillations through the trunk to the head compared to age-matched controls.

Experiment Three

The aims of this study were to; 1) investigate differences in general neuromotor function (i.e., lower limb strength, gait, balance, and simple reaction time) between ACLR individuals and healthy individuals of similar age, and 2) examine whether the reaction time responses (both choice and simple) assessed under forward stepping conditions are negatively affected by previous damage to the ACL.

Specifically, it is hypothesized that;

1. No differences will be found between persons with ACLR and the age-matched healthy controls in terms of lower-limb strength, walking speed, standing balance, and simple reaction time (performed under seated conditions).
2. The choice and simple reaction time responses assessed under postural stepping conditions will be slower for the ACLR individuals compared to age-matched controls.
CHAPTER II

LITERATURE REVIEW

In order to investigate the long-term neuromechanical effects of ACL reconstructive surgery it is first important to understand what the ACL is and how it functions. This first section of the literature review will discuss the general anatomy and function of the ACL within the knee. Further, this section will expound upon the implications of injury to the ACL, the treatment options as well as the rehabilitation after reconstructive surgery.

2.1. THE ANTERIOR CRUCIATE LIGAMENT

2.1.1. Anatomy of the knee

The knee consists of three articular surfaces that form two joints: the patellofemoral joint and the tibiofemoral joint, both contained within one joint capsule. The patellofemoral joint is made up of the patella and its articulation on the femur. The patella is a large sesamoid bone that is situated on the intercondylar notch of the femur and is embedded in the quadriceps femoris muscle proximally and the patellar tendon distally (Amis et al. 2003; Tecklenburg et al. 2006). The purpose of the patellofemoral joint is to improve the effectiveness of the quadriceps during knee extension by increasing the moment arm of the patellar tendon as well as reduces articular friction of the tendon over the joint (Hungerford and Barry 1979).

The tibiofemoral joint consists of the distal end of the femur and its articulation with the proximal end of the tibia (see figure 2.1). The distal end of the femur is made up of two femoral condyles, the medial and lateral femoral condyles (Yoshioka et al. 1987). Above the medial femoral condyle is the medial epicondyle where several muscles insert (i.e. medial collateral ligament, and adductor magnus) and originate (i.e. the medial head of the gastrocnemius) (Griffin et al. 2000). On the lateral side, the lateral epicondyle serves as the insertion for the
Figure 2.1. Representation of the anatomical structures that make up the knee.
lateral collateral ligament as well as the origin of the lateral head of the gastrocnemius (Griffin et al. 2000). The two femoral condyles are separated by the intercondylar notch, the aspect where the anterior and posterior cruciate ligaments originate and insert into the tibia (Girgis et al. 1975). The femora condyles articulate with the proximal tibia on two tibial plateaus. The tibial plateaus are concave and are covered with fibrocartilage known as the menisci (Ogden 1974). Between the tibial plateaus is the intercondylar eminence, serving as the attachment site for the anterior and posterior cruciate ligaments to the tibia (Girgis et al. 1975).

2.1.2. Anatomy of the ACL

The anterior cruciate ligament (ACL) is a thick band-like structure of connective tissue in the knee joint, intended to resist tibial translation and rotational loads experienced at this joint (Butler et al. 1980). Based on its structural design the ACL acts as the primary stabilizing ligament in the knee joint. It has been well documented that the ACL comprises of two bundles, the anteromedial bundle (AM) and the posterolateral bundle (PL) (Girgis et al. 1975; Duthon et al. 2006; Kopf et al. 2009). There are some who also report a third bundle, the intermediate bundle (IM) (Norwood and Cross 1979; Amis and Dawkins 1991). While some anatomical dissections suggest three or more bundles, it is believed that there are only two functional bundles, the AM and PL bundles (Petersen and Zantop 2007). The origin of the ACL is on the posteromedial surface of the lateral femoral condyle and the insertion is on the anteromedial aspect of the tibial intercondylar spine for the AM and posterolateral aspect of the tibial intercondylar spine for the PL (Girgis et al. 1975). Both bundles play a role in resisting tibial translation whereas the PL plays a greater role in providing rotational stability (Zantop et al. 2007).

With an understanding of the structural components of the ACL understood, the next step is to delve deeper into the cellular mechanisms that comprise the ACL. Mechanoceptors are located at the proximal and distal portions of the ACL and provide proprioceptive information for postural changes. When activated these receptors elicit a motor response of the muscles surrounding the knee, also referred to as the ACL reflex (Duthon et al. 2006). Evidence of the ACL reflex has been confirmed through the detection of sensory evoked potentials after direct
stimulation of the ACL (Ochi et al. 2002). The mechanoreceptors in the ACL provide sensory information with regards to position, dynamic movement, and loading of the knee. Receptors on the spinal cord receive the sensory information and through spinal interneurons, ascend this information to supraspinal sensory cortex and allow for adaptations and control of movement. Thus the mechanoreceptors in the ACL play an important role in providing afferent feedback needed for normal knee function (Krogsgaard et al. 2002).

2.1.3. Injury to the ACL

The ACL is one of the most commonly injured ligaments in the knee, accounting for roughly 200,000 injuries per year in the United States (Agel et al. 2016). The ACL is most vulnerable to injury when the knee is in 30° to 90° of flexion, and when the bundles are twisted and most taut (Zantop et al. 2007). Rupture of the ACL can occur through contact or noncontact mechanisms. It is possible that there is an isolated tear of either one of the bundles depending on the mechanism of injury. Contact mechanisms occur most commonly when the ACL is taut and a translational force is applied to the knee while the leg is fixed (Hewett et al. 2006b). Noncontact ACL tears occur during multi-planar mechanisms, that is to say mechanisms from the sagittal, frontal, and transverse planes contribute to the incidence of a noncontact ACL tear (Shimokochi and Shultz 2008; Quatman et al. 2010). The majority of ACL injuries occur from non-contact mechanisms. In addition, females have a 4- to 6-fold greater risk of sustaining an ACL injury than do males (Hewett et al. 2005). Acute ACL tears are characterized by an audible “popping” sound and sensation at the time of injury. Further, persons who sustain an ACL tear may have knee pain as well as knee instability, swelling of the knee joint, pain with range of motion tasks, as well as haemarthrosis. Diagnosis of an ACL tear can be done with special tests such as the Lachman’s Test and Anterior Drawer Test. Injury to the ACL is often not an isolated event and is associated with injury to the meniscus, subchondral bone, collateral ligaments (medial or lateral), or a combination of injuries (Noyes et al. 1980; Spindler et al. 1993; Spindler and Wright 2008). To confirm the diagnosis of an ACL tear, and other associated injuries, magnetic resonance imaging (MRI) is considered the gold standard (Crawford et al. 2007).
2.1.4. Treatment of ACL Tears

Despite the major role the ACL plays in stabilizing the knee, rupture of this ligament does not automatically result in functional impairment of the knee joint. Individuals who are not physically impaired by this injury are classified as ACL deficient copers, versus individuals who are debilitated by this injury are classified as non-copers (Snyder-Mackler et al. 1997). It is estimated that one-third of active individuals who sustain an ACL rupture can conservatively manage the injury and are able to return to pre-injury levels of activity (Anderson et al. 2016). Conservative management of ACL rupture involves rehabilitation to strengthen the surrounding musculature of the ACL deficient knee to return the person to functional performance. That being said, it is recommended that individuals who want to return to sports that require cutting and rapid changes in direction to have the ACL surgically reconstructed for the health and longevity of the knee joint (Kaplan 2011). The greater the trauma to the knee associated with the ACL rupture, the more likely surgical treatment will provide the best results for the individual (Frosch et al. 2013). Additionally, there is a greater likelihood of additional damage to surrounding structures (e.g. meniscus) than an isolated ACL tear, therefore reconstruction allows for management of any additional structural damage (Spindler et al. 1993).

When determining the best treatment option for an ACL tear, it is important to look at the quality of life outcomes of each. When examining the quality of life of ACL deficient individuals classified as copers compared to ACL reconstructed individuals and to healthy controls, the knee-related quality of life, irrespective of treatment, is reduced in those who have sustained an ACL tear (Filbay et al. 2015). Interestingly, the quality of life of ACL deficient copers were similar to those who underwent reconstructive surgery (Filbay et al. 2015). The long-term biomechanical difference between those who opt for conservative management versus surgical treatment of and ACL rupture remains unknown. What has been shown is that ACL reconstructed individuals exhibit greater activity levels than their ACL deficient counterparts (Farshad et al. 2011; Anderson et al. 2016). However, the risk of initially injuring an ACL occurs 1 in 80 people, the rate of re-injury or injury to the contralateral ACL post-reconstruction is 1 in
Figure 2.2. Conceptual diagram of factors that impact the ACL during standing and gait-related tasks as well as factors that impact the knee post-ACLR.
17 within the first two years after surgery (Wright et al. 2007; Paterno et al. 2010). Therefore, it is important to determine what strategy is best for each individuals’ lifestyles as well as the option that will best maintain the integrity of the knee joint.

2.1.5. ACL reconstructive surgery

The ACL is reconstructed using either an autograft or an allograft, a choice that remains controversial among researchers and surgeons as to which one is superior. An autograft is a tissue graft from the patient’s own tissue whereas an allograft is a tissue graft from a donor, transplanted from one person to another. Autograft reconstruction typically utilizes tissue from either the patellar tendon or hamstring, more recently the quadriceps tendon has been used as a graft choice as well (Poolman et al. 2007; Slone et al. 2015). The benefits of using an autograft are that the graft is not rejected by the body nor are infections transmitted from the graft. One of the greatest downfalls of an autograft is the result of donor-site morbidity. An allograft reconstruction can utilize donor grafts that are extracted from the donors’ patellar tendon, hamstring, or intact ACL. The benefits of using an allograft are that there is no donor-site morbidity, as well as surgery and rehabilitation time are shorter. The disadvantages are that the body can reject an allograft and infections are more easily transmitted. Irradiation of the allograft is common practice to reduce transmission of infections. However, irradiating the graft has shown to alter the properties of the graft, increasing the rate of failure compared to nonirradiated allografts (Balsly et al. 2008). Synthetic ACL grafts can also be used for reconstruction, however despite the good short term results there is no research showing the long-term effectiveness of this type of graft (Machotka et al. 2010).

The use of an autograft shows greater patient satisfaction than an allograft (Anderson et al. 2016; Zeng et al. 2016). In terms of function, some have found no difference between autografts and nonirradiated allografts while others have found autografts to show greater knee function than allografts (Mariscalco et al. 2014; Anderson et al. 2016).
2.1.6. Rehabilitation following ACL reconstructive surgery

The goal of rehabilitation after surgery is to restore the functionality of the limb to the person without adversely affecting the recovery of the graft. Unfortunately, most of the information we do have about the biological healing process of an ACL graft is based on mammalian models that are then applied to human models. Therefore the information with regards to the necrosis and revascularization the graft undergoes in humans is predominantly reliant on studies conducted with various mammals (Kondo et al. 2012; Giordano et al. 2015). There are a few studies that have used human models to study the healing process of an ACL graft in a less invasive extent (Horstman et al. 1993; Rougraff et al. 1993; Falconiero et al. 1998; Lee et al. 2004; Zaffagnini et al. 2007). What these studies show is that full maturation of autogenous ACL grafts occur after one year post-reconstruction (Rougraff et al. 1993; Falconiero et al. 1998; Zaffagnini et al. 2007). That being said, vascularization and morphology of the graft fibers appear to be mature as early as 6 months (Rougraff et al. 1993; Falconiero et al. 1998). Allografts tend to take longer to mature, taking more than two years to mature post-surgery (Horstman et al. 1993; Lee et al. 2004). The studies done on human grafts tend to have small sample sizes and therefore further studies are necessary to fully understand the biological healing process of human ACL grafts.

The information with regards to the healing process after ACLR as well as an understanding about how much strain can be placed on a newly implanted graft is the bases from which ACL rehabilitation programs are designed. Anterior cruciate ligament strain has been investigated with cadavers and in-vivo, information that enables researchers to most appropriately determine how different exercises or motions strain the healing graft (Butler et al. 1980; Beynon and Fleming 1998). Based on the graft healing process and strain on the ACL, rehabilitation can be performed in a manner that least hinders the recovery of the graft. Rehabilitation programs are designed to return the person to their normal activity-level of functional movement, restore their self-efficacy while diminishing fear-avoidance, and promote education and awareness of injury prevention (Nyland et al. 2016). Clinicians working with the person with an ACLR progress each program based on individual deficits, targeting cognitive,
biomechanical, structural, and neuromuscular factors that may predispose the individual to re-injury (see figure 2.2).

The primary postoperative goal after an ACLR is to restore range of motion (ROM) at the knee (Noyes et al. 1987; Biggs et al. 2009; Shelbourne et al. 2012). Normal ROM is considered to be 0° of extension and 135° of flexion, though this range is unique to each person and thus each person’s “normal” range should be determined based on the ROM of their contralateral limb (Shelbourne et al. 2012). ROM can be assessed by using passive ROM and with goniometric measurements. Passive ROM allows the clinician to bring the leg into extension and then into flexion to compare the amount of movement and the quality of the end feel to the contralateral healthy limb (Shelbourne et al. 2012). Goniometric measurements provides a more objective measure of ROM, providing how many degrees of extension/flexion a person has based on these measurements. The loss of ROM has been shown to have long-term deficits on gait mechanics, and has been shown to be associated with the development of osteoarthritis, therefore it is vital to regain full ROM (Harner et al. 1992; Shelbourne et al. 2012).

Following ROM, regaining strength is the next phase in restoring functionality after ACLR. Closed kinetic chain (CKC) and open kinetic chain (OKC) exercises are both used to improve strength deficits rehabilitation. Closed kinetic chain exercises are when the distal segment is fixed to an immobile surface and multiple joints and muscle groups are activated simultaneously to produce compound movements. Open kinetic chain exercises are when the distal segment is not in contact with a surface often leading to isolated movements of a single joint. It is believed the CKC are safer than OKC, however there is no evidence to support this claim (Kvist 2004; Lobb et al. 2012). Often CKC are introduced first in the rehabilitation program due to its reduction in shear forces as well as being more functional exercises and OKC exercises are introduced after sufficient bone healing has occurred. Regardless, no study has found one exercise better than the other, rather we see greater improvements in strength and faster recovery times when both CKC and OKC are used in the rehabilitation program (Grodski and Marks 2008; Tagesson et al. 2008). Therefore, strengthening programs for ACLR rehabilitation should include both OKC and CKC exercises that are regulated with graft healing
to promote the most effective environment for healing as well as preventing muscle atrophy (Fleming et al. 2005; Saka 2014).

Rehabilitation programs are designed to target deficits present following ACL reconstructive surgery to return the person to pre-injury levels. Retraining balance and proprioception positively impacts joint position sense. Balance and proprioceptive training not only improves the control of bodily awareness, it can improve the response time to external forces as well as muscle strength (Hewett et al. 2002; Liu-Ambrose et al. 2003). This will assist in returning the person to full activity following reconstructive surgery. Abnormal neuromuscular control of the lower limbs during the execution of athletic movements such as cutting and jump landing tasks have been shown to contribute to ACL injury. Neuromuscular training allows clinicians to target deficits in tasks requiring dynamic joint stabilization and improve neuromuscular control, specifically while performing athletic movements. Along the same lines, neuromuscular training has been shown to effectively restore physical function by incorporating a combination of strength, balance, plyometric, and sport-specific exercises into the person’s rehabilitation program. Neuromuscular training has been used as an injury prevention program and has been shown to successfully reduce the risk of ACL injury (Hewett et al. 2006a; Myer et al. 2012; Sugimoto et al. 2012). After ACL reconstructive surgery, the rate of a second ACL injury is astronomically greater than people who have never sustained an ACL injury (Paterno et al. 2012; Wiggins et al. 2016). The risk of initially injuring an ACL occurs 1 in 80 people, the rate of re-injury or injury to the contralateral ACL post-reconstruction is 1 in 17 within the first two years after surgery with decreased neuromuscular control being a predictor of a second ACL injury (Wright et al. 2007; Zazulak et al. 2007; Paterno et al. 2010). Neuromuscular training after the first ACLR has been shown to reduce the risk of a second ACL injury (Di Stasi et al. 2013a).

2.1.7. Accelerated versus non-accelerated rehabilitation

Over the past couple of decades the treatment and rehabilitation of an ACLR has come a long way. Early treatment included complete immobilization of the leg followed by immobilization of the knee joint itself (Clancy Jr et al. 1981; Amiel et al. 1986). Today we know
that immobilization after ACL reconstructive surgery has adverse effects on the knee joint and the incorporation of immediate mobilization is optimal (Henriksson et al. 2002). In addition, we know that accelerated rehabilitation programs can be just as safe and effective as traditional non-accelerated programs. As opposed to the traditional 6-12 month rehabilitation programs, accelerated rehabilitation programs are aimed to return individuals to full activity in less than 6 months. One of the fears of an accelerated program is that placing strain on the graft too soon will result in elongation of the graft that would produce abnormal laxity of the knee joint (Clancy Jr et al. 1981). When accelerated programs are compared to traditional non-accelerated programs, however, there is no difference in knee joint laxity between participants who underwent either type of programs (Shelbourne and Nitz 1990; Beynnon et al. 2005; Beynnon et al. 2011; Kruse et al. 2012). Additionally, there are no differences in strength, proprioception, functionality, patient satisfaction, or activity level of those who have undergone either an accelerated or non-accelerated rehabilitation program (Beynnon et al. 2005; Beynnon et al. 2011). Overall, accelerated rehabilitation has shown no adverse effects on the integrity of the graft or the person’s ability to return safely to activities (Kruse et al. 2012). That being said, Nagelli and Hewett (Nagelli and Hewett 2017) have suggested that joint health and function are achieved at two years and thus return to sport should be no less than that amount of time to reduce the rate of second ACL injury. With second ACL tears being common in those who undergo ACL reconstructive surgery, it would be of interest to see if the incidence of second ACL tears correlate with the type of rehabilitation program the individual undergoes.

2.1.8. Outcome assessments of ACLR and return to play

Postoperative rehabilitation programs aim to return the person to previous levels of sport participation or activity level. Additionally, one of the anticipated outcomes of people who undergo ACL reconstructive surgery is to return to their pre-injury levels of activity and sport participation (Feucht et al. 2016). To accomplish this, as mentioned above, rehabilitation programs are designed to regain full ROM, strength, balance, proprioception, and neuromuscular function. Shelbourne and colleagues (1990, 1992) have provided a rehabilitation protocol outlining specific phases (Shelbourne and Nitz 1990; Shelbourne et al. 1992). In order to advance through the five phases, each phase has specific criteria that need to be met with regards
Table 2.1. Progression through ACLR Protocol adapted from Shelbourne et al. (1992).

<table>
<thead>
<tr>
<th>Phase</th>
<th>Criteria</th>
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| I (Pre-reconstruction) | - Reduce Swelling  
- Restore full knee motion  
- Strengthen injured knee (as possible)  
- Mental preparation for surgery and rehabilitation process |
| II (Post-reconstruction) | - Reduce swelling and joint effusion  
- Obtain full passive knee extension (compared bilaterally)  
- Obtain 0-90° passive knee flexion  
- Maintain quadriceps control and activation  
- Patellar mobility  
- Wound healed |
| III | - Achieve independent ambulation and normal gait pattern  
- Minimal to no joint effusion  
- No joint line or patellofemoral pain  
- Obtain a minimum of 60% quadriceps strength (compared bilaterally)  
- Obtain 135° active knee flexion |
| IV | - Achieve sport-specific activities (cutting, jumping, running, stopping)  
- Minimal differences in strength measures bilaterally  
- Improve endurance and power bilaterally  
- At the completion of Phase IV the patient should be able to return fully to activities of daily life, and sports with strength, ROM, and with no pain. |
to ROM, strength, balance, proprioception, neuromuscular function, clinical evaluation, endurance, and sport specific tasks and scores. Table 1 outlines the specific criteria for each of the phases of an ACLR rehabilitation protocol based on recommendations from Shelbourn and colleagues (1992) as well as Wilk and colleagues (2003) (Shelbourne et al. 1992; Wilk et al. 2003). Regardless of the rehabilitation protocol after ACL reconstructive surgery, the timeline for progression will vary from person to person.

Once a person has gone through their ACL rehabilitation program, the next step is to determine if they are ready to return to play. There are a handful of objective measures that can be used to determine if the person has fully regained function of the reconstructed knee and are physically as well as mentally ready to return to play. To date, there is no set criterion on what measures best evaluate a person’s readiness to return to sport, however the battery of tests used to clear an athlete to return to play are designed to assess risk factors associated with ACL injury. These include a clinical examination, functional test evaluation, and patient reported outcomes (PROs). The clinical examination includes a Lachman test, a pivot shift test, and the use of a knee ligament arthrometer (i.e. the KT-2000). These tests allow a clinician to evaluate the integrity of the graft manually, assessing the laxity and stability of the ligament. Functional evaluation includes vertical jump test, single-leg hop test, and single-leg hop test over a distance. These jumping tasks have been shown to provide a reliable means of objectively measuring the ability of the lower-extremity to tolerate external forces (Brosky Jr et al. 1999). In addition, quadriceps and hamstring strength are measured bilaterally either by measuring the thigh circumference and/or the quadriceps-to-hamstring ratio. One of the main reasons people do not return to pre-injury levels of activity following reconstructive surgery is a fear of re-injury (Kvist et al. 2005). As a result, psychological effects of ACL injury and surgery are important measures for a return to sport evaluation. Patient reported outcome questionnaires such as the International Knee Documentation Committee and the Knee Injury, Osteoarthritis Outcome Scale, Psychovitality questionnaire, the Emotional Responses if Athletes to Injury Questionnaire, the Tampa Scale for Kinesiophobia, and the ACL Return to Sport Index allow for clinicians to evaluate the psychological readiness of a person post-ACLR to return to play (Kvist et al. 2005; Webster et al. 2008; Langford et al. 2009; Ardern et al. 2011). While there is no consensus on a set battery of tests that should be used to determine a person’s readiness to return to sport
following ACLR, a multifaceted approach encompassing physical as well as psychological outcomes is important to help determine whether someone is ready to return to play.

While a set of criteria for return to play assessments is missing from the literature, an important consideration is to examine the validity of current return to play test batteries. The battery of tests include, but are not limited to, measures of strength, functional assessments, and PROs of knee symptoms and function. The proportion of people who pass their return to play criterion before returning to sport on average is less than 50% (Di Stasi et al. 2013b; Logerstedt et al. 2014; Toole et al. 2017). What has been shown is that individuals typically pass specific portions of their return to play criterion (i.e. functional assessments), however only a small percentage are able to pass all portions of the return to play criteria (i.e. functional assessments, PROs, strength assessments) (Toole et al. 2017). This point is emphasized when individuals already participating in sport were evaluated and were unable to adequately pass return to play criteria. The pass rate has been shown to be around 25%, indicating that individuals are returning to play without meeting basic return to play criteria (Webster and Hewett 2019). However, when performing similar evaluations on individuals who have no history of ACL injury, the functional performance has been shown to be equivocal to those who did have an ACLR (Fältström et al. 2017). What is clear from these studies is that further research is needed to determine the battery of tests can optimize return to sport criteria to reduce the rate of secondary injuries as well as increase the return to pre-injury levels of sport after ACLR.

### 2.2. DEFICITS FOLLOWING ACLR

While the goal of reconstructive surgery is to enable a person to return to their previous levels of sport activity with improved functional capacity of the knee, this has not been the case for the majority of people undergoing this surgery. Despite the high rates of successful surgical and rehabilitative outcomes, only 63% of individuals return to their pre-injury levels of sport participation (Ardern et al. 2011; Ardern et al. 2014). In addition, more than 60% of these individuals develop symptoms of post-traumatic osteoarthritis (PTOA) 10-15 years after surgery, with the number increasing above 80% if the injury was not isolated to just the ACL (Øiestad et al. 2010). These individuals have shown a 10-fold increased risk of developing osteoarthritis...
compared to those who have never sustained a knee injury (Roos et al. 1998; Gillquist and Messner 1999). PTOA is observed after injurious impact loading in the joint and articular surface (Anderson et al. 2011). Of those who develop OA of the knee 9.8% of these are considered PTOA and thus, the estimated annual expenditure on those with PTOA alone is over three billion dollars in the United States (Brown et al. 2006). Besides the financial burden, the fact that PTOA affects mainly younger adults for whom joint replacement is not desirable, further understanding of the development and progression of this disease is needed to develop successful rehabilitative strategies that reduce these residual effects.

Changes in cartilage-loading patterns as a result of altered gait mechanics have been hypothesized to be a contributing factor in the long-term health following ACL reconstructive surgery (Andriacchi et al. 2009). Therefore several studies have investigated the biomechanical properties in persons with ACLR to help determine if certain structural and/or functional components of the knee may be the contributing factors to knee health.

2.2.1. Deficits in structural components of the knee

Strength

Following ACL injury there are several modifications observed in motor output. Specifically, neuromuscular inhibition of the quadriceps as well as increased activation of surrounding musculature to facilitate in protecting the joint. The inhibition observed at the quadriceps muscle after ACL injury is known as arthrogenic muscle inhibition (AMI), which is induced by distention of the knee capsule from swelling and/or damage in the joint (Snyder-Mackler et al. 1994; Torry et al. 2000; Hopkins et al. 2001). While ACLR improves functional stability, AMI can still persist after reconstruction (Hopkins and Ingersoll 2000). The progression of an ACL rehabilitation program is to have minimal differences in strength when compared bilaterally before they are allowed to return fully to athletic activities (Shelbourne et al. 1992). Typically this rehabilitation process takes between 6 months and a year, depending on the person. Research has shown quadriceps strength deficits of 20% persist 6 months following reconstruction (Ingersoll et al. 2008). Quadriceps activation failure due to AMI may cause
persistent weakness that can impede rehabilitation. Within the first 3 months there is significant quadriceps weakness in ACL reconstructed individuals compared bilaterally and to healthy controls (Morrissey et al. 2004; Drechsler et al. 2006). However, research has shown in the same time frame following surgery there is no difference bilaterally in quadriceps muscle force production when using a superimposed twitch at the femoral nerve (Snyder-Mackler et al. 1994) suggesting these individuals were not voluntarily able to recruit the same amount of motor units as the uninvolved limb. There is some controversy across the literature of whether ACL reconstructed individuals regain their quadriceps strength within the first years following surgery or if this recovery takes longer. There are those who find quadriceps weakness persists 6 to 12 months following surgery (Pfeifer and Banzer 1999; Mikkelsen et al. 2000; Bush-Joseph et al. 2001b; Lewek et al. 2002; Hart et al. 2010b).

One issue with ACL strength studies is that they do not often discern between those who have high quadriceps function (i.e. no evidence of muscle inhibition) as opposed to those who have low quadriceps function (i.e. evidence of muscle inhibition) in the ACL reconstructed groups. At one year post-reconstruction there are those who restore at least a 90% of their quadriceps strength (10% bilateral difference is normal), and those who are barely at 80% of their bilateral quadriceps strength (Lewek et al. 2002). This could be a limitation to studies because research shows that those with high quadriceps function pass return to sport criteria, have higher self-reported function, have greater functional performance, and their neuromuscular movement patterns are comparable to healthy controls whereas those with low quadriceps function are the opposite (Lewek et al. 2002; Schmitt et al. 2012; Di Stasi et al. 2013c; Palmieri-Smith and Lepley 2015; Pietrosimone et al. 2015). These controversial findings across the literature are seen spreading two years following surgery. Two years following surgery there are those who find differences in the strength of the contralateral limb and healthy controls (Hiemstra et al. 2007), and those who find no strength deficits with that of the contralateral limb and healthy controls (Roewer et al. 2011; Hall et al. 2012). These variances in findings could be the result of methodological differences. None of the studies report how the strength of the contralateral limb is maintained from the time of surgery to evaluation. Nevertheless, long-term assessments have shown that quadriceps strength remains consistent with that of the contralateral limb and healthy controls at two to even six years post-reconstruction (Mattacola et al. 2002;
Roewer et al. 2011; Hall et al. 2012; Kaur et al. 2016). Since quadriceps weakness has been shown to contribute to altered movement patterns, it is important that quadriceps strength is maintained to help ensure the health of the knee joint (Herzog et al. 2003). Regardless, whether a person is considered to have high or low quadriceps function does not discern whether they will or will not develop related secondary issues or osteoarthritis.

While maximal quadriceps activation is not a requirement to perform gait tasks, strength deficits can influence walking patterns. Quadriceps weakness is associated with decreased knee flexion angles as well as knee extension moments (Bush-Joseph et al. 2001b). It is believed that individuals with an ACLR walk with less quadriceps contraction to avoid tibial translation, a phenomena known as “quadriceps avoidance” (Berchuck et al. 1990; Andriacchi and Biracchi 1993). In addition, greater hamstring contractions have been observed as well, which is believed to be an effort to pull the tibia backwards to avoid translating the tibia during gait (Solomonow et al. 1987). These adaptations, if not corrected, could lead to altered gait patterns and could reduce the ability to attenuate forces about the knee joint during gait as well as decrease functional performance that increase risk of a second ACL tear (Hart et al. 2010a; Schmitt et al. 2015). On the other hand, some studies have also observed restored quadriceps strength within the year post-surgery (Timoney et al. 1993; Roewer et al. 2011). These conflicting results amongst the research makes it difficult to discern clear conclusions about the implications of ACL injury and reconstruction on strength.

**Proprioception**

Proprioception about the knee provides information about the movement and position of the knee joint. Proprioceptive feedback provides the CNS with information it needs to make adjustments to the environment. Loss of proprioception could result in instability of the joint and poor position sense that could impact joint movement and/or joint moments. Within the first three months post-reconstruction, there are decreases in proprioception (Fremerey et al. 1998). By six months, proprioception is restored at full flexion and extension (Fremerey et al. 2000; Reider et al. 2003). Three years post-reconstructive surgery there are no observable differences in proprioception throughout any point in knee joint motion (Barrett 1991). While reconstruction
and rehabilitation do improve knee joint position sense, proprioceptive training has been shown to improve proprioception as well as quadriceps and hamstring strength and is therefore an important component of a rehabilitation program (Cooper et al. 2005).

Balance

The synchrony between balance, muscle activation, knee joint movement sensation, and joint position allows for dynamic knee stability and function during gait. The use of balance and proprioceptive training programs have been successfully utilized to reduce the risk of initial ACL injury. Immediately following reconstructive surgery there are balance, proprioception, and neuromuscular deficits present in the surgical limb as well as the contralateral healthy limb (Hoffman et al. 1999; Henriksson et al. 2001; Hiemstra et al. 2007; de Fontenay et al. 2015). Balance and proprioceptive training not only improves the control of bodily awareness, it can improve the response time to external forces as well as muscle strength (Hewett et al. 2002; Liu-Ambrose et al. 2003). Within the first couple of months post-reconstruction there continue to be differences in balance and lower limb postural control (Shiraishi et al. 1996; Chmielewski et al. 2002). However, there are those who have found no difference in balance when compared to healthy controls within the first 6 months following surgery (Hoffman et al. 1999; Henriksson et al. 2001; Moussa et al. 2009; Howells et al. 2011). Overall, for any balance and proprioceptive deviations there may have been, the tendency is for these to improve to be comparable to healthy controls by 1 year post-operation (Mattacola et al. 2002; Shi et al. 2010; Angoules et al. 2011; Anderson et al. 2016).

2.2.2. Deficits in functional skills following ACL reconstructive surgery

Spatiotemporal Parameters

It has been observed that while walking, stance time, stride length, cadence and velocity return to normal within a year following reconstruction (Devita et al. 1997; Gao and Zheng 2010). There are instances where altered gait characteristics emerge, however these differences occur when there are asymmetries in limb strength of individuals post reconstruction (Di Stasi et
al. 2013b). These spatiotemporal variables remain similar eight years after surgery (Erhart-Hledik et al. 2017).

**Biomechanical assessment of gait in the frontal plane**

Within the first 12 months following reconstructive surgery, individuals have similar peak knee adduction angles compared to healthy controls (Georgoulis et al. 2003; Webster and Feller 2011; Webster and Feller 2012; Karimi et al. 2013). One study did observe lower peak knee adduction angles in persons with ACLR within the first year, however, this same study found no differences when examined three years post-operation (Wang et al. 2013a; Patterson et al. 2014). Overall the majority of studies have shown no difference in ACLR individuals’ knee adduction angles more than a year after surgery (Bulgheroni et al. 1997; Georgoulis et al. 2003; Webster and Feller 2011; Webster and Feller 2012; Patterson et al. 2014; Kaur et al. 2016).

When looking at the frontal plane moments there seems to be some disagreement across the literature. Some research suggests that there are no differences in knee adduction moments when measured at 6 months, 12 months, or three years after reconstruction (Bulgheroni et al. 1997). There are others who report observing the opposite, finding higher adduction moments in ACLR compared to healthy controls (Butler et al. 2009; Karimi et al. 2013). A meta-analysis by Kaur et al. (2016) found strong evidence indicating significantly lower peak knee adduction moments in the first year and by five years post-reconstruction these moments become similar between the two cohorts (Kaur et al. 2016).

**Biomechanical assessment of gait in the sagittal plane**

Individuals with ACLR have been observed to walk with greater knee flexion angles than healthy controls in the first 10 months following surgery (Ferber et al. 2002; Ferber et al. 2003; Gao and Zheng 2010; Shabani et al. 2015), as well as have decreased range of motion observed at the hip and ankle (Karimi et al. 2013). By the one-year mark, however, these trends seem to improve and the individuals walk with improved knee extension in that limb (Hart et al. 2010a). One year post-surgery, peak knee flexion angles of ACLR persons are similar to healthy controls.
(Lewek et al. 2002; Georgoulis et al. 2003; Hall et al. 2012; Webster and Feller 2012; Noehren et al. 2013). In addition, tibial translation is restored to normal limits within the first year (Wang et al. 2013a). One systematic review and meta-analysis even found peak knee flexion angles to be smaller in the ACLR groups than controls and the contralateral limb (Slater et al. 2017). When looking ahead several years post-reconstruction, the evidence suggests that there are no differences in knee flexion angles between ACLR and controls (Hall et al. 2012; Noehren et al. 2013). However, decreased hip flexion angles are still observed several years later (Noehren et al. 2013).

There is a tendency for increased knee extension moments during the early stance phase of gait during the first three months following surgery (Ferber et al. 2002; Ferber et al. 2003). By six months several research studies have found ACLR persons demonstrate knee extension moments comparable to healthy controls (Webster et al. 2005; Karimi et al. 2013). On the other hand, DeVita and colleagues (1998) observed increased joint torques at the knee as well as at the hip 6 months following surgery (Devita et al. 1998). Research has shown that 10 months after surgery individuals with ACLR have similar knee flexion moments as those found in healthy controls (Timoney et al. 1993; Webster et al. 2005). The increased joint moments within a year post-surgery has been proposed to attribute to the onset of OA (Hart et al. 2010a). Culvenor and colleagues (2015) found evidence of tibiofemoral OA as well as patellofemoral OA within a year following reconstructive surgery (Culvenor et al. 2015).

Within the first couple of years post-operation knee extension moments tend to decrease, with strong evidence indicating that ACLR have lower knee flexion moments than healthy controls (Bush-Joseph et al. 2001a; Hall et al. 2012; Noehren et al. 2013; Zabala et al. 2013). Despite the correction in joint forces within only a couple of years following surgery, the damage may have already be done and OA inevitable. We do see the impact of the surgery affecting other limbs, for instance there tends to be increased hip extension moments at least two years following surgery (Devita et al. 1998; Osternig et al. 2000; Hall et al. 2012). The increased moments at the hip may be a compensatory strategy to reduce tibial shear (Osternig et al. 2000).
Biomechanical assessment of gait in the transverse plane

Within the first year tibial internal and external rotation appears to be similar to healthy controls (Bulgheroni et al. 1997; Georgoulis et al. 2003; Gao and Zheng 2010; Webster and Feller 2011; Wang et al. 2013b; Czamara et al. 2015). In addition, there have been no observed differences in tibial rotation when compared to the contralateral limb either (Webster and Feller 2011; Sato et al. 2013).

When looking at internal and external rotational moments, the literature suggests that there are no differences in the forces produced in the transverse plane for ACLR gait (Karimi et al. 2013).

Long-term implications of ACL reconstructive surgery

Longitudinal studies demonstrate that the reconstructed limb, as well as the contralateral limb show changes over a decade. What is interesting about this study is that the changes seen in the reconstructed limb were also seen in the contralateral limb, however the rate of osteoarthritis developing prematurely in the contralateral limb is 3-fold less than the affected limb (Barenius et al. 2014; Erhart-Hledik et al. 2017). This suggesting that alternate factors other than mechanical changes are playing a key role in the onset and development of PTOA (Erhart-Hledik et al. 2017).

Measures of movement variability

Based on the above information utilizing traditional measures of gait mechanics, individuals with ACLR appear to have normal walking mechanics. A theory as to why there are increases in the joint moments is increased stiffness of the joint after surgery due to a loss in sensory information necessary to maintain normal gait patterns (Fremerey et al. 2000). The absence of a natural ACL may limit a person’s ability to adapt to the inevitable variability innate to everyday walking, thus placing them at an increased risk of degeneration (Moraiti et al. 2010). There is the potential that individuals who rupture the ACL and have a subsequent reconstruction
are limiting the sensory feedback in the knee necessary to maintain normal gait patterns (Fremerey et al. 2000). This could result in changes in the strategies utilized to for joint coordination and knee stabilization, thus producing changes in movement patterns. If this is the case, then traditional gait analysis and strength measures may not provide an analysis sensitive enough to detect changes in coordination strategies post reconstruction.

Irregular movement patterns have been shown in ACL reconstructed gait (Decker et al. 2011). Research has shown that ACL reconstructed gait exhibits more variable and unpredictable knee flexion-extension movement patterns (Moraiti et al. 2009; Moraiti et al. 2010). Further, ACL reconstructed individuals seem to adapt altered gait strategies, illustrating a loss of coordination around the knee joint during gait (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011). These authors suggest that the alterations in gait variability in ACL reconstructed individuals is due to a lack of proprioceptive information that would be typically transmitted with an intact ACL. With this reduced sensory feedback, it is believed that ACL reconstructed individuals may not be able to fully gain function after their knee has been repaired. That being said, there are only three studies that have looked at gait variability in ACL reconstructed individuals (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011), therefore the changes in gait variability as a result of reconstructive surgery remain somewhat allusive. The examination of movement variability provides a unique window into the changes that occur with regards to neuromuscular function after ACL reconstructive surgery. The information gained from analyzing variability could provide insight into the physiological and mechanical adaptations that occur in the ACL reconstructed population.

2.2.3. Neural implications of ACL injury and reconstruction

Injury

The neuromotor system is an intricate network designed to regulate and control motor outputs. The mechanoreceptors in the ACL provide sensory information with regards to position, movement, and loading of the knee (Ochi et al. 2002). Receptors on the spinal cord receive the sensory information and through spinal interneurons. These interneurons function to transmit
excitatory as well as inhibitory signals to other neurons as well as to alpha and gamma motoneurons to allow for adaptations and control of motor function. Following joint injury there are several alterations observed in the neuromotor output. These changes are seen across the central nervous system from muscular to cortical excitability.

Alterations to gait and neuromotor patterns are believed to be the result of intra-articular knee joint effusion (Torry et al. 2000). To assess this, experimental knee joint effusion models have been used to simulate acute injury. Through injecting fluids into the joint capsules of healthy individuals, there is a reduction in the Hoffman reflex as well as reduced muscle activity of the quadriceps muscles (Hopkins et al. 2001). Effusion in the knee joint has also been associated with diminished quadriceps spinal-reflex excitability thus weakened physical function (Spencer et al. 1984; Jensen and Graf 1993; Torry et al. 2000; Hopkins et al. 2001; Palmieri et al. 2004; Palmieri-Smith et al. 2007). This is further substantiated by transcranial magnetic stimulation (TMS), demonstrating altered cortical excitability of the quadriceps following ACL injury, with increased excitability in the cerebral cortex (Héroux and Tremblay 2006). These results suggest a reorganization strategy of the CNS to compensate for the ACL injury.

Changes in voluntary muscle activation are seen following ACL injury. Mechanoreceptors in the joint capsule and/or ligaments stimulated by swelling or structural deformation sends information to the spinal cord (Hopkins and Ingersoll 2000; Hopkins et al. 2001). As discussed earlier, inhibition of neuromuscular input also known as arthrogenic muscle inhibition (AMI), designed to protect the joint after distension or injury by altering the neural input to the surrounding musculature (Snyder-Mackler et al. 1994; Pietrosimone et al. 2012). Once afferent information reaches the spinal cord about the knee joint, modulations both presynaptically and postsynaptically occur in response to injury (Palmieri et al. 2004; Palmieri et al. 2005). The term AMI describes an inability to maximally contract a muscle without structural damage of the muscle or nerve (Hopkins and Ingersoll 2000). After knee joint injury, AMI often manifests in the quadriceps muscles. There is decreased excitation of the quadriceps is accompanied by increased hamstring, hip extensor, and soleus activation, suggesting a compensatory strategy of the system to facilitate in protecting the injured joint (Torry et al. 2000; Palmieri et al. 2004; Ingersoll et al. 2008). This altered lower limb activation pattern resulting
from knee injury is often referred to as quadriceps avoidance pattern (Torry et al. 2000). While AMI aims to protect the joint from further damage, the neural inhibition may create problems for rehabilitation such as muscle weakness (Hart et al. 2010b), atrophy (Lorentzon et al. 1989), and altered neuromuscular control (Ingersoll et al. 2008).

Mechanoreceptors within the ACL and afferent connections can be traced within the central nervous system to the spinal cord as well as supraspinal regions of the brain stem and cerebellum (Park et al. 2005). One theory is that after injury the CNS adapts to the sensorimotor dysfunction through cortical reorganization (Valeriani et al. 1996; Valeriani et al. 1999; Courtney et al. 2005). Evidence of plastic changes in brain activation patterns have been observed in ACL deficient persons roughly two years post-injury (Héroux and Tremblay 2006; Kapreli et al. 2009). When stimulating the common peroneal nerve of the ACL directly after injury to the ACL there is reduced afferent response indicated by reduced somatosensory evoked potentials (SEPs) (Pitman et al. 1992; Ochi et al. 1999; Valeriani et al. 1999; Ochi et al. 2002; Courtney et al. 2005). The loss of afferent information from these areas alters the somatosensory information sent to the cerebral cortex and thus could result in a reduction of position sense (Valeriani et al. 1996; Ochi et al. 1999). Changes in force sensation have also been detected after ACL injury. Individuals with ACL damage demonstrate a reduced acuity in discerning different weights (Héroux and Tremblay 2006). Functional magnetic resonance imaging (fMRI) shows decreased activation in sensorimotor cortical areas and increased activation in supplementary motor areas; i.e. the posterior secondary somatosensory area, presupplementary motor area, and posterior inferior temporal gyrus (Kapreli et al. 2009). Studies utilizing EEG have found changes in cortical activation of individuals who have had an ACLR (Baumeister et al. 2008; Baumeister et al. 2011). These changes are most prominent in the frontal Theta frequency, a signal that plays a major role in working memory functions including attention in cognitive and sensorimotor tasks. Increased power of the frontal Theta frequency has been observed in ACL reconstructed persons during joint position sensing tasks, indicating greater neurocognitive involvement and recruitment to perform the same task as healthy controls (Baumeister et al. 2008; Baumeister et al. 2011). Further, the increased power in frontal Theta frequency does not translate to altered neuromuscular strategies nor altered afferent information in the ACL reconstructed participants, rather the distribution of muscle activity is similar to healthy controls (Baumeister et al. 2011).
The neuroplasticity of the brain allows persons who sustain an ACL injury to adapt and compensate to the damage and change in sensorimotor information, however after reconstruction the CNS pathways appear to remain altered.

Reconstruction

To improve stabilization and physical function of the knee joint after ACL injury, reconstruction is often performed. A key goal of the rehabilitation process is to regain strength of the lower extremity musculature, specifically the quadriceps. That being said, quadriceps strength deficits of 20% persist 6 months following reconstruction (Ingersoll et al. 2008). The belief is that after injury there are neural adaptations that impact neural excitability of the area (Urbach et al. 1999). Improving voluntary quadriceps activation has been positively associated with increasing quadriceps strength (Lepley et al. 2015). Being able to target diminished voluntary activation may lead to improved strength gains in those who are still experiencing quadriceps weakness during rehabilitation. In addition, these people with quadriceps activation failure demonstrate reduced corticomotor excitability (Pietrosimone et al. 2015).

Despite the changes observed in voluntary activation in some individuals post-reconstruction, their spinal reflex excitability returns to normal levels (Kuenze et al. 2015; Harkey et al. 2016). These results contradict the hypothesis of a reflex inhibition contributing to reduced muscle activation. Interestingly, those ACL reconstructed individuals who do regain voluntary quadriceps activation show greater spinal-reflex excitability, believed to be a compensatory neuromuscular strategy to maintain a desired level of function (Pietrosimone et al. 2015). These same individuals who regain voluntary quadriceps activation demonstrate corticomotor responses similar to healthy controls, suggesting neural reparation across systems (Pietrosimone et al. 2015). This is supported by evidence of sensory innervation of reconstructed ACL’s, with detectable SEPs after direct stimulation of the ligament (Ochi et al. 2002). Electroencephalography (EEG) measures have shown that after reconstruction force sensation returns to normal and the distribution of motor output in the reconstructed limbs quadriceps muscle is comparable to healthy controls (Baumeister et al. 2008; Baumeister et al. 2011). That being said, there have been observed changes in the cortical activation patterns. Baumeister and
colleagues (1999, 2008) found greater activation in the frontal lobe of ACL reconstructed individuals, reflecting an increased focus of attention during force tasks. ACL injury is a deafferentation injury, suggesting modifications and the reorganization of CNS pathways. While ACLR improves the mechanical stability of the knee joint it does not seem to reverse the impact it has on the CNS (Valeriani et al. 1996; Valeriani et al. 1999; Courtney et al. 2005; Kapreli et al. 2009). The neuroplasticity of the brain allows persons who sustain an ACL injury to adapt and compensate to the damage and change in sensorimotor information, however after reconstruction the CNS pathways remain altered. With that in mind, perhaps rehabilitation strategies should be revised to accommodate neuromuscular strategies to reduce long-term implications for ACL reconstructed individuals.

While much of research on ACL reconstructed individuals has focused on the local consequences of this surgery, neurophysiological changes post reconstructive surgery may also provide insight as to the reason for the increased rate of re-injury. An understanding the pathways of the CNS that afford us the ability to maintain postural control and coordinate locomotor tasks in healthy adults is paramount in understanding whether there are neurophysiological alterations that occur after ACL injury.

2.3. NEUROMOTOR CONTROL FOR GAIT AND BALANCE

2.3.1. Neural pathways for gait and balance

Walking, or any form of human locomotion for that matter, is one of the most automatic of voluntary movements. The basic rhythm underlying locomotion is derived from two systems, the central pattern generator (CPG) and the spinal cord. The CPG is a neuronal network that can act independently from sensory input generated by the central nervous system (CNS) (Grillner and Zangger 1975). While the neuronal identity of the CPG remains unclear, evidence suggests that the CPG is a result of peripheral input from muscle reflexes onto the spinal cord that transform these commands into a rhythmic stepping pattern (Sherrington 1910; Hiebert et al. 1996; Guertin 2009). The regulation and control of locomotion is provided through supraspinal input.
The brainstem integrates visual and vestibular information with somatosensory input for initiating and controlling locomotor movements. Visual information from the motor cortex enables smooth, coordinative movements while vestibular information regulates balance with information about the position and movement of the head (Gibson 1958; Fernandez and Goldberg 1976; Fitzpatrick et al. 1999). Combined with somatosensory input, this information also enables adaptability of locomotor movements (Kennedy et al. 2003).

When stimulated, the mesencephalic locomotor region (MLR), located in the midbrain, sends a signal that activates the CPG to initiate movement and control speed (Jordan 1998; Jordan et al. 2008). The speed of gait is related to the amount of electrical stimulation to the MLR, low stimulation produces slower walking patterns and high stimulation produces running patterns (Grillner et al. 1997). The MLR projects to the reticulospinal neurons that relay the signal to descending locomotor pathways to initiate the CPG.

To regulate movement patterns, information from the CPG and somatosensory receptors is sent through the spinocerebellar pathways to the cerebellum (Robinson 1995). The cerebellum influences several brainstem nuclei; the vestibular nuclei, the red nucleus, and the nuclei in the medullary reticular formation; that induces the regulation and control of movement. The basal ganglia is responsible for the control of motor planning and programming. Stimulation of the basal ganglia has shown to influence the initiation and termination of voluntary movement, control of posture, as well as the rhythmicity of limbs (DeLong and Wichmann 2007).

The ability to produce locomotor movement allows us to navigate throughout our environment. The integration of information from the spinal cord and the brain enables us to adapt and modify what is otherwise a rhythmic movement pattern. The beauty of this system is that there is not just one area responsible for the production of a stepping pattern or postural control and balance, therefore if one of the supraspinal cortices is damaged we are still able to ambulate.
2.3.2. Muscle activation patterns for locomotion

To achieve locomotor movement, muscle activation is needed to produce the intended motion, as well as muscle activation to prevent unintended motions. This complex organization of muscle contractions and the rhythmicity provided via the CPG are what allow for ambulation (Grillner and Zangger 1975). Gait can be broken up into two phases: the swing phase and the stance phase. The swing phase during gait is the commencement of the flexion phase, when the hip, knee and ankle are all flexed. In anticipation of the foot initiating stance, the knee and ankle extend midway through the swing phase. This places the foot in front of the body, enabling the limb to take on the weight of a person. As the body moves onto the grounded leg in early stance phase the knee and ankle begin to move into flexion once again, however because bodyweight is being transferred to that limb there is a strong contraction of the extensors allowing the body to move over the foot. Towards the end of stance the hip, knee, and ankle all extend to provide the propulsion needed to keep ambulating. Though there are moments during gait that extensors or flexors are the predominant movements, during all four phases there is constant flexion and extension of joints occurring to maintain postural balance and control of movements (Grillner and Zangger 1975). The timing and amplitude of flexor and extensor activation allows for the achievement of locomotion (Sherrington 1910; Hiebert et al. 1996).

Proprioceptors within the muscles, tendons, ligaments, and joints provide feedback about movement, specifically with regards to internal force generation and bodily orientation. Proprioceptive information about movement direction, location, velocity, and muscle activation is sent to the central nervous system via afferent pathways (Deliagina et al. 2006). This information is then used to adjust the body or limbs to maintain control of the body during motion. Proprioceptive information plays a considerable role in coordination and the control of movement. The input of proprioception information helps regulate locomotor movements by adapting to expected and unexpected changes in the environment (Kavounoudias et al. 1999). The proprioceptive mechanoreceptors found in muscles, ligaments, and tendons help to prevent injury by inhibiting muscles that are stretched too far, and activating the antagonist’s muscles to help in shortening the muscles as quickly as possible (Lephart et al. 1997). They are also responsible in regulating the timing and amplitude of movement. The feedback generated allows
for automatic adjustments of force in the hip, knee and ankle that enables us to transition from stance to swing in response to changes in loading and unloading during gait (Sherrington 1910; Hiebert et al. 1996). While humans are able to perform movement tasks in the absence of proprioceptive information, the precision and timing of movements is altered. Kelso and colleagues (1980) demonstrated this when looking at individuals who have prosthetic metacarpophalangeal joints. While those with the prosthetic joints were able to perform the movements, there was a distance of the movement was less accurate. Kelso et al. (1980) attribute these finding to the lack of natural joint receptors in the metacarpophalangeal joint to modulate the movements more precisely (Kelso et al. 1980). Additionally, when looking at individuals who are deafferented compared to healthy individuals, both are able to perform the task of pointing at a target. However, the deafferented individuals performed the task with more spatial errors, that is to say with less accuracy (Blouin et al. 1993). The findings in the literature indicate that while movement is possible without proprioceptive input, information provided through proprioception allows movements to be performed more accurately.

2.3.3. The influence of sensory information on gait and balance

Sensory receptors provide information to the central nervous system about movement occurring in the body. It is the integration of information provided by the sensory systems that provides a person an external representation of their surroundings (Horak et al. 1990; Peterka 2002). As previously stated, visual, vestibular, and somatosensory inputs allow adaptions in locomotor movements. The visual system provides information about an object’s size, shape, location, and orientation. This information about the surrounding environment provides humans with the ability to navigate obstacles and detect changes to the environment (e.g. walking surface, orientation), to maintain balance during gait. The influence visual information has on one’s balance is evidenced in research by Lee and Aronson (1974). In this research the subjects experienced the illusion of a moving room. What they found was that the subjects adopted a movement strategy that resembled the oscillations of the moving room (Lee and Aronson 1974). This effect emphasizes the impact visual perception has on postural control and the implications it may have on controlling locomotion. Visual information can be provided almost instantaneously due to the rapid propagation of light waves (Patla 1997). This allows for the
system to efficiently interpret the surroundings and take preparatory actions to avoid perturbations (Patla 1991). In gait, for example, visual input plays a significant role in anticipatory planning of foot placement while walking (Paulus et al. 1984). Similarly, evidence suggests that stimulation of the vestibular system provides humans with information that is vital in maintaining postural control during locomotion.

The vestibular system is often referred to as our “sixth sense”, it is sensitive to linear and angular accelerations that provide the CNS with accurate inertial information (Angelaki and Cullen 2008). This system provides the CNS with the orientation of the body with respect to gravity and during head movements (Nashner et al. 1989). When the head rotates, the internal contents within the semicircular canals, the endolymph, places forces on the hair cells found within the canals that sensitive to to angular accelerations. The otolith organs operate in a similar manner as the semicircular canals, they detect linear accelerations during motion as well as static accelerations due to gravity. Displacement of the hair cells in the vestibular system, rendered as the magnitude and orientation of both angular and linear accelerations, are then sent to the CNS to establish an appropriate response (Goldberg and Fernández 1975). This allows people to maintain balance while walking, or if you are an Olympic figure skater allows you to maintain balance while performing a camel spin. However, information from the vestibular system alone will not provide the CNS with discernable information. The vestibular system cannot discern linear accelerations due to changes in orientation (e.g. tilting of the head to the left), or translation (e.g. moving to the right) (Jaeger et al. 2002). For that Olympic skater to spin, maintain balance and offset nystagmus they rely on visual input to allow them to maintain control of the movement. Therefore CNS relies on the integration of the sensory systems, such as the visual system, to respond to vestibular information.

The somatosensory system is located throughout the body, with cutaneous, joint, and proprioceptive inputs providing information with regards to balance and orientation. While a loss of somatosensory fibers does not prevent us from being able to ambulate effectively, we have observed changes in balance and postural responsiveness. For example, persons with peripheral neuropathy are at an increased falls risk compared to neurotypical controls (DeMott et al. 2007). Postural responsiveness in individuals with peripheral neuropathy is delayed up to
roughly 25 milliseconds (Inglis et al. 1994). In addition, their ability to scale amplitudes of translational movements is impaired which impacts the CNS ability adjust appropriately to sensory feedback, resulting in adaptations to movement strategies (Inglis et al. 1994; Bunday and Bronstein 2009). Individuals with peripheral somatosensory loss after injury also experience alterations in somatosensory responsiveness. This is observed in persons who have torn their anterior cruciate ligament as well as in individuals who suffer from ankle sprains (Valeriani et al. 1996; Courtney et al. 2005; Hertel 2008). While damage to somatosensory information does not result in total loss of locomotive capabilities, damage to this system does appear to change response strategies. One sensory modality alone does not provide the information the CNS needs to provide optimal postural stability and orientation. It is through the integration of all three systems that the CNS is given an accurate representation of our postural control and what is needed to correct for changes to that stability (Macpherson and Horak 2013).

2.3.4 Coordination of movement

The CNS requires information about the environment to coordinate joints and segments for smooth, goal-directed gait to occur. Coordination involves integrating different degrees of freedom and organizing motor activity to perform as a functional unit (Turvey 1990). To perform a specific movement, all systems needed in the construction of movement (e.g. from the intention, the muscle activity, and information from the CNS) need to coordinate to perform that movement task (Latash et al. 1996). For example, gait requires a sequence of complex muscle patterns that are spatially distributed to synchronize muscle contractions required for walking. Through the study of oscillatory movements we have been able to grasp a better understanding of the coordination of gait. Human gait can be characterized as self-optimizing pattern, oscillating at the most stable and energy efficient coordination pattern (Kelso and Schöner 1988; Holt et al. 1990). Legs have a tendency to oscillate at a 1:1 coordination, and during gait this coordination is found to be most stable when walking at one’s preferred stride frequency (PSF) (Russell et al. 2010).

There are two forms of coordination, absolute coordination and relative coordination. Absolute coordination refers to when two oscillators that are phase and frequency locked,
whereas oscillators in relative coordination maintain a similar phase coupling but a more flexible frequency and phase relationship (Holst 1973). The phase relation between two oscillators is an essential variable in revealing the coordination of synergies. Oscillators will tend to coordinate either inphase or antiphase ($\phi=0$ or $180^\circ$, respectively). In human walking the legs move together in an antiphase fashion; absolute coordination between legs during gait would result in both limbs having a relative phase around $180^\circ$. On that same note, if we looked at the coordination of the left leg and the right arm during gait they would move together in an inphase relation, with a relative phase about $0^\circ$. The stability of the coordination is quantified using the standard deviation of relative phase; as the standard deviation of relative phase increases so does the coupling strength.

Changes in gait coordination are revealed in individuals with chronic pain or post-injury (Hamill et al. 1999; Heiderscheit et al. 2002; Miller et al. 2008; Drewes et al. 2009; Hamill et al. 2012; Herb et al. 2014; Yen et al. 2017). Studies of patellofemoral pain, iliotibial band syndrome and chronic ankle instability have all revealed changes in the pattern of coordination between segments or joints of the involved limb as well as reduced coordination variability (Heiderscheit et al. 2002; Miller et al. 2008; Drewes et al. 2009; Herb et al. 2014; Yen et al. 2017). These altered phase relations have been interpreted as an effort of individuals with either chronic pain or injury to constrain the degrees of freedom to reduce pain (Miller et al. 2008). Once a person with ACLR finishes rehabilitation the idea is that they have a stable knee joint with no pain, therefore the patterns of reduced coordination variability in injured populations should not be applicable to this population. There has been only one study that has examined coordination in ACL reconstructed individuals. Changes in coordination variability have been observed within the involved reconstructed limb, however these changes were not significantly different from healthy controls (Kurz et al. 2005). With one study on coordination in this population, more research is needed to substantiate these results. In addition, the impact of ACLR on between limb coordination has not been examined therefore the extent at which coordination variability during gait is impacted in ACL reconstructed individuals is unknown.
2.3.5 Postural control

Postural control has been defined as the control of balance and orientation of the body’s position in space (Shumway-Cook and Woollacott). Postural control is considered to be a complex motor skill that is a reactive response from the interaction of multiple somatosensory processes and involved several physiological mechanisms (Macpherson and Horak 2013). In order to maintain postural control requires the ability to maintain postural orientation as well as postural equilibrium against acting forces (Horak 1987). Postural orientation relies on sensory information that allows for constant adjustments to maintain balance while postural equilibrium is the coordination of somatosensory strategies while stabilizing ones center of mass in response to external postural disturbances (Horak 2006). Both postural orientation and postural equilibrium are maintained through continuous adjustment and response of the postural control system (Horak 1987).

To maintain dynamic postural stability, the postural system implements a response strategy to preserve postural control. In order to understand how the body maintains control during gait, researchers have represented the trunk as an inverted pendulum that pivots about the hip (Winter 1995). This model has been used extensively throughout the literature to describe the movement patterns of the head, arms and trunk (HAT) segments as they move about the hip to preserve control during standing and walking tasks (Winter et al. 1990; Winter et al. 1993). During standing and walking, the HAT segments have been shown to remain erect, roughly 1.5 degrees away from being completely vertical (Thorstensson et al. 1984; Winter et al. 1990). The torque generated in the vertical direction about the hip are minimal, and therefore these results indicate that the torque resulting from anterior-posterior accelerations are the cause of upper body movement during gait (Winter et al. 1990). While the moments at the trunk are generated by accelerations of the trunk, the hip muscles are activated to produce moments of similar magnitude to maintain postural control of the upper body in the frontal plane (MacKinnon and Winter 1993; Winter 1995). Sagittal plane stability of the upper body is governed by the abductors and adductors of the hip, which control the lateral position of the foot that then dictates the step width of the gait pattern (MacKinnon and Winter 1993). The trunk controls the body’s center of mass during gait, steering the center of mass in the direction of movement with a trunk
roll motion about the hip joint to reduce the angular displacement of the upper body (Thorstensson et al. 1984; Patla et al. 1999). In addition, a compilation of sensory input allows one to navigate and coordinate movement around the environment with as much control as possible (Patla et al. 1999). The trunk serves several of these stabilizing functions during gait, from minimizing the angular displacement of the trunk (Thorstensson et al. 1984), to navigating the direction of the body (Patla et al. 1999), to filtering gait-related oscillations as they travel up the body to the head (Winter 1995).

*Maintaining balance during stance*

The adjustments made by the CNS to maintain postural control are all in an effort to control the body’s center of mass and rotations about the center of mass (Horak and Nashner 1986). The center of mass is a representative point of the body’s total body mass. During stance there are two main goals of the postural control system 1) to maintain support of the center of mass against gravity, and 2) maintain control of the center of mass (Macpherson and Horak 2013). Humans are inherently unstable, therefore we need a manner in which to counteract that instability. A complex pattern of muscle activation from the CNS allows us to produce direction-specific forces to control the center of mass (Horak and Nashner 1986; Winter et al. 1998). This will help us adjust to our intrinsic swaying and subtle movements produced throughout our body.

*Response to disturbances to the postural system*

Disturbances to the postural system can be initiated internally or externally. Several physiological systems are involved in detecting postural changes, relaying the required adjustments needed to offset and subsequently recover from the disturbance (Horak et al. 1990). When balance is disrupted there is an automatic response from the CNS to regain balance and equilibrium. Muscles are strategically recruited to counteract the disturbance to bring the center of mass back to a stable position (Horak and Nashner 1986; Kuo 1995). One strategy is to activate muscles to recover the displaced center of mass and bring it back over the base of support (Horak 1987). Another strategy would be to widen the base of support by taking a step.
Figure 2.3. Reaction time paradigm adapted from Schmidt & Lee (1999).
(McIlroy and Maki 1996). The strategies employed by the CNS allow for changes to occur in the postural system without hindering stability (Buchanan and Horak 2001). If one of these physiological systems is impaired, the CNS utilizes compensatory strategies to maintain postural control (Dieringer 1995; Horak and Hlavacka 2001). While there are several means from which the CNS can provide balance through compensatory means, impairments place an individual at greater risk of instability (Runge et al. 1999; Kim et al. 2009).

2.3.6 Reaction time

The speed at which you can react to a stimulus has been shown to be quite important in sporting events; i.e. how quickly a sprinter can get off the blocks, or how quickly a boxer can avoid a punch. In research, the same measures are often used to evaluate performance of a motor skill, or how quickly information can be processed to produce a response. Response time is the overall time taken to initiate the response to the completion of the movement response (Schmidt and Lee 1988). The response time can be broken down into two separate measures: reaction time, and movement time. Reaction time is a measure used to determine the amount of time it takes to initiate a movement in response to a stimulus. This is not a measure of a movement itself but rather the time it takes from the initiation of a stimulus to the time it takes to initiate movement, for example how long will it take for someone to push down on a button with their index finger after they hear the “go” signal (see Figure 1). Measuring reaction time provides insight into cognitive processing because it measures the time it takes to process the stimulus, decipher the proper response, and then initiate the response (Stelmach and Worringham 1985). Movement time occurs after the reaction time, it is the time from the instigation of a response to the completion of the movement to that response. To gain a better understanding of the processing involved in reaction time, electromyography (EMG) has been utilized to measure the muscle activity during a response.

Types of Reaction Time

There are three types of reaction time scenarios: simple reaction time, choice reaction time, and discrimination reaction time. In simple reaction time there is one stimulus and one
possible response to that stimulus; e.g. a light turns on and the response is to push a button with your index finger. With choice reaction time there is more than one stimulus and each stimulus has a specific response to perform; e.g. when the yellow light turns on the response is to push down with your index finger and when the blue light turns on the response is to push down with your ring finger. Finally, discrimination reaction time has more than one stimulus, however the response is only initiated by one of these stimuli; e.g. the response to push down with your index finger is triggered by the red light, even though there is a blue and yellow light that will turn on as well.

The cognitive difference in simple reaction time and choice reaction time is the information-processing stages, starting from the stimulus and eliciting the intended motor response (Sternberg 1969). With simple reaction time the process requires detecting the stimulus and producing a motor response. In addition, the required movement is known therefore the motor response can be set ahead of the stimulus. When we make the cognitive task more complex, such as adding a choice component, processing the information will take longer. Choice reaction time requires detecting the stimulus as well as identifying what the stimulus represents, then selecting the correct response and executing the response (Smith 1968). Hicks law (or the Hick-Hyman Law) demonstrates that the time it takes for a person to make a decision based on the number of choices provided, the greater the number of choices, the greater the amount of time it will take to make a decision. While this processing may have sequential as well as parallel components, information processing takes longer with a choice reaction time task (Sternberg 1969).

**Reaction Time Reveals Cognitive Processing and Neuromuscular Function**

The ability to respond to stimuli efficiently is essential to navigate every-day life (Lord et al. 2003). This allows us to adapt and correct ourselves to avoid falling, or any situation that may threaten our well-being (Grabiner and Enoka 1995). As previously mentioned, the process of adapting to a stimulus requires processing the stimulus, determining the most effective response, and sending a signal to initiate and perform the desired response. Differences in reaction time have been shown in populations with delays in cognitive processing and/or decreased
neuromuscular function. For example disorders with cognitive processing delays, such as Alzheimer’s disease or Parkinson’s disease, show slower simple and choice reaction times (Gordon and Carson 1990; Hausdorff et al. 2006). In addition, individuals with peripheral neuropathy show a decline in neuromuscular function from decreased sensory function, muscle strength and proprioceptive input (Martyn and Hughes 1997). Studies looking at the effects of peripheral neuropathy on neuromuscular function have found these individuals slower reaction times in both their hands and feet (Morrison et al. 2010; Morrison et al. 2014). The normal aging process reveals declines in cognitive processing, neuromuscular properties, reaction time, and postural reaction times (Vandervoort 2002; McDowell et al. 2003; Tucker et al. 2008; Tucker et al. 2009). Compared to young adults, older adults exhibit slower simple reaction times (Tucker et al. 2008; Tucker et al. 2009) and are a strong predictor of older adults at falls risk (Lord et al. 1991; Lord et al. 1994; Lord and Clark 1996; Lord and Fitzpatrick 2001).

Measures of reaction time have also revealed alterations in sensorimotor function following injury. Individuals with chronic ankle instability (CAI) exhibit delayed responsiveness of the peroneal muscles (Hoch and Mckeon 2014). Chronic ankle instability refers to the condition where after sustaining a lateral ankle sprains, persons experience ankle instability and are prone to repetitive ankle sprains. When the ankle is forced into inversion suddenly the mechanoreceptors send an afferent signal to the spinal cord which in turn sends an efferent signal to the peroneal muscles to contract and protect the ankle from the forced inversion. To measure the reflexive response in individuals with CAI is to observe the timing response of the peroneal muscles. A common method of measuring the reaction time of the peroneal muscles has been to use unanticipated ankle perturbation techniques. These measures have shown conflicting results, some found delays in muscle responsiveness in CAI people and others found no differences (Kavanagh et al. 2012; Hoch and McKeon 2014). These differences may be a consequence of the perturbation techniques used, these methods capture a multitude of somatosensory responses that may mask any neuromuscular deficits specifically at the ankle. One study by Kavanagh and colleagues (2012) looked at the voluntary responsiveness of the peroneus longus and tibialis anterior. To avoid stimuli that may impact the reaction time, the participants were given a visual or audio stimulus from which they were to perform the movement task. The researchers observed that individuals with CAI have a slower response time of the peroneus longus when moving the
ankle to eversion, indicating hindered neuromuscular function of the ankle (Kavanagh et al. 2012). While there have been no studies examining the reaction time of ACL reconstructed individuals, the results from CAI research indicate that neuromuscular function may be hindered.

**Stepping reaction time**

If we increase the complexity of the reaction time task, we see an increase in the time it takes to respond. For instance, the reaction time for tapping a finger will be much quicker than taking a step. When walking the body is tasked with maintaining postural control while 80% of the time our center of mass moves outside of the base of support (Winter 1995). Maintaining postural control requires considerable visuospatial processing, the inclusion of a secondary task during a postural task is more cognitively challenging to process and, subsequently, perform a motor response (Kerr et al. 1985; St George et al. 2007). Taking a step for walking increases the complexity of information processing, requiring the synchronization of information from several systems to be processed to walk (Woollacott and Shumway-Cook 2002). In order to successfully maintain balance while taking a step, a quick response time is paramount (van den Bogert et al. 2002). This is evidenced when looking at the difference in response times between older adults who are at risk of falling and those who are not. Older adults with a falls risk have significantly slower stepping reaction times than those who are not at risk of falling (Lord and Fitzpatrick 2001). A systematic review by Hsu et al. (2012) supports these findings revealing stepping reaction time and secondary tasks requiring the processing of an additional rapid response is highly associated with predicting falls (Hsu et al. 2012). In addition, when participants were given more complex tasks concurrent to the stepping reaction time, such as stepping over an object, the results further demonstrate the difficulty older adults have in coordinating movements as swiftly as young adults. When given a secondary cognitive task alongside performing a walking task, older adults performed poorer on the cognitive task and performed poorer as well as slower in the walking task (Miyake et al. 2000; Verghese et al. 2002; Kressig et al. 2008). Studies have investigated the response of participants when stepping over an object and found older adults had a slower stepping reaction time, those at a high risk of falls an even slower execution time (St George et al. 2007; Sturnieks et al. 2008). Together, these results suggest that
individuals with delays in cognitive processing and/or decreased neuromuscular function may be unable to make proper corrective responses when the system is challenged.
CHAPTER III

EXPERIMENT ONE

COORDINATION STABILITY BETWEEN THE LEGS IS REDUCED AFTER ACL RECONSTRUCTION

3.1 INTRODUCTION

The primary role of the anterior cruciate ligament (ACL) is to stabilize the knee joint during locomotion. When torn, ACL reconstruction (ACLR) is often recommended to restore stability of the knee. Some of the major draw-backs of this surgery are an increased rate of re-injury to the ACL or to the ACL of the contralateral limb, and the exponential rate at which post-traumatic osteoarthritis (OA) occurs in these individuals (Lohmander et al. 2007). Early onset of OA post-ACLR has been considered to be strongly attributed to abnormal gait (Andriacchi et al. 2009), although measures of strength and walking kinematics in people with ACLR exhibit similar patterns one year post-reconstruction as healthy controls (Gao and Zheng 2010; Karimi et al. 2013; Kaur et al. 2016). However, more than a year after surgery, alterations in knee and hip joint moments during walking have been observed in people with ACLR (Butler et al. 2009; Noehren et al. 2013; Kaur et al. 2016). It appears that standard kinematic and spatiotemporal parameters are not sensitive enough to detect modifications. Instead, measures that quantify coordination between body parts (e.g., joints or segments) may be better able to identify differences between healthy controls and individuals with ACLR that contribute to changes in kinetics (Kurz et al. 2005).

Gait coordination of individuals with chronic pain or injury has been found to differ from healthy individuals (Hamill et al. 1999; Heiderscheit et al. 2002; Miller et al. 2008; Drewes et al. 2009; Hamill et al. 2012; Herb et al. 2014; Yen et al. 2017). Studies of patellofemoral pain, iliotibial band syndrome and chronic ankle instability have all revealed changes in the pattern of
coordination between segments or joints of the involved limb as well as reduced coordination variability (Heiderscheit et al. 2002; Miller et al. 2008; Drewes et al. 2009; Herb et al. 2014; Yen et al. 2017). These findings of increased coordination stability of joints within a limb have been interpreted as people with injuries developing an altered and more constrained gait pattern in an effort to reduce pain while walking or running (Miller et al. 2008). While an individual with ACLR is expected to have a stable knee joint without any pain after rehabilitation, changes in coordination between segments of the involved leg have been observed, but no significant difference was observed in coordination variability compared with healthy controls (Kurz et al. 2005). With only a single study on coordination in ACLR gait there is a need for more research. There are also a number of limitations in these studies that should be considered. Firstly, these studies examined coordination within the involved limb and not between limbs. Secondly, gait was performed on a treadmill which we have previously shown can obscure coordination variability effects present in overground walking (Russell et al. 2010; Russell et al. 2016). Thirdly, by only considering one speed these studies did not provide an assessment of the changes in coordination across speeds/frequencies. And finally, the studies were not grounded in a model of rhythmic coordination that has been extensively investigated and supported by research on healthy individuals.

The extended Haken, Kelso and Bunz (HKB) coupled oscillator model has driven much of the research in human coordination of movements by providing specific testable hypotheses which have been empirically verified (Haken et al. 1985; Kelso et al. 1990). According to this model, rhythmic motion of a joint can be considered as an oscillator which is coupled with the rhythmic oscillations of another joint. The following equation can be derived from the model to make predictions about coordination between joints (Haken et al. 1985; Kelso et al. 1990):

\[ \dot{\phi} = \Delta \omega - a \sin \phi - 2b \sin 2\phi + \sqrt{Q} \xi_t, \]

where \( \phi \) is relative phase between two oscillators, \( \Delta \omega \) is the asymmetry (arithmetic difference in the resonant frequency of each oscillator), \( b/a \) represents the strength of the coupling between the oscillators, \( \sqrt{Q} \xi_t \) represents noise, and \( \dot{\phi} \) is the derivative of \( \phi \) with respect to time. This model
makes several predictions that have been supported experimentally (Sternad et al. 1992; Amazeen et al. 1995; Sternad et al. 1995; Amazeen et al. 1996; Sternad et al. 1996; Amazeen et al. 1998). For present purposes we highlight four hypotheses relevant to the current study that have been previously validated in walking. Firstly, oscillators will tend to coordinate either inphase or antiphase ($\phi=0$ or $180^\circ$, respectively). In human walking the legs move together in an antiphase fashion, while contralateral arm and leg move together in an inphase coordination pattern. Secondly, stability of the coordination (quantified inversely by the standard deviation of $\phi$, SD$\phi$) increases as coupling strength ($b/a$) increases. Previously we have shown that coupling strength decreases (SD$\phi$ increases) when moving at stride frequencies/speeds faster or slower than preferred (Russell and Haworth 2014a). Thirdly, when the limbs are asymmetrical (i.e., differ in their preferred movement frequency) the faster limb phase leads the slower limb ($\phi$ deviates from $0$ or $180^\circ$). Finally, the larger the asymmetry between the limbs the less stable the coordination (i.e., SD$\phi$ will increase). These later two hypotheses have been supported when asymmetries have been created between the legs, in healthy individuals, using ankle weights (Russell et al. 2010; Russell et al. 2016).

According to the HKB-model, ACLR could influence coordination between the knees by creating an asymmetry ($\Delta\omega\neq0$) or altering the coupling ($b/a$) between the legs. These two possibilities lead to different specific predictions. If ACLR changes the preferred movement frequency of the involved leg an asymmetry with the uninvolved leg would arise ($\Delta\omega\neq0$), leading to the following predictions for the ACLR group: (1) preferred stride frequency differs from healthy controls, (2) greater deviation in relative phase from $180^\circ$, (3) greater coordination variability (SD$\phi$) especially at slower or faster than preferred speeds (i.e., SD$\phi$ is a quadratic function of gait speed with greater quadratic and constant coefficients). Alternatively, ACLR may reduce the strength of the coupling ($b/a$) between the legs in which case (1) and (2) would not be observed, but (4) greater coordination variability would occur (i.e., SD$\phi$ is a quadratic function of gait speed with greater constant coefficient). These predictions were assessed in the current study by quantifying coordination between the knees while participants walked overground at five different speeds relative to their preferred speed. Additionally, to determine if individual differences in asymmetry or walking speed impacted coordination stability, multiple regression analyses were performed.
3.2. METHODS

3.2.1 Participants

Thirty-four individuals were recruited for the study, which was approved by the local institutional review board and conforms to the Helsinki Declaration. Seventeen of the participants had unilateral ACLR (\(\bar{x}\) age: 23.5 (SD: 2.73) years, height: 172.9cm (SD: 9.07cm), weight: 77.4kg (SD: 13.78 kg)). An additional 17 participants were recruited for the control group and were age, height, and weight matched to the ACL participant group (\(\bar{x}\) age: 25 (SD: 2.44) years, height: 173.2cm (SD: 10.52cm), weight: 75.7kg (SD: 14.97kg)). The inclusion criteria for subjects was that they had no history of neuropathy, and had not sustained an injury to their lower extremity in the past six months. Further inclusion criteria for the control group was that they had no previous history of lower extremity injuries and surgeries. The inclusion criteria for the ACLR group was unilateral ACLR with a minimum of one-year post-surgery. There were no significant anthropometric differences between groups (\(p’s > 0.05\)).

3.2.2 Procedures

Sagittal plane angular knee displacement was collected using two electrogoniometers (Delsys, Boston, MA, USA). Each electrogoniometer was secured over the lateral right and left knee joint and were affixed to the leg using double-sided tape and athletic tape. Electrogoniometers were calibrated with knees in peak extension and knees at 90\(^\circ\) flexion using the Delsys Trigno system (Delsys, Boston, MA, USA). Participants performed three trials of over-ground walking at their preferred speed over a distance of 55 meters. Wireless timing gates (Brower Timing Systems, Draper, UT, USA) were positioned at the beginning and end of the walking path to calculate the participant’s gait speed. The average of the three preferred walking trials was used to determine each individuals’ preferred gait speed (100%). Participants were also instructed to walk at four additional target speeds relative to their preferred: 50, 75, 125 and 150\%. The order of the different speeds was randomized for each participant. Trials outside of 10\% above and below of the intended gait speed were repeated. Each participant repeated a total of 1-4 trials.
Data Analysis

Discrete $\phi$ is a more appropriate method of quantifying coordination than continuous $\phi$, especially when phase plane motion does not approximate a circle and asymmetries may exist between oscillators, as in the case of the knees in the current study (Fuchs et al. 1995). For each stride ($i$) peak knee flexion was detected for both left and right legs. $\phi$ was then computed as the ratio of the time between successive peaks of the two legs ($\Delta t$) to the stride time ($T$) of the dominant (Control) or involved (ACL) leg:

Equation 2:

$$\phi_i = \left(\frac{\Delta t_i}{T_i}\right) \times 360^\circ$$

By multiplying this ratio by $360^\circ$, $\phi$ quantified coordination in degrees based on the unit cycle. Mean $\phi$ quantified the average coordination pattern between the knees, while the standard deviation ($SD\phi$) provided a measure of coordination stability. Additionally, mean stride frequency ($\omega$) was computed as the inverse of average $T$. Twenty-seven trials were excluded from the final data analysis due to corruption of files. However, there was at least two uncorrupted trials for every subject under each condition. All data were analyzed and processed using software developed in Matlab version 7.0 (Mathworks R14, USA).

Hierarchical linear models (HLM) were used to model the relationship between target speed and gait speed, and between the percentage of preferred gait speed and the dependent variables: $\omega$, $\phi$, and $SD\phi$. HLM was implemented using the maximum likelihood method, with group set as a fixed factor and target speed as linear, quadratic or cubic covariate fixed effects. Different models were evaluated which included subject, intercept and target speed or average gait speed as random effects and different covariance matrix structures. The model with the lowest Hurvich and Tsai’s criterion and with significant fixed effects was selected as the most appropriate model. In order to assess if individual differences in gait speed and $\phi$ effect coordination stability, multiple regressions were performed separately for each target speed and group. For this analysis, mean absolute $\phi$ ($|\phi|$) was computed, as the expectation is that
deviations above or below 180° will increase SDϕ. Statistical analysis was performed using IBM SPSS Statistics software (IBM Corporation, Armonk, NY, USA, Version 19), with the significance level set at 0.05.

3.3. RESULTS

All participants were able to walk at the target speeds set as a function of their preferred gait speed (Figure 3.1). A linear increase in speed from the 50% to 150% target speed was observed (Table 3.1), with a range from 41% to 159% of preferred speed. On average, the ACLR group walked significantly slower than the Control group. Walking at different speeds resulted in different stride frequencies that were a quadratic function of speed (Table 3.1, Figure 3.2). In contrast with gait speed, ACLR had no significant effect on walking cadence. Turning to measures of coordination, ϕ was approximately 180° as expected and did not vary significantly with gait speed or between groups (Table 3.1, Figure 3.3). Coordination stability (SDϕ) was a significant quadratic function of gait speed, with greater SDϕ at speeds faster or slower than preferred (Table 3.1, Figure 3.4). Additionally, the ACLR group had significantly greater SDϕ indicating reduced coordination stability.

Multiple regression analyses were used to determine if individual coordination stability can be predicted from gait speed and |ϕ|. For the ACLR group, only |ϕ| was a significant moderate predictor of SDϕ when walking at gait speeds approximately 75, 125 and 150% of preferred (Table 3.2). At these non-preferred speeds, the more a person with ACLR deviated from ϕ=180, the lower their coordination stability (greater SDϕ). In contrast, for the Control group SDϕ was only a function of gait speed for the 50, 100 and 125% target speeds (Table 3.2). The more individuals from the Control group deviated from their preferred speed the lower their coordination stability (greater SDϕ).

3.4. DISCUSSION

The general coordination properties observed in healthy controls and the ACLR groups confirm application of the HKB-model to walking. Coordination between the knees approximated anti-phase (ϕ=180°) and was not significantly influenced by walking speed, as to
be expected for legs that are approximately symmetrical. This differs from a previous study of healthy human walking which observed coordination between the knees deviating from anti-phase in a linear relationship with stride frequency/speed (Russell and Haworth 2014a). Rather than suggesting those participants had a significant leg asymmetry, the results are likely due to the instruction to match the right heel strike with an auditory metronome. Adjustments to correct any deviations from the target cadence would likely be focused at the right heel strike leading to deviations from anti-phase coordination between the legs. Without a metronome the current study found no significant relationship between gait speed and $\phi$. Where these two studies do agree is in the quadratic relationship between gait speed and $SD\phi$, where coordination stability was maximal (minimal $SD\phi$) at the preferred stride frequency, and decreased ($SD\phi$ increased) with faster or slower speeds. This finding can be interpreted as coupling strength ($b/a$) decreasing as speed or stride frequency deviates from preferred. What the current study does show is that without a metronome no asymmetry in coordination is evident but coupling strength between the legs is greatest at the preferred speed and decreases with faster or slower speeds.

The decrease in coordination stability after ACLR can be understood as arising from a reduction in coupling strength rather than an asymmetry between the legs. Contrary to the expectations for an asymmetry, stride frequency (Prediction 1), $\phi$ (Prediction 2), and the quadratic coefficient for the relationship between gait speed and $SD\phi$ (Prediction 3) did not differ significantly between individuals with ACLR and healthy controls. Instead, coordination stability was reduced in the ACLR group compared with the Control group, as expected for a decrease in coupling strength (Prediction 4). These findings suggest that ACLR does not significantly affect the preferred frequency of movement of the impaired leg, and therefore does not create an asymmetry between the legs. Indeed, there was no significant difference in cadence between the groups, although a small decrease in the average walking speed was observed. A slower walking speed without a concomitant change in cadence suggests the individuals with ACLR adopted shorter step lengths than healthy controls. Slower walking speeds with a shorter step length could be an effort to reduce the magnitude of gait related oscillations in the trunk and head which have been found to be altered in individuals with ACLR (Kavanagh et al. 2006a; Armitano et al. 2017). Reduced step length has been observed soon after ACLR surgery, but by eight months significant differences with healthy controls have disappeared (Knoll et al. 2004).
While previous research has failed to find kinematic differences a year after surgery, the use of five speeds, hierarchical linear modeling statistical analyses and measures of coordination in the current study have identified changes in walking after ACLR.

The small reduction in preferred speed achieved through shorter steps could stem from anxiety about the repaired knee, but individuals with ACLR are readily able to walk at different speeds when requested. Also, if the relationship between coordination stability and gait speed was due to the ACLR group simply adopting a slower speed, the quadratic relationship between SDϕ and gait speed would have the same quadratic and constant coefficients, but a greater linear coefficient than the Control group (i.e., the curve in Figure 3.4 would be shifted to the right). Instead, coordination stability of individuals after ACLR is lower across all five speeds tested, and the preferred speed coincides with maximal coordination stability even though it is slower than the preferred speed of healthy controls (i.e., equivalent to a quadratic equation with greater constant, but the same quadratic and linear coefficients as seen by the solid curve in Figure 4). Rather than anxiety leading to a slower preferred walking speed, the speed of maximal coordination stability between the knees is slower in individuals with ACLR, and these individuals freely elect to walk at this reduced speed suggesting they are sensitive to this change (Russell and Haworth 2014b).

The observed changes in walking more than a year after ACLR reveals that deficits remain even after rehabilitation. However, previous assessments of the repaired knee have yet to isolate the source of these deficits. With rehabilitation most individuals regain strength after ACLR (Lewek et al. 2002; Kaur et al. 2016), and muscle strength of the involved leg likely far exceeds requirements for walking speeds tested in the current study. Residual laxity in the knee joint after ACLR could influence mechanical stability of the knee (Ramesh et al. 2005), but by six months after surgery arthrometer measures of the injured knee are not significantly different from the uninjured knee (Muaidi et al. 2009). Similarly, damage to mechanoreceptors in the repaired ACL disrupts proprioceptive information about the knee (Fremerey et al. 2000), however the just noticeable difference in knee rotation position returns to normal values within six months of surgery (Muaidi et al. 2009; Dhillon et al. 2011). If properties such as strength,
Table 3.1. Results of the hierarchical linear modeling analysis for each of the dependent variables. Target speed was the covariate for the dependent variable gait speed, while for all other dependent variables the covariate was the measured percentage of preferred speed walked for each trial. Group was the effect of healthy controls compared with ACLR. Significant coefficients are reported in the table below, all others that were not significant are reported as NS.

<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>Constant (CI)</th>
<th>Group (CI)</th>
<th>Linear (CI)</th>
<th>Quadratic (CI)</th>
<th>p</th>
<th>R²LR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gait Speed</td>
<td>.063 (.042/.083)</td>
<td>.032 (.003/.060)</td>
<td>.012 (.012/.013)</td>
<td>NS</td>
<td>&lt;0.05</td>
<td>1.00</td>
</tr>
<tr>
<td>ω</td>
<td>.29 (.26/.32)</td>
<td>NS</td>
<td>.0079 (.74/.83)</td>
<td>-.000015</td>
<td>&lt;0.000</td>
<td>1.00</td>
</tr>
<tr>
<td>ϕ</td>
<td>179.30 (177.69/180.90)</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>&lt;0.000</td>
<td>NA</td>
</tr>
<tr>
<td>ωCV</td>
<td>.081 (.060/.102)</td>
<td>NS</td>
<td>-.0011 (-.0016/-0.00064)</td>
<td>.0000043</td>
<td>&lt;0.000</td>
<td>0.84</td>
</tr>
<tr>
<td>SDϕ</td>
<td>11.23 (9.95/12.51)</td>
<td>-.43 (-.83/-0.04)</td>
<td>-.13 (-.16/-0.11)</td>
<td>.00066</td>
<td>&lt;0.05</td>
<td>0.96</td>
</tr>
</tbody>
</table>
Figure 3.1. The mean gait speed of the ACLR and Control groups for the five target speeds. Significant best fitting lines are displayed for the ACLR (solid) and Control (dashed) groups. There was a significant difference between the two groups for the mean gait speeds across the five conditions (indicated by *). Error bars indicate one standard error of the group mean gait speed.
Figure 3.2. The mean stride frequency ($\omega$) for ACLR and Control groups plotted against the percentage of the preferred gait speed for each target speed condition. Significant best fitting quadratic curves are displayed for the ACLR (solid) and Control (dashed) groups. There was no significant relationship between $\omega$ and percentage of the preferred gait speed as well as no significant difference between the ACLR and Control groups. Error bars indicate one standard error from the mean $\omega$ and mean gait speed.
Figure 3.3. The mean relative phase ($\phi$) between knees for ACLR and Control groups plotted against the percentage of preferred gait speed for each target speed condition. No significant relationship was observed between $\phi$ and percentage of preferred gait speed. In addition, no significant difference was observed between the groups. Error bars indicate one standard error from the mean gait speed and mean relative phase.
Figure 3.4. The group mean standard deviation of relative phase between knees for ACLR and Control groups plotted against the percentage of preferred gait speed. The significant best quadratic curves are displayed for the ACLR (solid) and Control (dashed) groups. There was a significant difference between the two groups for the mean gait speeds across the target speed (indicated by *). Error bars indicate one standard error from the mean gait speed and SD\(\phi\).
mechanical stability and proprioception of the ACLR leg return to pre-injury levels, then perhaps subtle deficits in motor control of the limb are the source of the findings in the current study.

Given that antiphase coordination is a stable coordination pattern within the speeds of human gait, and without any significant physical differences between the repaired and uninvolved legs, then no significant difference in mean coordination between ACLR and Control groups would be expected. Rather, the findings here of reduced coordination stability point to greater noise in the control of the repaired knee or the coupling with the uninvolved knee. This increased noise could arise from afferent and/or efferent changes. Prior measures of strength, mechanical stability or proprioception may not be sensitive enough to detect deficits or the deficits may reside in the functional control of the knees. The decrease in coordination stability seen after ACLR contradicts the general claim in the literature that coordination stability increases with chronic pain or injury (Ferber et al. 2004; Kurz et al. 2005; Hiemstra et al. 2007).

Several methodological differences could explain the variance in results. The current study examined walking overground, while previous studies have investigated walking on a treadmill. The use of a treadmill has been found to stabilize movements and to obscure differences in conditions or between groups compared with overground walking (Dingwell et al. 2001; Russell et al. 2010; Russell et al. 2016). Additionally, the quantification of coordination and its stability also varied between studies. Continuous $\phi$, used in many previous studies (Miller et al. 2008; Drewes et al. 2009; Hamill et al. 2012), is compromised when computed between two oscillations that do not approximate simple sine-waves, such as lower limb joints during walking (Fuchs et al. 1996). While an alternative, vector coding, reduces this concern it has been shown to be inaccurate at turning points in the rhythmic motion (Wheat and Glazier 2006), but it is these turning points that are important in anchoring coordination (Byblow et al. 1995). To avoid these concerns, coordination was computed as discrete $\phi$ in the current study, as recommended by Fuchs and colleagues (1996) (Fuchs et al. 1996). Perhaps the most critical difference between the current and previous studies is the focus on coordination between the legs (inter-limb) rather than segments or joints within a leg (intralimb). If this is the case, increased coordination stability between the joints of the same leg could occur mechanically through increased activation of bi-articular muscles, which would constrain the motions of adjacent joints. In contrast, decreased stability in inter-limb coordination between the knees may result from greater
Table 3.2. Multiple regression analyses to predict coordination stability (SDφ) from gait speed and/or absolute relative phase (ǀϕǀ). Each regression analysis was performed separately for each group and target walking speed condition.

<table>
<thead>
<tr>
<th>ACL</th>
<th>Target speed (%)</th>
<th>Constant</th>
<th>Gait Speed</th>
<th>ǀϕǀ</th>
<th>P&lt;</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>6.91*</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NA</td>
</tr>
<tr>
<td>(5.42/8.40)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>75</td>
<td>2.89*</td>
<td>NS</td>
<td>0.51 (.67)*</td>
<td>.01</td>
<td>.40</td>
<td></td>
</tr>
<tr>
<td>(1.18/4.61)</td>
<td></td>
<td></td>
<td>(0.18/0.84)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td>4.22*</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NA</td>
</tr>
<tr>
<td>(3.49/4.96)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>125</td>
<td>4.12*</td>
<td>NS</td>
<td>0.12 (.66)*</td>
<td>.01</td>
<td>.40</td>
<td></td>
</tr>
<tr>
<td>(3.55/4.68)</td>
<td></td>
<td></td>
<td>(.04/.19)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>150</td>
<td>6.47*</td>
<td>NS</td>
<td>-0.12 (-.57)*</td>
<td>.05</td>
<td>.27</td>
<td></td>
</tr>
<tr>
<td>(5.48/7.46)</td>
<td></td>
<td></td>
<td>(-.23/-0.01)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Control</th>
<th>Target speed (%)</th>
<th>Constant</th>
<th>Gait Speed</th>
<th>ǀϕǀ</th>
<th>P&lt;</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>9.63*</td>
<td>-5.42 (-.54)*</td>
<td>NS</td>
<td>.05</td>
<td>.33</td>
<td></td>
</tr>
<tr>
<td>(5.92/13.34)</td>
<td></td>
<td>(-10.52/-0.32)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>75</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
</tr>
<tr>
<td>100</td>
<td>NS</td>
<td>3.22 (.51)*</td>
<td>NS</td>
<td>.05</td>
<td>.21</td>
<td></td>
</tr>
<tr>
<td>(0.12/6.33)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>125</td>
<td>NS</td>
<td>4.38 (.60)*</td>
<td>NS</td>
<td>.05</td>
<td>.32</td>
<td></td>
</tr>
<tr>
<td>(1.07/7.69)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>150</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
<td>NS</td>
</tr>
</tbody>
</table>

(*)Significant predictor of coordination stability.
noise in the control of the affected limb and its coordination with the unimpaired limb, leading to lower coupling strength.

While the current literature suggests individuals with ACLR return to a normal kinematic gait pattern, the high rate of re-injury, injury of the intact ACL or the development of OA indicate that underlying problems remain in at least some members of this population. Here, we have identified that after ACLR coordination stability is reduced, but we also found evidence of individual differences amongst the ACLR group. Persons with ACLR who walked with greater deviation from anti-phase coordination (|\phi-180|), indicative of greater asymmetry between their legs, walked with lower coordination stability. This was especially evident at non-preferred walking speeds. In contrast, the coordination stability of individuals in the Control group could be distinguished by their walking speed, with coordination stability decreasing the more gait speed differed from the individual’s preferred. The findings for the ACLR group suggest that some individuals did present asymmetry between the legs which resulted in further decreases in coordination stability. Whether these individual differences might identify those more likely to re-injure, injure the intact ACL or develop OA will require further research.

3.4.1 Study Limitations

The current study was designed to assess the applicability of the HKB-model to coordination between the knees for individuals with ACLR during overground walking. While this provided the capability to examine various speeds of overground walking over a long distance, this did also lead to some limitations that should be considered. Future research should determine if ACLR effects differ between inter-limb and intralimb coordination, by investigating multiple joints of both legs. Anterior tibial shear is considered the most direct loading mechanism on the ACL therefore the current study assessed sagittal plane coordination. Future studies should investigate rotational stability through the inclusion of coronal and transverse plane coordination. Given that hand dominance has been found to influence upper limb coordination (Treffner and Turvey 1995), foot dominance should be measured to assess its influence on lower limb coordination. Finally, to understand the effect individual differences after ACLR have on coordination, more participants will be needed with a more comprehensive
assessment of the reconstructed knee, to include measures of proprioception, mechanical stability, leg strength and motor control.

3.5. CONCLUSIONS

This study provided evidence of the applicability of the HKB-model to understanding coordination in the gait of healthy and injured individuals. Both groups walked with the greatest coordination stability between their knees at their preferred speed, while coordination stability decreased at faster or slower speeds. While previous studies have failed to observe kinematic differences in gait between healthy controls and individuals with ACLR, the current study revealed that people with ACLR walk with lower coordination stability. This can be interpreted as decreased coupling strength between the legs. On average, the ACLR group did not display systematic asymmetries between the legs, but some individuals did reveal asymmetries which resulted in further reductions in coordination stability. These changes in coordination after ACLR could contribute to increased risk of re-injury, injury of the other knee, or OA. The current study shows the potential of applying models from the motor control literature to movement disorders.
CHAPTER IV

EXPERIMENT TWO

UPPER BODY ACCELERATIONS DURING WALKING ARE ALTERED IN ADULTS WITH ACL RECONSTRUCTION

4.1. INTRODUCTION

The anterior cruciate ligament plays an important role in stabilizing the knee joint during walking and running activities. Following injury or damage to this ligament that results in a tear, reconstruction is often advised to improve stability of the knee joint. Unfortunately, while this surgery may lead to short-term benefits for maintaining knee stability, there are long term health concerns for persons who undergo anterior cruciate ligament reconstruction (ACLR). In particular, individuals who undergo ACLR surgery demonstrated a prevalence for developing osteoarthritis in the reconstructed knee in less than 12 years post-surgery (Lohmander et al. 2007). One reason for the development of early-onset osteoarthritis relates to changes in ones walking pattern following ACLR (Stergiou et al. 2007; Butler et al. 2009). However, previous reports have struggled to identify clear gait differences for individuals following this surgery. For example, many people with ACLR exhibit similar spatiotemporal parameters (Georgoulis et al. 2003; Gao and Zheng 2010) and gait kinematics (Devita et al. 1998; Gao and Zheng 2010; Webster and Feller 2011) compared to healthy controls. Additionally, similar kinetic and kinematic patterns are observed when compared to the contralateral limb within a single person (Di Stasi et al. 2013b). The result of these collective findings is that approximately 90% of all ACLR individuals appear clinically to have “normal” knee function, defined by similarities in spatiotemporal parameters and mechanical properties of the knee (Ardern et al. 2010). The only consistent difference has been with regard to the force developed during walking, with increases in vertical ground reaction forces (GRF) and joint moments being reported for ACLR persons (Devita et al. 1998).
While the majority of previous studies have focused on the localized effects of this reconstructive surgery (i.e. its effect on the knee joint itself), it is possible that the impact of ACLR could be manifested at other parts of the body. For example, the upper body (i.e., trunk and neck) play an important role in damping gait-related oscillations during locomotion (Pozzo et al. 1991; Winter 1995; Holt et al. 1999) to ensure the head is stabilized (Menz et al. 2004; Kavanagh et al. 2006b). This stabilization is evident in decreased amplitude of acceleration patterns at the head when compared to the lower trunk (Menz et al. 2003; Kavanagh et al. 2005b). Previous research has shown that the ability of the trunk-neck region to effectively control accelerations during walking is affected by both increasing age and for clinical populations (Yack and Berger 1993; Menz et al. 2003; Menz et al. 2004; Kavanagh et al. 2005a; Kavanagh et al. 2005b). If ACLR affects the capacity of the knee joint to appropriately dissipate forces within the lower limb during walking (Moraiti et al. 2010), then it is possible that impact of this surgery may also influence the dynamics of the upper body regions during gait. However, there has been no direct assessment of whether ACLR impacts the ability of the upper body to attenuate oscillations while walking.

This study was designed to assess and compare whether individuals who have had an ACLR demonstrate upper body acceleration patterns similar to healthy subjects. It was predicted that ACLR individuals would demonstrate a diminished capacity to compensate for gait-related oscillations, especially at higher frequencies.

4.2. METHODS

4.2.1. Participants

Seventeen individuals with unilateral ACLR (age: 23.5±2.73 years, height: 172.9±9.07cm, weight: 77.4±13.78kg) and 17 age/height/weight matched control (CTRL) persons (age: 25±2.44 years, height: 173.2±10.52cm, weight: 75.7±14.97kg) participated in the study. There were no significant differences between the two groups with regards to age, height, or weight (p’s>0.05). Inclusion criteria for the CTRL group included no previous history of lower extremity injuries, surgeries, or neuropathy. For the ACLR group inclusion criteria was a
minimum of one year since their last reconstructive surgery (average time since surgery: 4.1±2.6 years). Participants in the ACLR group could have had multiple ACLRs, so long as they were unilateral.

4.2.2. Procedures

Participants performed three trials of over-ground walking at their preferred pace over a distance of 55 meters. Wireless timing gates (Brower Timing Systems, Draper, UT) were positioned to calculate gait speed. Acceleration data were collected using three triaxial accelerometers sampled at 148 Hz using the Delsys Trigno system (Delsys, Boston, MA), positioned on the head, neck, and lower trunk as per our previous research (Morrison et al. 2015).

4.2.3. Data Analysis

Acceleration data were filtered using a second-order Butterworth low-pass filter (cutoff frequency: 20Hz). All data were analyzed and processed using Matlab version 7.0 (Mathworks R14). The following analyses were performed:

*Time Domain Analyses:* The root mean square (RMS) of the acceleration signal was calculated to provide a measure of average acceleration amplitude.

*Frequency Domain:* This was performed using Welch's averaged, modified periodogram method (window: 512 data points) with the primary measure being the frequency at which the peak power (peak power frequency, PPF) occurred. Based upon previous research, this analysis was performed within three discrete frequency bandwidths (0-3Hz, 3-6Hz and 6-10Hz)(Kavanagh et al. 2005b).

*Signal Regularity:* Approximate entropy (ApEn) was used to provide a measure of the degree of regularity of the accelerometer signals. This analysis produces values ranging from 0-2 with lower values reflecting increased regularity (Pincus 1991).
Figure 4.1. Representative acceleration profiles for the trunk and head during gait. Traces are shown for motion in the anterior-posterior (AP), mediolateral (ML), and vertical (VT) directions.
**Segmental Gain:** An estimation of the degree of attenuation or gain between the trunk-neck and neck-head combinations was determined by applying a transfer function to the RMS, frequency, and ApEn data. These results describe whether there was an overall gain (i.e., positive value) or attenuation (i.e., negative value) of the signal between adjacent segments (James et al. 2014; Morrison et al. 2015). For this analysis, data from the more superior segment were divided by data from the adjacent, lower segment (units: decibels (dB)). For the frequency data, this calculation was performed for each frequency bin as previously described.

4.2.4. Statistical Analysis

Preferred gait speed was analyzed using an independent t-test to determine differences between the groups. Other dependent variables were analyzed using mixed linear models with group as the between-subject factor (i.e., ACLR, CTRL) and segment the within-subject factor (i.e., head, neck, trunk). Least significant difference post hocs were used to determine the specific differences between segments for each direction and between groups at the same segment. Statistical analyses were performed using SAS statistical software (SAS Institute Inc., Release 8.0), with the significance level set at 0.05.

4.3. RESULTS

4.3.1. Descriptive Gait Data

There was no significant group difference for preferred gait speed between participants (ACLR: 1.24±0.15meters/sec, CTRL: 1.31±0.16meters/sec; \( t(132)=1.54, p=0.133 \)). An example of the typical acceleration profile for the head and trunk segments across the AP, ML and VT directions is shown in Figure 4.1.

4.3.2. Anterior-Posterior (AP) Accelerations

*RMS acceleration:* Figure 4.2, which contains mean RMS values for each segment across the two groups, illustrates that the gait-related accelerations decreased from the trunk to the head.
Figure 4.2. Bar graphs illustrating differences in mean RMS values between the ACLR and CTRL groups. Bar graphs in the left column depict the acceleration for the head, neck, and trunk segments in the AP, ML, and VT directions. Plots depicting the pattern of RMS attenuation for the head-neck and neck-trunk segments are also shown (right column). Error bars represent one SE of the mean.
In the AP direction, there was a significant segment effect ($F_{2,64}=4.65, p<.013$). Post hocs revealed that the RMS accelerations were significantly lower at the head and neck than the lower trunk. There was no significant group difference ($p>0.05$).

**Frequency:** The frequency profile in the AP direction was characterized by a prominent peak around 1-2Hz (coinciding with step frequency) as well as multiple harmonics of successively higher frequencies (Figure 3). A significant segment effect was observed for the PPF across all frequency bins (0-3Hz: $F_{2,64}=3.25$; 3-6Hz:$F_{2,64}=41.36$; 6-10Hz:$F_{2,64}=9.27$, $p$’s<0.05). Within the 0-3Hz and 6-10Hz ranges, post hoc analysis revealed that PPF values decreased from the trunk to the head for both groups. Contrastingly, within the 3-6Hz range, the PPF values increased from trunk to head. Table 4.1 contains the summary results for the frequency analysis.

**Signal Regularity:** There was a main effect of segment ($F_{2,64}=283.67, p<0.001$). Post hoc analysis revealed that ApEn values increased significantly from the trunk to the neck and decreased from the neck to the head. There were no significant group differences ($p>0.05$). Figure 4.4 demonstrates the mean ApEn values for each axis as a function of group and segment.

**Segmental Gain:** A significant segment effect was found for the RMS attenuation values ($F_{1,32}=310.99, p<0.05$). Subsequent analysis revealed that RMS attenuation was significantly greater from the trunk to the neck than from the neck to the head ($p<0.05$). There were no significant group differences ($p>0.05$). Attenuation of PPF showed significant main effects for group ($F_{1,32}=8.40, p<0.05$) and segment ($F_{1,32}=59.62, p<0.05$) between 0-3 Hz. For the segment effect, post hocs revealed that attenuation in the AP axis was greater for the trunk-neck combination compared to the neck-head ($p<0.05$). There were no differences for the other frequency bins (3-6Hz, 6-10Hz, $p$’s>0.05).

For ApEn attenuation, there was a significant segment effect ($F_{1,32}=683.73, p<0.001$) with both groups showing increased gain from the trunk-neck but attenuation from the neck-head. No significant group effect was observed ($p>0.05$).
Figure 4.3. Illustration of the general frequency pattern of the head, neck, and trunk accelerations in the AP, ML and VT directions for a single person from the ACLR group and a single individual from the Control group. Also included are ensemble averages of signal attenuation for the head-neck and neck-trunk values in the AP, ML and VT directions.
3.3.3. Medio-Lateral (ML) Accelerations

*RMS acceleration:* There was a main effect for segment for ML accelerations ($F_{2,64}=98.73, p<0.001$). Similar to the AP results, ML accelerations decreased significantly from the trunk to the head in both groups. There were no significant group differences for this measure ($p>0.05$).

*Frequency:* For ML accelerations, a prominent peak was seen at 1 Hz. A significant group by segment interaction was observed between 0-3 Hz ($F_{2,64}=19.35, p<0.001$). Post hocs revealed the PPF values for the CTRL group were significantly greater at the trunk (1.86 Hz) compared to the ACLR group (1.58 Hz). Additionally, PPF values for the CTRL group were significantly lower at the head (0.62 Hz) compared to the ACLR (0.90 Hz). Between 3-6 Hz, there was a significant segment effect ($F_{2,64}=3.26, p=0.045$) with PPF decreasing significantly from the trunk to the head for both groups. No significant differences were found between 6-10 Hz ($p>0.05$).

*Signal Regularity:* There was a significant main effect for segment ($F_{2,64}=859.26, p<.0001$). Post hocs revealed a significant increase in signal regularity (decreased ApEn) from the trunk to the neck and head across both groups. No significant group differences were found ($p>0.05$).

*Segmental Gain:* A significant segment effect was found for attenuation of ML RMS acceleration signals ($F_{1,32}=34.78, p<0.001$). Post hocs revealed attenuation was greater between the trunk-neck than for the neck-head combinations ($p<0.05$).

For the frequency results, a significant segment effect was seen for gain values between 0-3Hz ($F_{1,32}=59.62, p<0.05$) and 3-6Hz ($F_{1,32}=11.61, p<0.05$). Post hocs showed attenuation was greater for the trunk-neck combination compared to the neck-head ($p<0.05$). No differences were found between 6-10Hz. A significant group difference was found between 0-3Hz ($F_{1,32}=3.95, p<0.05$), with a significantly larger PPF attenuation or the CTRL compared to the ACLR group.
Figure 4.4. Figure depicting differences in mean approximate entropy (ApEn) values between the ACLR and CTRL groups. Values are shown for all three segments (i.e., head, neck, and trunk) and for the three planes of motion (i.e., AP, ML, and VT directions). Error bars represent one SE of the mean.
For ApEn, there was a significant segment combination effect ($F_{1,32}=478.36, p<0.001$) with post hocs revealing greater attenuation from the trunk-neck than for the neck-head complexes. No group differences were observed for any of the ML attenuation results ($p's>0.05$).

**Frequency:** Vertical accelerations were characterized by a peak at 1-2Hz at the step frequency and additional harmonics at higher frequencies (Figure 4.3). There were no significant segment or group differences ($p>0.05$).

**Signal Regularity:** There was a significant group by segment interaction effect for ApEn ($F_{2,64}=10.99, p<0.001$). Post hocs revealed that for both groups, ApEn values were lowest (i.e. more regular signal) for the neck, and increased at the head and trunk (Figure 4.4). Additionally, ApEn values were greater at the head and neck for the ACLR group compared to controls ($p's<0.05$).

**Segmental Gain:** For RMS attenuation, a main effect for segment was found ($F_{1,32}=495.17, p<0.001$) with post hocs illustrating the VT acceleration attenuated from the trunk-neck, but increased from the neck-head. No significant group difference was found ($p>0.05$). Frequency results revealed no significant segmental gain effects ($p's>0.05$).

For ApEn, there was a main effect for segment ($F_{1,32}=478.36, p's<0.001$) with significant attenuation from the trunk to the neck but a significant gain from neck-head. Additionally, there was a significant group difference ($F_{1,32}=7.76, p=0.008$). Post hocs revealed that the CTRL group exhibited greater attenuation compared to the ACLR group.

4.4. **DISCUSSION**

This study was designed to assess and compare the pattern of acceleration for the lower trunk, neck and head for individuals with unilateral ACLR compared to healthy adults. While there were similarities in the general acceleration patterns between the two groups, the ACLR individuals exhibited a diminished ability to attenuate frequency oscillations in the AP and ML directions. Further, the time-dependent structure of the acceleration signal for the head and neck
in the VT direction was more irregular (i.e. higher ApEn) for the ACL reconstructed individuals compared to controls.

4.4.1. Similarities in Acceleration Patterns across Groups

To date, the majority of studies examining changes in gait for persons with reconstructed ACL’s have focused on the effects this surgery has on knee joint function (Ardern et al. 2010; Hart et al. 2010a; Moraiti et al. 2010; Decker et al. 2011; Webster and Feller 2011). However, it is likely that any factor that negatively affects the knee during walking is also likely to influence other parts of the body. Consequently, we were interested in assessing whether the effects of ACLR were more widespread, impacting on the dynamic control of the upper body during walking. While accelerometry has been used in several studies to assess the role of the trunk and neck during gait as a function of age and/or disease (Yack and Berger 1993; Menz et al. 2003; Menz et al. 2004; Kavanagh et al. 2005a; Kavanagh et al. 2005b), there have been no previous reports of whether persons with ACLR exhibit differences in the pattern of acceleration through the trunk-neck-head axis during walking.

The results revealed that there were a number of similarities between the two groups with regards to the acceleration signals for the trunk, neck and head. In particular, the RMS amplitude of the gait-related oscillations showed a similar systematic pattern, decreasing (i.e. attenuating) from the trunk to the head, a result consistent with previous reports (Kavanagh et al. 2005a). This attenuation was most pronounced for acceleration in the AP and ML direction. In contrast, no significant change was found for the amplitude of the VT accelerations from the lower trunk to the head.

Inspection of the frequency profile (Figure 4.3) revealed a prominent peak in the AP, ML, and VT acceleration signals below 3 Hz. Specifically, for AP and VT accelerations at the trunk, the peak was found between 1.2-1.4 Hz (Table 1). For ML accelerations at the trunk, the peak was closer to 2 Hz (1.58-1.86 Hz), but decreased significantly from the trunk to the head for both the CTRL and ACLR persons. These peaks have been shown to reflect the stride and step
frequencies respectively during over-ground walking (Hirasaki et al. 1999; Moe-Nilssen and Helbostad 2004). From

4.4.2. Differences in Acceleration Patterns across Groups

One function the trunk plays during walking is to act as a low-pass filter, serving to dampen or attenuate gait-related oscillations (Kavanagh et al. 2005a). This attenuation process is important in that it minimizes the impact gait-related oscillations can have on head motion, ensuring a stable platform for the visual and vestibular systems (Winter 1995; Holt et al. 1999). Consistent with these findings, the ACLR persons had a diminished ability to account for lower frequency oscillations in the AP and ML directions (as indicated by the attenuation results). More specifically, within the 0-3Hz range, controls exhibited a higher PPF at the trunk compared to the ACLR group (Table 1). However, at the head, the ACLR group had a significantly higher PPF compared to controls. Together these results indicate that the ACLR persons had a reduced capacity to attenuate AP and ML gait-related oscillations within 0-3Hz. These results, it would appear that the general functional properties of the trunk, neck, and head regions of the ACLR person operate in a similar capacity to that of a healthy person.

While both groups exhibited a similar capacity to accommodate vertical gait-related fluctuations in terms of amplitude, there was a notable difference in the pattern of regularity (ApEn) of the VT acceleration signals. Specifically, these accelerations were more complex (higher ApEn) for the ACLR individuals compared to the healthy persons. As increases in signal complexity have been linked with decline in the ability to coordinate and/or smoothly control movement (Hamill et al. 1999; Cortes et al. 2014), this finding may indicate that the ACLR persons were unable to adequately compensate for gait-related accelerations in the vertical direction. Further, it may highlight that even though the primary point of this injury is the knee, the impact of this previous injury (and subsequent surgery) may be more widespread, affecting the trunk’s ability to smoothly control accelerations during walking.
Table 4.1. Summary of results for frequency analysis for the two groups. This table illustrates the frequency at which peak power (PPF) values were observed at trunk and head segments and within the three frequency bins (0-3 Hz, 3-6 Hz and 6-10 Hz). For clarity the values for the neck are not shown as they were not significantly different from the head.

<table>
<thead>
<tr>
<th>Direction</th>
<th>Frequency Range</th>
<th>Segment</th>
<th>Group</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Control</td>
</tr>
<tr>
<td>AP</td>
<td>0-3 Hz</td>
<td>Trunk</td>
<td>1.26 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>1.20 Hz*</td>
</tr>
<tr>
<td></td>
<td>3-6 Hz</td>
<td>Trunk</td>
<td>3.83 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>3.94 Hz*</td>
</tr>
<tr>
<td></td>
<td>6-10 Hz</td>
<td>Trunk</td>
<td>7.40 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>6.76 Hz*</td>
</tr>
<tr>
<td>ML</td>
<td>0-3 Hz</td>
<td>Trunk</td>
<td>1.86 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>0.62 Hz</td>
</tr>
<tr>
<td></td>
<td>3-6 Hz</td>
<td>Trunk</td>
<td>4.63 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>3.57 Hz*</td>
</tr>
<tr>
<td></td>
<td>6-10 Hz</td>
<td>Trunk</td>
<td>7.14 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>6.68 Hz</td>
</tr>
<tr>
<td>Vertical</td>
<td>0-3 Hz</td>
<td>Trunk</td>
<td>1.26 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>1.26 Hz</td>
</tr>
<tr>
<td></td>
<td>3-6 Hz</td>
<td>Trunk</td>
<td>4.24 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>4.43 Hz</td>
</tr>
<tr>
<td></td>
<td>6-10 Hz</td>
<td>Trunk</td>
<td>6.79 Hz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Head</td>
<td>6.78 Hz</td>
</tr>
</tbody>
</table>

* indicates a significant segment difference ($p<0.05$); † indicates a significant group difference ($p<0.05$).
A potential explanation for these group differences may be related to previous reports of changes in the GRF’s generated by persons with reconstructed ACL’s (Herzog et al. 2003; Hart et al. 2010a). Specifically, persons with reconstructed ACL’s have been shown to generate higher impact forces about the affected joint in the AP direction compared to healthy individuals (Hart et al. 2010a). Similarly, it has also been reported that ACLR individuals can generate increased torques about the knee during walking, leading to increased external GRF’s on the entire body (Herzog et al. 2003). For the ACLR person, the consequence of generating larger forces may reduce their ability to adequately attenuate accelerations, leading to a significant increase in higher frequencies throughout the body as shown for our results.

The increases in signal complexity within the upper body in the current study are consistent with previous reports for lower limb gait variability in the reconstructed limb and the intact contralateral limb (Kurz et al. 2005; Moraiti et al. 2010). When assessing differences of ACLR gait using nonlinear measures, the resultant pattern has been reported to be more irregular in ACLR individuals than healthy controls (Moraiti et al. 2010). This decreased regularity for the person with ACLR could reflect the general decline in proprioceptive input due to the absence of a natural ACL (Moraiti et al. 2010).

While the results of the current study highlight important differences between the ACLR and healthy persons with regards to the pattern of upper body accelerations, there are limitations that should be considered when interpreting the current findings. One issue that may influence results would be time since the last ACLR reconstruction. For the current cohort, the average time since surgery was 4.1±2.6 years. It is possible that the upper body acceleration of persons who had their surgery over a longer time period may exhibit differences compared to those who were assessed closer to the time of surgery. Another consideration is whether the number of surgeries would have an impact on the gait-related acceleration patterns.

4.5. CONCLUSIONS

While both groups demonstrated a similar pattern of gait-related accelerations during over-ground walking, subtle differences were found with ACLR individuals demonstrating
increased complexity of VT accelerations and a reduced capacity to compensate for accelerations in the AP and ML directions between segments. These results indicate that trunk-neck-head axes of ACLR persons may have a reduced ability to filter out gait-related fluctuations, which has the potential to negatively impact head control. Together, these findings indicate that the impact of previous ACL damage is not simply localized to the knee joint, but is widespread, affecting upper body control as well.
CHAPTER V

EXPERIMENT THREE

ACL RECONSTRUCTED INDIVIDUALS DEMONSTRATE SLOWER REACTIONS DURING A DYNAMIC POSTURAL TASK

5.1. INTRODUCTION

The anterior cruciate ligament (ACL) is the most commonly injured ligament in the knee joint, accounting for approximately 200,000 injuries per year in the United States (Prodromos et al. 2007; Spindler and Wright 2008). While ACL reconstructive surgery is considered highly successful, allowing many individuals to return to an active lifestyle, the rate of secondary ACL injury incidence is 1-in-17 within the first two years after surgery (Wright et al. 2007; Paterno et al. 2010). Persons with an ACLR have shown a 10-fold increased risk of developing post-traumatic osteoarthritis compared to those who have never sustained a knee injury, with up to 13% developing symptoms of post-traumatic osteoarthritis 10-15 years post-reconstruction (Roos et al. 1998; Gillquist and Messner 1999; Øiestad et al. 2010). While much of research on ACL reconstructed individuals has focused on the local consequences of this surgery, neurophysiological changes post injury may also provide insight as to the reason for the increased rate of secondary injury.

The central nervous system integrates sensory input from throughout the body to enable individuals to perform many everyday actions and movements. Within the knee joint, the ACL not only serves a stability function, but it also provides proprioceptive information regarding the position of the joint in space. Consequently, the loss of a natural ACL not only affects joint stability, but also alters sensory feedback from this structure. For example, ACL deficient individuals who regain pre-injury levels of function have demonstrated altered muscle activation patterns that are believed to be a compensatory mechanism related to their previous injury.
(Courtney et al. 2005). Further, there have been reports that persons with a reconstructed ACL exhibit more irregular, variable walking patterns (Moraiti et al. 2009; Moraiti et al. 2010) as well as a tendency to change their coordination strategies during gait to the point where alterations in coordination about the knee joint emerge (Kurz et al. 2005; Moraiti et al. 2010; Decker et al. 2011; Armitano et al. 2018). Together these results indicate that the lack of sensory feedback provided by a natural ACL leads to subtle changes during a dynamic task. As a result, the absence of an ACL may have greater neuromotor implications than has been considered to date.

A feature of many everyday actions is the ability to respond quickly and appropriately to unexpected stimuli within our environment. Reaction time (RT) is a measure commonly used to assess such an ability (Spirduso 1975). Many studies have utilized seated RT assessments to quantify response time in both the upper and lower extremities (Lord and Clark 1996; Lord et al. 2003). Such measures are able to discern declines in basic cognitive processes of response execution and have been used as a risk factor for falls in older people (Lord et al. 1991; Lord et al. 1994; Lord and Clark 1996). While RT’s are more commonly used within experimental designs to assess changes in processing speed related to age or a specific task constraint (Grabiner and Enoka 1995; Schmidt et al. 2018), they have also been employed to determine the impact of injury on the speed of responses. Previous research has reported that neuromuscular function of the ankle is hindered in persons with chronic ankle instability, and that the RT as well as movement time of the muscle is delayed in these individuals (Kavanagh et al. 2012). Without the sensory feedback provided by a natural ACL, we hypothesized that ACL reconstructed individuals may not be able to fully regain neuromuscular function and may have similar delays in RT as those with chronic ankle instability.

The assessment of RT requires both central and peripheral neurophysiological factors to initiate and control desired movements (Patla et al. 1993). Further, performing rapid RT-type responses under postural conditions provides insight into the ability of individuals to respond quickly and appropriately during a more challenging task. For example, stepping over a curb, climbing stairs, or turning a corner to enter a room are common locomotor tasks that are performed in changing environments and require quick reactions. Changes in sensory,
neuromuscular, and cognitive systems that produce a decline in function have the potential to negatively affect the ability of persons to correctly execute such daily tasks. Several studies have examined postural RT in older adults and reported these individuals exhibited slower response times and a reduced ability to regain postural control as quickly compared to healthy young adults (Lord and Fitzpatrick 2001; Tucker et al. 2008; Tucker et al. 2009). One extension from these findings is that individuals who experience sensory and neuromuscular changes resulting from injury (i.e. ACL damage) may also exhibit reduced reactions during postural tasks. While several studies have focused on the ability to perform proper athletic movements after injury (e.g. jump landing tasks, lateral cutting tasks), RT paradigms have not been commonly used to assess the impact of injury.

The purpose of this study was to examine whether individuals with a history of ACLR exhibit differences in postural reaction time, seated RT, gait, balance, ankle range of motion, patellar tendon reflex latency, and lower limb strength compared to healthy age-matched controls. It was predicted that the individuals with an ACLR would have slower RT responses under postural conditions. Further, we believe there will be no differences found between persons with ACLR and the age-matched healthy controls in terms of lower-limb strength, walking speed, standing balance, and simple seated reaction times.

5.2. METHODS

5.2.1. Participants

Sixteen participants with unilateral ACLR were recruited for this study (age: 29.25±6.86 years, height: 173.43±7.46 cm, weight: 81.46±14.56 kg) in addition to 16 age/height/weight-matched healthy participants for the control group (age: 28.90±6.24 years, height: 170.66±8.91 cm, weight: 74.61±17.01 kg). This sample size is based on a power calculation (Effect size (Hedge’s G)=0.61, α=0.05, 1-β=0.95) from a similar reaction time study that assessed RT in individuals with chronic ankle instability (Kavanagh et al. 2012). Inclusion criteria for the ACLR group required that participants were 2-15 years post-ACL reconstructive surgery (average time since surgery: 8.9±5.97 years) and had no additional history of lower extremity injuries,
surgeries, or neuropathy. ACLR participants could have had multiple ACLRs, as long as the surgeries were unilateral. For this study there was one participant with more than one unilateral ACLR. Inclusion criteria for the control groups included no previous history of lower extremity injuries, surgeries, or neuropathy. All participants provided written informed consent prior to beginning the study.

5.2.2. Procedures

In addition to matching the groups by age, height, and weight, participants were also matched based on physical activity level using the International Physical Activity Questionnaire (IPAQ) (average activity level for the ACLR group: 4443.1±3359.01 metabolic equivalents (MET)-minutes a week, and control group: 4876.36±3380.58 MET-minutes a week (both groups were considered to have high activity level participants on average). Once completed, participants were asked to kick a soccer ball to determine foot dominance. The ball was placed a foot in front of the participant where they were asked to kick the ball straight ahead however, they felt most comfortable. This task was repeated three times. The foot they used most often to kick the ball was determined to be their dominant foot. Assessments were performed on each person’s simple reaction time (seated), postural reaction time, gait, proprioception, joint laxity, balance, ankle range of motion, reflex responses (i.e. patella tendon), and quadriceps strength. The order of the limb assessment was counter-balanced between participants to offset any order affects. Details of each assessment are as follows.

5.2.3. Simple Reaction Time (Seated)

The simple RT responses of each person were assessed under seated conditions. The RT responses were attained for both the upper limb (i.e. index finger) and lower limb (foot). To evaluate RT, all participants responded to a light stimulus by depressing a timing switch. For the upper body seated RT task, participants responded to the stimulus by depressing a switch with their finger. For the lower body seated RT task, the participants responded with the depression of a pedal foot switch placed on the floor with the distal portion of their foot. All of the participants completed five practice trials to familiarize themselves to the protocol. Following the practice
trials, the participants performed 10 recorded trials for both upper and lower limbs bilaterally. The methodology of the seated RT task is based on previous research by Lord and colleagues (1996).

5.2.4. Postural Reaction Time

For this task, individuals performed both simple and choice RT tasks under postural conditions. Participants stood on two force plates (AMTI OR6-6 Force Platform, AMTI, UK) facing a light panel consisting of a right and left light source (see Figure 5.1). The force plate data were used to compute the participants’ center of pressure (COP). Both light sources were located two feet directly in front of the participants on a four-foot tall platform. When one of the lights came on, participants were instructed to take a step forward as quickly as possible with the foot that corresponded to the side the light came from (i.e., When the right light came on, participants were asked to step forward with their right foot and vice versa). A reaction time device was developed to detect when either of the lights turned on. Both the reaction time device and the force plates were synced to Vicon motion capture system (Vicon, Oxford, UK) to capture the onset of the light stimulus as well as the onset of movement. Both simple RT and choice RT tasks were evaluated under the postural RT condition. For the simple RT assessment, participants were told to step forward with the foot being examined for 10 practice trials followed by 10 experimental trials. Selection of the order of the limb being evaluated was counter balanced between individuals.

For the choice postural RT assessment, participants were told to step to the side indicated by the light. The trials for the choice RT task were counter-balanced to either right or left. Participants were unaware of which direction to step until the light for that side turned on. Participant’s performed 10 practice trials followed by 20 recorded trials for the choice RT condition.

The RT data and the COP data were both sampled at 2000 Hz on the Vicon motion capture system. Determination of the postural reaction time was attained using a custom algorithm using both of these data. This algorithm was designed to calculate the time period
Figure 5.1. Illustration of the postural reaction time methodology. The two force plates and the light stimulus are synchronized to Vicon motion capture system to capture the onset of the light stimulus as well as the onset of movement. Movement is represented by changes in COP in the AP and ML directions.
between the onset of the light signal to the first observable change in center of pressure (COP) (see Figure 5.2). Changes in COP in the anterior-posterior (AP) and medio-lateral (ML) directions were both analyzed using the raw COP data. The determination of the onset of COP movement was based upon the COP signal exceeding 2 SDs of a designated threshold as per previous research (Tucker et al. 2009). The threshold value was calculated as average position of COP activity in the first 300 ms of the trial prior to the visual cue.

5.2.5. Gait

Participants performed three trials of over-ground walking under two conditions: their preferred walking pace and a fast walking pace. For the preferred walking pace participants were asked to “walk at your preferred walking pace” for three trials, and for the fast walking pace participants were asked to “walk as fast as possible without breaking into a jog” for the fast walking pace. A 20 ft. Protokinetcs (Havertown, PA) pressure sensitive walking surface provided spatio-temporal gait measures (i.e. velocity, cadence, step length and width) during each trial. Participants walked a total of 28ft. with the Protokinetcs pressure sensitive walking surface positioned in the center of the path, which eliminated capturing the speeding up and slowing down on the mat. Spatiotemporal data were sampled at 150 Hz and processed using the Protokinetcs PKMAS software (Havertown, PA).

5.2.6. Proprioception and Joint Laxity

Proprioception testing was performed using a lower limb alignment task (Lord et al. 2003). Participants were examined in a seated position with their eyes closed. They were asked to align their legs simultaneously on either side of the vertical acrylic board with the inscribed protractor (measurements in degrees). Participants were asked to align their legs on various heights of the board, first for 2 practice trials and then for 5 recorded trials.

To measure the integrity of the ACL, laxity of the knee joint was assessed using a KT-2000 arthrometer (MEDmetric Corp., San Diego, CA). Each participant was assessed bilaterally
using a manual maximum test performed by a certified athletic trainer (Myrer et al. 1996). The displacement of each limb was recorded for a total of three trials.

5.2.7. Balance and Ankle Range of Motion

Each participant’s dynamic balance was assessed using the Y-Balance test for the lower body (Perform Better Europe, München, Germany) conducted according to a published protocol (Plisky et al. 2009). The Y-balance test is an assessment of dynamic stability, where the performance score is based on a relative reach distance that is normalized to leg length. This measure also provides an indication of their risk of injury. Ankle range of motion (ROM) was also assessed using the weight-bearing lunge test. Decreased ankle dorsiflexion is a risk factor for lower extremity injuries. The weight-bearing lunge test is a functional measure of dorsiflexion range of motion and was conducted according to a previously published protocol (Chisholm et al. 2012). Both of these clinical assessments consisted of three trials for each leg.

5.2.8. Reflex Testing

The patellar tendon reflex time was measured to assess the responsiveness of the lower extremity motor neurons. A certified athletic trainer administered all of the reflex tests. A positive test was indicated by the leg extending following the tap to the tendon. A total of three positive tests were recorded for each leg. To determine the latency of the reflex response, electromyographic (EMG) electrodes were placed on belly of the rectus femoris muscle of both the right and left limbs. In addition, an accelerometer was placed on the reflex hammer to determine when the hammer hit the patellar tendon. EMG data and acceleration data were collected using the Delsys Trigno system (Delsys, Boston, MA) at a sampling rate of 2000 Hz through Vicon Nexus software (Vicon, Oxford, UK). EMG data were filtered using a second-order Butterworth low-pass filter (cutoff frequency: 400 Hz).
Table 5.1. Summary of results for strength, clinical, gait (preferred and fast conditions) measures. Values are means ± standard deviation unless otherwise stated. Additionally, the data measures that required both feet were reported only in the “ACL Affected” column.

<table>
<thead>
<tr>
<th>Variable</th>
<th>ACLR group</th>
<th>Control Group</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Affected</td>
<td>Non-Affected</td>
</tr>
<tr>
<td>Strength (N·M/kg)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Concentric 60° per s</td>
<td>169.76±53.43</td>
<td>163±42.85</td>
</tr>
<tr>
<td>Concentric 120° per s</td>
<td>144.45±48.50</td>
<td>152.55±39.08</td>
</tr>
<tr>
<td>Eccentric 60° per s</td>
<td>189.42±54.02</td>
<td>194.99±54.65</td>
</tr>
<tr>
<td>Eccentric 120° per s</td>
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<td>204.37±52.18</td>
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<td>Clinical Measures</td>
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<tr>
<td>KT-2000 (°)</td>
<td>6.51±3.29</td>
<td>5.32±3.27</td>
</tr>
<tr>
<td>Y-Balance (%)</td>
<td>94.17±5.55</td>
<td>94.18±5.36</td>
</tr>
<tr>
<td>Forward Lunge (cm)</td>
<td>9.07±0.24</td>
<td>10.02±0.24</td>
</tr>
<tr>
<td>Proprioception (°)</td>
<td>0.93±0.77</td>
<td>0.96±0.75</td>
</tr>
<tr>
<td>Patellar Tendon Tap (ms)</td>
<td>27.43±0.41</td>
<td>26.19±0.18</td>
</tr>
<tr>
<td>Seated Simple RT</td>
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</tr>
<tr>
<td>Hand (ms)</td>
<td>211.18±34.58</td>
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</tr>
<tr>
<td>Foot (ms)</td>
<td>260.76±32.09</td>
<td></td>
</tr>
<tr>
<td>Gait (Preferred gait speed)</td>
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<td></td>
</tr>
<tr>
<td>Cadence (steps per min)</td>
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<td></td>
</tr>
<tr>
<td>Velocity (cm/s)</td>
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<td></td>
</tr>
<tr>
<td>Step length (cm)</td>
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<td></td>
</tr>
<tr>
<td>Stride width (cm)</td>
<td>9.83±2.56</td>
<td></td>
</tr>
<tr>
<td>Gait (Fast gait speed)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Velocity (cm/s)</td>
<td>203.91±40.58</td>
<td></td>
</tr>
<tr>
<td>Cadence (steps per min)</td>
<td>143.35±20.80</td>
<td></td>
</tr>
<tr>
<td>Step length (cm)</td>
<td>85.40±8.89</td>
<td></td>
</tr>
<tr>
<td>Stride width (cm)</td>
<td>10.19±3.00</td>
<td></td>
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</table>

Values are means ± standard deviation unless otherwise stated. Additionally, the data measures that required both feet were reported only in the ACL Affected column and Control Dominant Limb column. None of these variables were significantly different from the control group.
Figure 5.2. A representative figure depicting the onset of the light stimulus as well as the onset of COP movement in the AP and ML directions (COP signal exceeding 2 SDs of the threshold).
5.2.9. Quadriceps Strength

Bilateral isokinetic strength measures were used to assess strength differences both between limbs (within the same person) and between the ACL and control groups. The order of the limb assessment for each group was counter-balanced between participants to offset any order effects. This assessment was performed using a Biodex Multi-Joint System PRO (Shirley, NY). Concentric and eccentric quadriceps strength was assessed bilaterally using a bilateral isokinetic knee protocol at a low (60° per second) and high speed (120° per second) as per previous research (Willigenburg et al. 2014). Participants were given three practice trials before each condition and for each leg assessed. The testing procedure consisted of five repetitions at each of the given speeds with at least one minute of rest in between trials.

5.2.10. Statistical Analysis

All dependent variables were analyzed using mixed generalized linear models with group (i.e., controls, ACL) as the between-subject factor, and limb as the within-subject factor. For the between-subject factor, the affected limb for the ACLR group was compared to the dominant limb of the control group. The dependent variables analyzed for the group effect included lower limb strength (N-M/kg), seated RT of the hand and foot (sec), postural RT (sec), proprioception (°), dynamic balance (%), ankle dorsiflexion range of motion (cm), walking ability (cadence, velocity, step length, stride width) and reflex latency (sec).

There were two levels for the limb effect analysis; dominant versus non-dominant limbs were compared for the control group, and affected versus non-affected limbs were compared for the ACLR group. For the limb effect, seated RT of the foot, postural RT, balance, ankle range of motion and reflex latency were the variables assessed. All statistical analyses were performed using SAS statistical software (version 9.3, SAS Institute Inc., NC), with the alpha level set at p<0.05.
5.3. RESULTS

Simple RT (Seated): The results of this analysis revealed no significant group differences between the ACL and control group for both the hand ($F_{1,30} = 1.03, p = 0.32$) and foot RT values ($F_{1,30} = 0.47, p = 0.50$). Additionally, there was no limb effect for either group in the simple RT of the foot (ACL: $F_{1,15} = 0.60, p = 0.45$; control: $F_{1,15} = 1.21, p = 0.30$)(Table 5.1).

Postural Reaction Time: Figure 5.3 and Figure 5.4 illustrate the differences between groups under the postural RT conditions. The ACLR group was significantly slower in the AP and ML directions compared to the control group for both the simple (AP: $F_{1,30} = 33.76, p < 0.05$; ML: $F_{1,30} = 7.30, p < 0.01$) and choice RT postural tasks (AP: $F_{1,31} = 4.54, p < 0.05$; ML: $F_{1,31} = 34.3, p < 0.05$). The RT results for the ACLR group were 85 ms (484 ms ±6.17ms) slower in the simple RT task than the control group (399 ms ±1.95 ms) and 105ms slower under the choice RT condition (ACL: 550 ms ± 43 ms, control: 445 ms ± 12.25 ms). There were no main effects for limb or limb-by-group interaction effects (ACL: $F_{1,15} = 0.45, p = 0.51$; control: $F_{1,15} = 0.72, p = 0.41$).

Gait Data: As shown in figure 5.5, during the preferred walking trials, no significant differences between the two groups were found for velocity ($F_{1,31} = 2.09, p = 0.16$), cadence ($F_{1,31} = 0.41, p = 0.53$), step length ($F_{1,31} = 1.65, p = 0.21$), or stride width ($F_{1,31} = 0.37, p = 0.55$). Similarly, no significant group differences were found at the fast walking speed for velocity ($F_{1,31} = 1.90, p = 0.18$), cadence ($F_{1,31} = 1.00, p = 0.33$), step length ($F_{1,31} = 1.18, p = 0.29$), or stride width ($F_{1,31} = 0.81, p = 0.38$).

Proprioception and Joint laxity: The results from the proprioception test revealed no group differences ($F_{1,31} = 0.01, p = 0.92$) (Table 5.1). Additionally, no significant differences were found in ligamentous laxity between the ACL reconstructed group’s affected limb and the control group ($F_{1,31} = 0.33, p = 0.57$) (Figure 5.5).

Balance and Ankle Dorsiflexion Range of Motion: There were no group differences for either dynamic balance ($F_{1,30} = .068, p = 0.80$) or dorsiflexion range of motion weight bearing
Figure 5.3. Bar graphs illustrating differences in mean simple RT (postural) values between the ACL and control groups. The results from the dominant and non-dominant hands were compared between groups for the upper extremity assessment. The simple RT (postural) results for the foot compared the affected leg of the ACLR group to the dominant foot of the control group. No differences were found between limbs in either group nor between groups. Error bars represent one SD of the mean.
Figure 5.4. Bar graphs illustrating differences in mean postural choice RT values between the ACL and control groups. Error bars represent one SD of the mean.
lunge test ($F_{1,30} = 3.57, p = 0.07$). Further, there was no limb effect found in either group for the Y-balance test (ACL: $F_{1,15} = 0.31, p = 0.76$; control: $F_{1,15} = 1.27, p = 0.22$) or forward lunge test (ACL: $F_{1,15} = 0.21, p = 0.83$; control: $F_{1,15} = 0.89, p = 0.38$) (Table 5.1).

Reflex Testing: There were no group differences found for the latency of the patellar tendon tap between the ACLR group and the control group ($F_{1,31} = 1.34, p = 0.27$). Further, no differences were found between limbs of the ACLR group and the control group ($p$'s $>0.05$) (Figure 5.5).

Strength Measures: No group differences were found at the 60° per second speed for either concentric $F_{1,31} = 0.92, p = 0.34$ or eccentric quadriceps strength ($F_{1,31} = 0.18, p = 0.67$) (Table 5.1.). Additionally, no differences were found at the 120° per second condition (concentric: $F_{1,31} = 2.55, p = 0.12$; eccentric: $F_{1,31} = 1.78, p = 0.20$). There were no between limb differences for concentric (ACL: $F_{1,15} = 0.93, p = 0.36$; control: $F_{1,15} = 0.45, p = 0.65$) or eccentric quadriceps strength (ACL: $F_{1,15} = 1.32, p = 0.20$; control: $F_{1,15} = 0.56, p = 0.58$).

5.4. DISCUSSION

This study was designed to assess whether differences in postural reaction time, seated RT, gait, balance, ankle dorsiflexion range of motion, patellar tendon reflex, and lower limb strength for individuals who have had ACL reconstructive surgery compared to healthy age-matched controls. The main finding was that the ACLR group reacted slower than the control group under postural RT conditions regardless of stepping with the affected or unaffected ACLR limb. This slowing of responses was found irrespective of which limb (affected or unaffected) was used for stepping. In contrast, no differences between groups were found for any of the other measures, with the ACL individuals exhibiting no discernable differences in lower limb strength, (seated) RT, proprioception, balance, walking ability and the latency of their reflex responses compared to the healthy controls. These novel findings suggest that only under more challenging postural conditions does the impact of reconstructed ACL on motor responses emerge.
Figure 5.5. Figure depicting mean results for spatiotemporal gait measures, proprioception, knee joint laxity, and patellar tendon reflex. Error bars indicate one standard deviation from the mean.
5.4.1. Slowing of Postural Reaction Time with ACLR

When performing many everyday tasks, the ability to respond quickly and optimally to unexpected stimuli is critical to maintain balance (Kerr et al. 1985; Grabiner and Enoka 1995; St George et al. 2007). This is particularly true for when taking a step, where the ability to execute a quick, accurate response is paramount for ensuring balance and preventing stumbling or falls (van den Bogert et al. 2002). The results of the current study reveal that, for persons with a reconstructed ACL, their postural reaction times were significantly slower than for the control individuals (by about 80 ms). What is compelling about this finding is that this result was observed even though no differences were observed between the groups with regards to their simple RT responses (for either the hand or foot). Consequently, it would seem that the added challenge of performing a dynamic postural task (i.e., having to take a step) led to a slowing of stepping responses under both simple and choice RT situations. Typically, reaction time measures are performed to provide insight as to cognitive processing demands (Stelmach and Worringham 1985), with more challenging tasks leading to slowing of responses. While slower reaction times are traditionally associated with decreased central processing, there were no differences in the seated simple RT task. As a result, declines in cognitive processing speed are unlikely to be the primary underlying reason for the increased latency of responses observed, but rather the task of dynamic, postural conditions.

While assessments of changes in postural reaction time for ACL reconstructed persons has not been previously investigated, this protocol has been used in other contexts. For example, slower RT responses under postural conditions have been shown to differentiate between healthy young and older adults (Tucker et al. 2008) and are linked to increased falls risk (Kavanagh et al. 2005a). Declines in reaction time are a typical feature of the normal aging process (Fozard et al. 1994). This overall slowing has been attributed to a number of reasons ranging from specific neurophysiological changes including changes in neuromotor processing, reduction in neural dopamine receptor binding, and loss of neurotransmitter connectivity, to more deliberate changes in movement strategy including prioritization of accuracy over speed (Wood and Jennings 1976; Fozard et al. 1994), and a more cautious response strategy (van Dyck et al. 2008; Lu et al. 2011). While the risk of falling increases with age, older adults who are considered to be at a high risk
of falling have significantly slower response times than those who are deemed to be of lower risk. Interestingly, the slower responses are not limited to postural reactions, with declines in simple RT (under similar seated conditions) also reported (Lord and Fitzpatrick 2001; Tucker et al. 2009). For both these aging studies and our investigation, it would appear that the slowing of RT responses is magnified when the system is assessed under more challenging (i.e., postural) conditions. Given the lack of any other significant differences, it is unlikely that the reason of the slowing is driven by changes in strength, balance, sensation, reflex latency or joint laxity. Rather, the slowing of stepping response in ACL reconstructed individuals may indicate a more complicated neuromotor response which may involve a tradeoff in movement prioritization of stability over mobility.

Previous reports have shown that the fear of re-injury influences individuals with ACLRs to alter their participation in sports or recreational activities (Kvist et al. 2005). This is also highlighted in studies performed with older adults, particularly in those individuals with an increased fear of falling. Older adults with a fear of falling are observed to have slower postural responses as well as diminished postural stability compared to healthy older adults (Yack and Berger 1993; Kavanagh et al. 2005b; Horak 2006). Further, older adults as well as those with an increased fear of falling demonstrate a greater degree of coupling between the trunk and head (Kavanagh et al. 2005a), revealing an increase in trunk stiffness associated with a more cautious gait strategy (Morrison et al. 2015). While measures of gait from the current study and previous literature report gait variables are similar to healthy controls, a fear of re-injury may elicit similar alterations in coordination strategies to those seen in older adults. Interestingly, our previous research has demonstrated that ACL reconstructed persons exhibit a reduced ability to compensate for gait related oscillations from the trunk to the head, suggesting a similar stiffening of the trunk as older adults (Armitano et al. 2017). Further investigation is needed to examine the potential impact increasing stiffness throughout the body has on the coordination of upper and lower body segments during gait and balance tasks to understand how ACL reconstructed individuals may alter their coordination strategies during movements of this nature.
5.4.2. ACL Injury Affects the Execution of a Postural Task

It is important to highlight that the slowing of reaction responses under postural conditions was not reflected by similar declines in any of the selected neuromotor measures. More specifically, no differences were found between the ACL and control groups with regards to walking ability, knee joint proprioception, joint laxity, balance, dorsiflexion range of motion, reflex latency, and quadriceps strength. This would appear to be in contrast to other studies that have reported changes in one or more of these measures. Within the first six months following an ACLR, these ACL reconstructed individuals have demonstrated decreases in proprioception (Fremerey et al. 1998). Within the same time frame there have been reports that ACL reconstructed individuals exhibit reduced balance and postural control compared to healthy controls (Shiraishi et al. 1996; Chmielewski et al. 2002). In addition, a number of studies have reported quadriceps weakness persisting 6 to 12 months following surgery (Pfeifer and Banzer 1999; Mikkelsen et al. 2000; Bush-Joseph et al. 2001b; Lewek et al. 2002; Hart et al. 2010b). The reason for the differences found in the current study could be time since surgery. Studies that report differences between ACL and control groups tend to be one to two years post-reconstruction (Lewek et al. 2002; Hart et al. 2010a; Hart et al. 2010b) while the average time since surgery of the ACLR group participants in the current study was roughly 9 years post-reconstruction. Long-term assessments, however, report similarities in gait, proprioception, joint laxity, balance, range of motion, reflex latency, and quadriceps strength (Fremerey et al. 2000; Mattacola et al. 2002; Roewer et al. 2011; Hall et al. 2012; Anderson et al. 2016; Kaur et al. 2016; Erhart-Hledik et al. 2017; Hoch et al. 2019). Further, the majority of studies assessing neuromotor or mechanical changes after ACL reconstructive surgery have focused on one or two variables of mechanical function (e.g. strength or gait or range of motion) within a given cohort. While long-term assessments focused on one or two of these variables help direct us to a clearer understanding of the impact of ACL reconstructive surgery, it is only through the comprehensive assessment of this range of factors within the same individuals that we gain a clearer picture of the underlying long-term effects of ACLR.
5.4.3. Limitations

There are limitations that should be taken into consideration when interpreting the findings of this study. The average time since surgery for the current cohort for this study was 8.9±5.97 years post-reconstruction. The onset of post-traumatic osteoarthritis has been shown to occur within 10-15 years after ACL reconstructive surgery. While none of the participants in the current study were diagnosed with any form of osteoarthritis, radiographic images were not obtained to determine the presence of osteoarthritis in the ACLR participants. While these are considerations for future studies, the ACLR participants were age-matched, height-matched, as well as matched for activity level to the participants in the control group. As a result, regardless of the onset of post-traumatic osteoarthritis or not in the ACLR group, individuals who had an ACLR were considerably slower postural reaction times than the healthy control group. Finally, a more in-depth evaluation of the ACLR persons’ fear of re-injury would clarify whether the slower postural reaction times are primarily due to physiological changes, psychological biases, or a combination of both.

5.5. CONCLUSIONS

Overall, the findings of this study highlight that persons with an ACLR demonstrate slower stepping responses under time-challenging postural conditions. However, this slowing of postural reactions was not reflected by any noticeable change in the other biomechanical and neuromuscular measures. One suggestion is that the slowing of stepping response in ACL reconstructed individuals indicate a more complicated neuromotor response where the individual prioritizes stability over mobility. The selective response may, in part, be due to increased fear of re-injury rather than selective neuromuscular changes based muscle weakness, slower reflexes or altered sensory responses.
CHAPTER VI

GENERAL DISCUSSION

6.1. SUMMARY OF EXPERIMENTAL FINDINGS

The general purpose of this dissertation was to examine the neuromechanical deficits in persons with an ACLR compared to healthy controls to determine the implications of this injury and surgery. It was anticipated that the findings would provide a more comprehensive understanding of how ACL injury and subsequent surgery may impact the entire body. It was also expected that the examination of upper body accelerations and inter-limb coupling during gait, and postural reaction time measures would provide insights into the impact an ACL injury has across body systems to ultimately develop better rehabilitation strategies that reduce the risk of secondary injuries. The following section provides a summary of the three experiments that comprise this dissertation.

Chapter 3. Coordination stability between the legs is reduced after an ACLR

This study examined inter-limb coordination of ACL reconstructed individuals and healthy controls while walking at five different gait speeds ranging from 50% to 150% of their preferred walking speed. Coordination between the knees was not significantly different from anti-phase coordination (180° relative phase) for either group across the different speeds. Both groups demonstrated maximal coordination stability when walking at their preferred gait speed (100%), however the ACLR group displayed reduced coordination stability across all five speeds compared to the healthy control group. These findings were interpreted as the ACLR group having reduced coupling strength between the knees than the healthy controls, but no overall significant asymmetry between the legs. However, multiple regression analyses revealed individual differences within the ACLR group. ACLR individuals with more asymmetry between the knee movements (i.e., deviation from 180° relative phase) showed even larger reductions in coordination stability. The results of this study indicate that even after ACLR and rehabilitation
changes in coordination evident in gait could contribute to increased risk of secondary injuries. Individuals with asymmetry between the legs and greater coordination instability may be at even greater risk of injury.

*Chapter 4. Upper body accelerations during walking are altered in adults with an ACLR*

The aim of this study was to examine the anterior-posterior, medio-lateral and vertical acceleration patterns from the lower trunk, neck and head of individuals who have had an ACLR compared to healthy age-matched controls. Both groups demonstrated similar acceleration patterns from the trunk to the head while walking overground. The main findings were that when compared to healthy controls, the ACLR individuals demonstrate a reduced capacity to compensate for high frequency oscillations during gait in the AP and ML directions, as well as increased complexity in the VT accelerations. Taken together, the results of this study indicate that individuals after ACLR are unable to adequately compensate for gait-related oscillations and have a reduced ability to coordinate these accelerations to control movements smoothly. These findings imply that damage to the ACL is not localized to the knee, but rather affects upper body control as well.

*Chapter 5. ACL reconstructed individuals demonstrate a slowing during postural reaction time tasks*

The first two studies found neuromotor deficits present in walking after ACLR and rehabilitation, which do not appear to be isolated to the involved knee. The third study aimed to isolate the source of deficits by investigating differences in neuromotor function of ACL reconstructed individuals and age-matched healthy individuals. This included measures of strength, proprioception, knee joint laxity and reflex latency, as well as performance of more functional tasks including, balance, gait and stepping reaction time. The results revealed no significant differences between groups in strength, proprioception, knee joint laxity, reflex latency response, hand or foot simple reaction time, balance, and gait. During the postural stepping reaction time task the ACLR groups’ responses were significantly slower in both affected and non-affected limbs than the healthy controls, during both simple and choice reaction
time conditions. The findings of this study reveal a latency in the execution of a voluntary postural stepping task in ACLR individuals. Reduced reaction time speeds are often associated with changes in sensory or motor function, however, based on the experimental findings, this latency in a stepping task is not due to differences in muscular strength, position sense, or changes in central processing (i.e., no group differences in hand or foot simple reaction time). Older adults who have an increased fear of falling also present with slower postural reaction times. While the ACL reconstructed individuals may not have a fear of falling, there may be heightened fear of injury or re-injury that is causing them to prioritize stability over mobility. The slowing of the stepping postural reaction time tasks in ACL reconstructed individuals indicates a more complicated neuromuscular response where the priority is of stability over mobility of movement. Our belief is that this response may be due to an increased fear of re-injury rather than changes in strength, proprioception, knee joint laxity and reflex latency, as well as performance of more functional tasks including, balance, and gait.

6.2. SYNTHESIS OF EXPERIMENTAL FINDINGS

The implications of ACL injury followed by reconstructive surgery has been described with regards to the effects on the knee joint itself and the surrounding segments and has not been assessed in a manner that comprehensively describes the impact throughout the body and across several systems. Overall, these studies provide a greater understanding of the impact ACLR has across systems as well as the manifestation of these implications in altered movement patterns. The current study provides new insight into the global effect ACLR has on movement dynamics in persons with an ACLR. With assessments of inter-limb coupling strength, combined with upper body acceleration patterns and postural reaction time measures, there is a greater understanding of the impact ACLR has on a person’s dynamic movements. This understanding will allow future research to be directed at these subtle changes to provide more robust rehabilitation strategies to target these changes to ultimately reduce the sequelae of symptoms after ACL reconstructive surgery.

In general, persons who had an ACLR have similar gait patterns and characteristics as healthy age-matched controls. ACL reconstructed individuals have similar walking patterns as
healthy controls and are both most stable while walking at their preferred gait speed. Further, while walking the trunk acts as a low-pass filter, attenuating gait-related oscillations from the trunk to the head. However, there were some observable differences found when assessing these measures while walking under different conditions. As would be expected, the coordination between limbs of the ACLR groups and the control group was approximately anti-phase and was not influenced by walking at various speeds. However, evidence of reduced coordination stability shown in the ACLR group suggests a decrease in inter-limb coupling strength that has not been previously assessed in this population. These findings are contrary to those traditionally found in persons with chronic pain or injury where gait coordination patterns shown increased coordination stability (Herb et al. 2014; Yen et al. 2017). This has been understood as these individuals are developing a more constrained walking pattern to reduce the instances of pain. A potential explanation for the decrease in coordination stability rather than the increase shown in other injured populations could be 1) methodological differences, where previous research has looked at intra-limb coordination and this study examined inter-limb coordination, and 2) more importantly the differences in these injuries. The outcome after reconstructive surgery is that the knee joint is stable and a main goal of a successful rehabilitation is that the person has no knee pain. The decrease in coordination stability may reside in the functional control of the knee. Alterations in movement control strategies may be the result of changes in the prioritization of control during dynamic tasks as well as deficits in mechanical stability that reduce the ability to walk smoothly.

To navigate everyday life there is a certain amount of variability necessary for adaptability and to produce smooth, purposeful movement. Gait-related acceleration patterns measured from the trunk to the head have been used to describe movement dynamics under postural conditions and to understand dynamic postural control to illustrate propagation of accelerations from the trunk to the head (Kavanagh et al. 2006a). In the AP and ML directions, the ACLR individuals had a diminished ability to attenuate oscillations up the trunk. One of the functions of the trunk during gait is to dampen gait-related oscillations to minimize the impact on head motion (Menz et al. 2004). The reduced attenuation is highlighted by the greater peak power at the head in the ACLR group. Additionally, the VT accelerations were more complex in the ACLR group indicating a reduced ability to coordinate and control movement as well as an
inability to effectively compensate for gait-related oscillations in the VT direction. Previous reports of increased ground reaction forces in ACL reconstructed individuals during walking tasks (Herzog et al. 2003) could explain the reduced ability to adequately attenuate accelerations up the body. Interestingly, the preferred gait speed of the ACLR group was slower than the age-matched controls without changes in cadence. This implies there may be an effort to reduce the magnitude of the impact gait related oscillations. Combined with the reduced inter-limb coupling strength, it is apparent that persons with an ACLR are adapting altered strategies to walking and the impact of this change expands further than just the knee joint and surrounding segments.

Changes in central and/or peripheral neurophysiological function could be the underlying reasons for altered movement strategies during a dynamic task after ACLR. What was revealed in this dissertation is that there were no significant differences in quadriceps strength, knee joint laxity, proprioception, balance, or general gait measures from healthy control participants, revealing no inherent changes in general sensory or motor function at the knee. Given there were no significant differences from healthy controls in latency of the patellar tendon reflex or during hand or foot simple reaction time tasks, indicates there are no general changes in central or peripheral neuromotor processes. Further the lack of differences in the balance assessments or walking assessments indicate there is no evident decline in balance control. The one measure to reveal group differences was postural reaction time, where the ACLR group responded significantly slower than the age-matched controls. This result combined with the reduced coordination stability and attenuation of oscillations of the trunk during gait may indicate some deficits in dynamic stability, which lead the ACLR individuals to prioritize stability over mobility. Postural reaction time tasks require the integration of several systems to execute the movement accurately and smoothly, and while there were no differences when looking at the measures separately, a postural task may show differentiation between the groups because of the greater amount of processing required. The slower response during a postural task may be due to an increased fear of re-injury rather than changes in muscle strength, altered sensory responses or slower reflexes.
6.3. FUTURE DIRECTIONS

The present study identifies subtle differences in persons with an ACLR while performing a dynamic task that are not simply localized to the knee joint. While these individuals are able to walk in a similar manner as healthy controls, there appears to be a divergence in movement strategy that has not been shown in traditional sensory and mechanical measures. However it is of interest to determine the underlying cause of the adjustment in their walking strategy that will help adapt rehabilitation of this surgery to reduce the changes seen years after surgery. One potential underlying adjustment made by these individuals is the recruitment of muscles to perform dynamic movements. Changes in naturally occurring muscle activation patterns may be due to the declines in sensory and motor function after surgery and going through the rehabilitation process. The integration of several systems across the body are required to produce smooth, goal-directed movements and the absence of one of these for a prolonged period of time may alter the natural activation pattern of muscles while walking and running. The activation patterns of the lower extremity muscles during gait have not been quantified in an ACL reconstructed population. To gain a better understanding of changes in movement strategy, quantifying muscle activation may provide evidence of altered recruitment patterns.

While assessments of postural reaction time have not been previously investigated in persons with an ACLR, slower postural reaction times have been observed in older adults. Declines in sensory and motor function as well as neuromotor processing have been shown to be a natural characteristic of aging therefore slower reaction times are to be expected in this population. Older adults at risk of falling, however, are significantly slower than older adults who are not at risk. A fear of falling causes them to adapt a more cautious movement pattern that is evidenced by slower walking speeds, a shorter stride length and wider stride width reflecting their priority of maintaining stability over mobility (Prince et al. 1997). While the group with the ACLR are not what would be considered at risk of falling, there could be apprehension in their movement to reduce the impact of ground reaction forces during a postural task or to reduce the risk of injury. It is well documented that the majority of people who have an ACLR are not able to return to the same activity levels they participated in prior to injury. A fear of re-injury is one
of the common reasons underlying this decline in sport participation (Ardern et al. 2011). Taken together with changes experienced in sensory and motor information, a potential reason underlying the slower postural reaction time seen in the ACLR group could be an apprehensive strategy of the system that prioritizes stability of a movement above other processes. If the change in movement strategy is a reaction that is similar to a fear of falling response, it could be concluded that alteration to movement control strategy was due to apprehension that persists in the performance of dynamic tasks.

Based on the results of this dissertation, the absence of a natural ACL has a more disruptive systemic impact than previously investigated. Another interesting research direction for the future would be to utilize these findings to distinguish the ACL reconstructed individuals who are at an increased risk of re-injury. Individual differences in inter-limb coupling strength may be used as a predictive measure in determining those who may be at a greater risk of re-injury. Further investigation into individual differences in postural control and postural reaction times could lead to the development of a clinical predictive model of the ACL reconstructed persons at risk of a secondary injury. Previous research has shown the ability to maintain upper body coordination stability while walking and the slowing of postural stepping responses are affected by age and are exacerbated with the addition of a fear of falling. A fear of re-injury in ACL reconstructed individuals may elicit similar alterations in neuromotor strategies similar to those seen in older adults with a fear of falling. It would be of interest to determine whether coordination stability, inter-limb coordination, and postural reaction time could be used as biomarkers to predict who is at greater risk of re-injury to the ACL.
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A. SUPPLEMENTAL SOURCES CONSULTED

Experimental Brain Research citation style
**VITA**

Cortney Noel Armitano  
College of Health Sciences  
Norfolk, VA 23529

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<th>Formal Education</th>
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| 2014-2019        | Ph.D. | Old Dominion University, Norfolk, VA  
School of Physical Therapy and Athletic Training  
Thesis Title: The impact of anterior cruciate ligament reconstructive surgery on neuromotor function.  
Major: Kinesiology and Rehabilitation  |
| 2011-2013        | M.S. | University of Rhode Island, Kingston, RI  
Department of Kinesiology  
Thesis Title: Effectiveness of a surfing intervention on children with disabilities.  
Major: Exercise Science  |
| 2006-2010        | B.S. | Campbell University, Buies Creek, NC  
Department of Exercise Science  
Major: Athletic Training  |

**Professional Appointments**

| 2018- Present | Visiting Lecturer in the Athletic Training Post-Professional Program  
School of Physical Therapy and Athletic Training, Old Dominion University, Norfolk, VA  |
| 2014- Present | Research Assistant at the Center for Brain Research and Rehabilitation, School of Physical Therapy and Athletic Training, Old Dominion University, Norfolk, VA  |

**Research and Publications**


