MEASURING CHANGES TO FORCE CONTROL VARIABILITY USING LYAPUNOV EXPONENTS: LOOKING AT THE INFLUENCE OF ANTERIOR CRUCIATE LIGAMENT INJURY AND RECONSTRUCTION AND HIGH PERFORMANCE ATHLETICS

by

Amelia S. Lanier

A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

Spring 2016

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by

Amelia S. Lanier

Approved: _____
Thomas S. Buchanan, Ph.D.
Professor in charge of thesis on behalf of the Advisory Committee

Approved: _____
Charles B. Swanik, Ph.D.
Chair of the Department of Biomechanics and Movement Science

Approved: _____
Kathleen S. Matt, Ph.D.
Dean of the College of Health Sciences

Approved: _____
Ann L. Ardis, Ph.D.
Senior Vice Provost for Graduate and Professional Education

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Signed: _____

Thomas S. Buchanan, Ph.D.
Professor in charge of dissertation

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: _____

Lynn Snyder-Mackler, Sc.D.
Member of the dissertation committee

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: _____

Jill S. Higginson, Ph.D.
Member of the dissertation committee

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: _____

Nicholas Stergiou, Ph.D.
Member of the dissertation committee

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ABSTRACT

Rupture of the anterior cruciate ligament is a common sports related injury that results in significant financial burden. A significant amount of research has been conducted to understand function and recovery after ACL tears. Of particular interest, is understanding biomechanics of running and cutting maneuvers as these tasks lead to a majority of ACL tears. Recently, the use of nonlinear analysis techniques have provided promising results in a number of biomechanical applications. Specific to ACL injury, researchers using maximal Lyapunov exponents found significant changes to kinematic control at the knee. However, nonlinear kinetic changes that may occur after ACL injury have yet to be explored. The goals of this dissertation were to explore the use of nonlinear techniques as they relate to the control of multidirectional ground reaction forces in recreational athletes (uninjured controls), high performance athletes, ACL deficient patients, and ACL reconstructed patients and to evaluate how nonlinear measures relate to both injury mechanism and task performance.

As a part of this dissertation we developed a force control task that utilizes real time visual feedback of a person's force production to understand force control variability, as measured by maximal Lyapunov exponents. For the first two aims the force control task was used to study recreational athletes, high performance athletes, ACL deficient individuals, and ACL reconstructed individuals. The third aim was divided into two parts, one which explored force control variability and ACL injury mechanism, and one which evaluated the relationship between force control variability and task performance. For the first part of aim 3, ACL deficient subjects were divided by injury mechanism, contact versus noncontact and LyE values were

assessed. For the final part of this aim, high performance athletes completed a series of sport-like tasks including a cutting maneuver, a lateral shuffle, and a deceleration task and biomechanical data from these tasks were then correlated with LyE data from the force control task.

From this work we found that recreational and high performance athletes had similar mGRF control in their right and left limbs. Recreational athletes additionally generated similar LyE in both the anterior/posterior (AP) and medial/lateral (ML) directions, while high-performance athletes generated lower LyE values in the ML direction indicating superior mGRF control in that direction. In our exploration of force control variability after ACL injury and reconstruction, we found significantly larger force control variability in both ACL-d and ACL-r subjects when compared to healthy controls and high performance athletes. This deficit in mGRF control may be a significant contributor to ACL re-injury as it may affect the ability to perform different running and cutting tasks. Those who experience a contact injury exhibit mGRF control deficits through LyE measures which suggests that secondary tissue commonly seen in contact injuries correlate with average stance time in all sport-like tasks tested. As LyE values increased, which indicates poor control, average stance time decreased.

Understanding not only how mGRF control changes after ACL injury and reconstruction, but the dynamic interplay between LyE and injury mechanism and LyE and sports performance, helps biomechanists get a more complete picture of force control variability. This work can assist in developing new rehabilitation techniques and training programs that can not only enhance performance but improve injury

recovery. Nonlinear analysis techniques have the potential to be a powerful addition to our current clinical knowledge base.

Chapter 1

INTRODUCTION

1.1 Background

1.1.1 Anterior Cruciate Ligament Injury and Treatment

Anterior cruciate ligament tears are a common sports injury prevalent in young adults. In fact, over the past two decades the number of ACL ruptures in those 18 years of age and younger has been increasing (LaBella, Hennrikus, and Hewett 2014). There are approximately 100,000 ruptures annually in the United States (Frank and Jackson 2005; Prodromos et al. 2007). The most common form of treatment is reconstructive surgery which replaces the torn ligament, overall about 60-90% of those injured undergo surgery while in younger cohorts as much as 98% of patients undergo surgery (Linko et al. 2005; Sanders et al. 2016). ACL reconstruction is the most common treatment and has been shown to reduce knee instability and reduce the likelihood of meniscal injuries (Meunier, Odensten, and Good 2007). Unfortunately, about 25% of those injured, regardless of treatment course, will experience a re-tear in the first year after meeting return to sport criteria (Mark V. Paterno et al. 2012; Wiggins et al. 2016).

After ACL injury subjects experience deficits in strength, ROM, and neuromuscular control. Following injury there are noted reductions to quadriceps strength in involved limb when compared to uninvolved limb (REF). Additionally, those who experience an ACL injury are unable to fully extend their injured knee after

surgical reconstruction. Neuromuscular changes post injury include increased electromechanical delay (De Ste Croix et al. 2015), reduced proprioception in both the sagittal and frontal plane (Muaidi et al. 2009), and increased muscle co-contraction (Rudolph et al. 2001). Physical therapy aims to recover these deficits. In most cases patients receive both pre-surgical and post-surgical physical therapy. Research indicates treating deficits before surgery lead to better outcomes during post-op recovery (Adams et al. 2012). Physical therapy aims to address the many different biomechanical deficits seen after injury progressively as patients meet specified criteria (Adams et al. 2012; Wright et al. 2015). Initially, therapists work to improve range of motion, reduce joint effusion, and improve quadriceps strength. As the patient progresses new exercises and training are included in therapy but only after the patients meet clinical milestones. In conjunction with criterion based rehabilitation protocols, patients are not allowed to return to sports until passing a number of tests both performance based and self-reported. For patients to pass return to sport (RTS) criteria they must be at least 12 weeks post-op, achieve 90% symmetry on all hop tests and quadriceps strength, and report 90% self-assessed function on the global rating score and Knee Outcome Survey of Activities of Daily Living (Adams et al. 2012). In meeting this criteria patients are then released to return to sports participation although this should be approached gradually. Overall, approximately 81% of people return to some form of sport activity, 65% return to their previous level of sport, and 55% return to competitive sports after ACL reconstruction (Ardern et al. 2010). While a majority of patients do return to pre-injury levels of activity additional treatment approaches based on new measurement techniques are valuable in improving outcomes post-surgery.

Even after enduring extensive rehabilitation about 1 in 4 athletes go on to experience a secondary ACL tear (Mark V. Paterno et al. 2012; Wiggins et al. 2016). Those who have experienced an ACL injury and undergone ACL reconstruction are approximately 15 times more likely to experience a secondary injury than an athlete with no history of ACL injuries (Mark V. Paterno et al. 2012). Unfortunately, the high risk of re-injury persists beyond the first year patients return to sports. In the first 24 months post return to sport patients who underwent ACL reconstruction were 6 times more likely to endure the injury again when compared to uninjured athletes (Mark V Paterno et al. 2014). Surprisingly, most athletes experience a secondary injury in the contralateral limb rather than a re-tear (Mark V. Paterno et al. 2012; Mark V Paterno et al. 2014). The high risk of re-injury highlights the need to fully understand patient function and recovery before returning to sports.

Acutely following injury there are well documented deficits to strength, range of motion, and motor control among other impairments. Unfortunately, even after these issues are resolved in rehabilitation joint health may irrevocably impaired. In the long term, 50% of those who have experienced a ACL rupture (whether it be a single tear or multiple tears) will develop post traumatic osteoarthritis (OA) within 10-13 years after their initial injury (Aaron and Racine 2013; Andersson, Samuelsson, and Karlsson 2009). Health costs related to OA add up to approximately 100 billion dollars annually imparting a significant financial burden (London, Miller, and Block 2011; Michaud et al. 2006). OA is a very damaging condition causing joint pain, swelling, and stiffness (Abhishek and Doherty 2013). This disease has also been linked to increased knee joint loading and increased loading frequency (Andriacchi et al. 2006; Buchanan and Manal 2008; Maly 2008) which may further progress the

disease. With the high prevalence of ACL ruptures, re-ruptures, and subsequent development of OA, research to develop optimized rehabilitation is imperative.

1.1.2 ACL Injury and Cutting Maneuvers

A majority of ACL injuries occur through a noncontact mechanism, when there is a combination of full knee extension, internal or external tibial rotation, and knee valgus (Boden, Griffin, and Garrett Jr. 2000; Rochcongar et al. 2009; Serpell et al. 2011). This injury mechanism leads to a high instance of ACL tears that occur during cutting maneuvers. Therefore, sports that involve running and cutting, such as football, soccer, and basketball, see the highest numbers of ACL tears (Hewett, Di Stasi, and Myer 2012; Mark V. Paterno et al. 2012; Rosa et al. 2014). To better understand ACL ruptures much research has been done quantifying the biomechanics of running and cutting maneuvers.

Cutting tasks require precise neuromuscular control in conjunction with intense power generation. During these maneuvers athletes must quickly and suddenly change direction. Athletes perform theses maneuvers to avoid obstacles, which may be unanticipated, or to properly position themselves which may be anticipated. Unanticipated cutting tasks put athletes at particular risk as they are not prepared for the movement. As athletes initially decelerate prior to changing direction they must generate significant anterior/posterior ground reaction forces (Besier and Lloyd 2001). Once deceleration is complete they must then redirect the center of mass hence producing greater medial/lateral ground reaction forces (Besier and Lloyd 2001). As speed and power increases, modulation of these multidirectional forces (mGRF) are crucial in providing proper body support and positioning to protect against injury. Kinetic measures provide insight into the speed and power generated during cutting,

and have been shown to be related to cutting performance. Specifically, greater ML GRF impulse is significantly correlated to greater performance during side step cutting maneuvers (Havens and Sigward 2014a). Elucidating the impact of mGRF control is important in treating injury and preventing re-injury as athletes are required to generate these forces as they return to sports and return to running and cutting.

The noncontact injury mechanism of ACL rupture indicates a motor control deficit may be a contributing factor. Current research indicates significant increases to electromechanical delay induced by fatigue in female soccer players (De Ste Croix et al. 2015). These results suggest athletes may not be able to activate muscles quick enough to stabilize the knee joint to prevent injury. Additionally, motor control deficits are present after ACL injury and reconstruction. Post ACL injury there are significant losses to proprioception in both the sagittal and transverse planes; the inability to sense joint position may prevent protective neuromuscular control strategies (Muaidi et al. 2009). Electromyography has highlighted increased hamstring co-contraction and altered hamstrings timing during gait in those classified as non-copers (Rudolph et al. 2001). A number of rehabilitation protocols have been developed focusing on recovering neuromuscular control. Hewett and colleagues have utilized neuromuscular training during drop vertical jumps to prevent injury prospectively (Hewett, Ford, and Myer 2006). Snyder-Mackler and colleagues have developed perturbation training that helps develop personalized motor control strategies both before and after ACL reconstruction (Chmielewski et al. 2005; Hartigan, Axe, and Snyder-Mackler 2009; Hurd, Chmielewski, and Snyder-Mackler 2006; Di Stasi and Snyder-Mackler 2011). The prevalence of motor control alterations

and neuromuscular focused rehabilitation emphasize the importance of this component in ACL injury, reconstruction, and rehabilitation.

1.1.3 Nonlinear Dynamics and Biomechanics

A number of functional changes occur after ACL injury and reconstruction. Post injury there are significant increases in knee joint laxity (Daniel et al. 2002). In both cadaveric and in vivo studies there are significant increases to tibial translation and rotation (Georgoulis and Papadonikolakis 2003; Hofbauer et al. 2014; Li et al. 2007; Stergiou et al. 2007). Reduced joint stability measured by tibial motion occurs during walking, jumping, and stair climbing (Georgoulis and Papadonikolakis 2003; Hofbauer et al. 2014; Stergiou et al. 2007). There are also significant reductions to coordination variability post ACL reconstruction (Pollard et al. 2014). All of the previously identified deficits were captured using linear analysis tools. However, linear analyses do not capture the subtle time varying changes that may occur from stride to stride or cycle to cycle. These small changes may be crucial to understand recovery as they can indicate neuromuscular health (Decker et al. 2011).

Nonlinear analyses have recently become a popular tool in understanding various biomechanical issues. These analyses include estimations of approximate entropy and maximal Lyapunov exponent. Nonlinear analyses are valuable in their ability to capture subtle time-varying changes within a signal of interest (Harbourne and Stergiou 2009). Of particular interest is the maximal Lyapunov exponent (LyE) which measures the rate of divergence of infinitesimally close trajectories. Research reports the maximum or largest LyE which indicates the maximal divergence of trajectories or alterations to the structure of variability of the signal of interests. LyE analysis has numerous biomechanical applications including understanding gait in

patients with lower limb amputation (Wurdeman, Myers, and Stergiou 2013, 2013; Wurdeman et al. 2014) and diabetic neuropathy (Dingwell and Cusumano 2000; Harbourne et al. 2010).

Research focused on ACL injury indicates that utilizes nonlinear analysis found reduced kinematic control, as measured by LyE calculated for knee flexion during gait, after ACL injury and reconstruction (Decker et al. 2011; Stergiou et al. 2004a; Zampeli et al. 2010). In fact alterations to kinematic control were common to all reconstruction types (Moraiti et al. 2010b). Larger LyE values measured from joint kinematics in these populations indicates an inability to respond to environmental demands or poor control on kinematics (Decker et al. 2011). In order to gain a better understanding of neuromuscular control post ACL injury and reconstruction in a context related to running and cutting kinetics new analyses are needed.

1.2 Specifics Aims and Hypotheses

In an effort to expand upon the existing body of work using nonlinear analysis techniques and address mGRF control after ACL injury and reconstruction this doctoral work contains three aims. The first aim was to established a standard for force control variability in recreational athletes. The second aim defined changes to force control variability after initial injury and reconstruction and in high performance athletes. While the third aim explored the relationship of force control variability to both sport performance and ACL injury mechanism.

1.2.1 Aim 1: Evaluate mGRF force production and control using nonlinear analyses.

A safe novel task was developed where subjects produced multidirectional ground reaction forces (mGRFs) to be used for nonlinear analyses. Healthy uninjured

individuals were instructed to generate force in the anterior/posterior (AP) and medial/lateral (ML) directions continuously and alternately. We evaluated force control variability using force profiles generated during the force control task. These calculations helped establish force control variability standards to be used for further analyses. We hypothesized there would be no difference in force control variability when comparing the right and the left limb. Additionally, we hypothesized larger force control variability in the AP direction when compared to the ML direction due to the musculature of the lower limb.

1.2.2 Aim 2: Explore force control variability as it relates to ACL injury, ACL reconstruction, and high performance athletics.

For this aim those who have experienced ACL injury and also undergone ACL reconstruction performed the force control task as described in Aim 1. Additionally, high performance athletes also completed the force control task. Within both ACL injured (ACL-d) and ACL reconstructed (ACL-r) subjects we hypothesized the injured limb would generate significantly greater LyE in both the AP and ML direction when compared to uninjured limb. Within high performance athletes we hypothesized they would exhibit similar LyE between the right and left limb and between the ML and AP direction. Across groups we hypothesized ACL-d and ACL-r subjects would exhibit significantly greater force control variability when compared to uninjured recreational athletes and high performance athletes. Lastly, we hypothesized those who participate in high performance athletics will have superior force control variability noted by significantly lower LyE values as their training requires running and cutting maneuvers.

1.2.3 Aim 3a: Determine the relationship between force control variability and ACL injury mechanism.

To further understand changes to force control variability as a result of ACL injury and reconstruction we dichotomized ACL-d and ACL-r subjects based on injury mechanism, contact versus noncontact. We then evaluated changes to LyE based on injury mechanism. We hypothesized that, in ACL-d subjects, those who experienced a non-contact injury would have larger force control variability (LyE) when compared to those who have experienced a contact injury. Along the same lines we hypothesized within ACL-r subjects, those who experienced non-contact injuries would exhibit larger LyE when compared to contact injuries. Lastly, we hypothesized that this trend, larger LyE in non-contact subjects compared to contact subjects, would be present in LyE measured from both the AP and ML directions.

1.2.4 Aim 3b: Determine the relationship between force control variability and performance of functional sport like tasks.

For this aim high performance athletes completed a cutting maneuver, a lateral shuffle, and a deceleration task. Using standard motion capture techniques we collected kinematics and kinetics during these tasks. Correlational analysis was used to determine the relationship between LyE values measured from Aim 2 and performance metrics measured during the cutting, lateral shuffle, and deceleration tasks. Performance metrics included both average and standard deviation stance time, medial/lateral ground reaction force impulse (ML GRI), and anterior/posterior ground reaction force impulse (AP GRI). For the lateral shuffle and cutting task we hypothesized that average ML GRI, standard deviation ML GRI, average stance time, and standard deviation of stance time would correlate with ML force control variability. We additionally hypothesized that average AP GRI, standard deviation of

AP GRI, average stance time, and standard deviation of stance time would correlate with AP force control variability in the deceleration task.

1.3 Innovation

This work will be the first to utilize nonlinear analyses to understand kinetics in both ACL injured and ACL reconstructed populations. Nonlinear analyses, particularly Lyapunov exponents, have been used in a number of studies with positive results. Research indicates greater kinematic variability after initial ACL rupture that continues after ACL reconstructive surgery, as measured by Lyapunov exponents. However, nonlinear analyses have yet to be used to understand the kinetic consequences of ACL injury and reconstructive surgery. To explore changes to the control of mGRFs our lab developed a force control task that uses real time visual feedback of force production. We used LyE analysis on the mGRF force profiles generated by subjects before and after ACL reconstructive surgery. Deficits in force control variability can help identify those who are at risk for further injury as they return to competitive sports and may be a modifiable factor for use in rehabilitation.

This work will be the first to utilize nonlinear analyses to understand kinetic variability in high performance athletes. Most ACL injuries occur during sports requiring running and cutting, like football, soccer, and handball. ACL injuries can be devastating as recovery requires lengthy rehabilitation forcing athletes to miss up to a full season of play. Understanding force control variability in high performing athletes may also help in reducing ACL injury incidence rates by establishing a standard or identifying a deficit. Utilizing real time visual feedback of force production we conducted LyE analysis to understand force control variability. We

established a spectrum of force control variability based on performance by comparing the LyE values calculated for healthy uninjured subjects, those with ACL injury, those with ACL reconstruction, and high performance athletes.

This work will be the first to explore the relationship between nonlinear analysis techniques and ACL injury mechanism. A majority of ACL tears occur through a noncontact mechanism, typically when athletes perform a cut maneuver or land from a jump. This injury mechanism points to neuromuscular control deficits as the inciting factor of injury. While there are a number of training programs aimed to improve neuromuscular control, both prospectively and after ACL reconstruction, injury rates remain the same. Nonlinear techniques have the ability to capture subtle changes to a signal which may be indicative of neuromuscular changes. Utilizing the force control task described earlier in a cohort of ACL-d and ACL-r subjects that experienced both contact and noncontact ACL injuries we identified differences based on injury mechanism.

This work will be the first to explore the relationship between nonlinear analysis techniques and dynamic sport related tasks. Because cutting maneuvers increase ACL injury risk understanding the biomechanics is vital. During these tasks athletes generate force in many directions, these mGRFs are crucial to completing these tasks. And so the ability to successfully control mGRFs may also be important in cutting performance. Current research correlates larger kinetic variables measured during cutting tasks to better cutting performance. In addition to the force control task the high performance athletes performed a series of running and cutting maneuvers. We used correlational analysis to understand the relationship between biomechanical measures from the running and cutting maneuvers and the LyE values calculated

during the force control task. Establishing these relationships will inform therapists and athletes as patients return to sports.

Chapter 2

EVALUATING MULTIDIRECTIONAL GROUND REACTION FORCE PRODUCTION AND CONTROL USING NONLINEAR ANALYSES

2.1 Introduction

Rupture of the anterior cruciate ligament (ACL) is a common injury that leads to significant financial burdens. Of those injured, about 25% will experience a re-tear in the first year (Mark V. Paterno et al. 2012). ACL injuries most commonly occur through a noncontact mechanism that is associated with cutting maneuvers (Boden, Griffin, and Garrett Jr. 2000; Krosshaug et al. 2007). Hence, sports that require many directional changes also have a higher incidence of ACL tears. Cutting and turning tasks require modulation of multidirectional ground reaction forces (mGRFs). For example to turn during gait an individual first decelerates generating a posteriorly directed GRF and then adjust the GRF medially or laterally to change direction (Glaister et al. 2008). In more dynamic tasks like cutting Havens and colleagues found that as cutting angle increases medial-lateral GRF (ML GRF) impulse also increases (Havens and Sigward 2014b). Additionally, ML GRF impulse correlates with better performance (Havens and Sigward 2014a). With multiple changes to the direction and magnitude of GRFs during not only gait but also running and cutting, the ability to control these forces are especially important when considering ACL ruptures.

Many studies that aim to understand ACL recovery utilize biomechanical measures (joint angle, joint moment, etc.) in conjunction with linear analysis tools, like variance. However, these analyses do not capture the subtle time varying changes that may occur from stride to stride. These small changes may be crucial to understand recovery as they can indicate neuromuscular health (Decker et al. 2011). Nonlinear

analyses, such as the Lyapnuov Exponent (LyE), are able to capture these subtle changes and describe the structure of variability or how variability changes as a function of time. Research reports the maximum or largest LyE which indicates the maximal divergence of a trajectory. LyE analysis has numerous biomechanical applications including understanding gait in patients with lower limb amputation (Wurdeman, Myers, and Stergiou 2013, 2013; Wurdeman et al. 2014) and diabetic neuropathy (Dingwell and Cusumano 2000). Research focused on ACL injury indicates more divergence (larger LyE) of knee flexion angles during gait after ACL injury and reconstruction with more kinematic divergence common to all reconstruction types (Moraiti et al. 2010b; Stergiou et al. 2004a). Changes to divergence of knee flexion angle during gait post ACL injury and reconstruction may indicate functional deficits exist but the full extent of this finding remains unseen as the kinetic consequences of this injury have yet to be evaluated using nonlinear analyses.

This study aims to utilize real time visual feedback of a subjects force production in conjunction with nonlinear analysis techniques in an effort to gain insight on control of mGRFs. This study will quantify force control variability within a single limb in multiple directions. We believe this study can aid in ACL injury recovery and rehabilitation. However, for this analysis we must first establish a standard with uninjured healthy participants to make meaningful comparisons. In our analysis of a healthy uninjured cohort we hypothesize there will be no difference in force control variability (LyE) when comparing the right and left limb. We additionally hypothesize healthy subjects will have superior mGRF control, noted by lower LyE values, in the AP direction when compared to the ML direction as the

musculature for the lower limb is organized for mostly anterior/posterior motion and strength.

2.2 Methods

2.2.1 Subjects

For this study 10 healthy individuals with no previous history of knee injuries were included (Table 2.1). All participants were active in >50 hrs/yr of level I and II sports which include running and cutting activities. This study was approved by institutional review board at the University of Delaware and all subjects provided informed consent.

Table 2.1: Demographic data (average ± standard deviation) for healthy uninjured recreational athletes.

	n	Age (Years)	BMI (kg/m^2)
Male	3	22 ± 0.8	22.9 ± 0.7
Female	7	22 ± 0.3	21.9 ± 2.3

2.2.2 Force Control Task

All subjects completed a force control task. For the force control task subjects stood on two separate force platforms (AMTI OR-6, Watertown, MA) with a foot on each force plate and data sampled at 1000 Hz. Subjects received real time visual feedback of their AP or ML force production via custom written Labview code. Visual feedback was presented on a screen in front of subjects and included one slider that responded to force production and two stationary sliders that served as goals for

the subjects (Figure 1 and 2). The two stationary sliders represented 50% of the subjects' maximum strength in that direction, based on a calibration trial (Figure 1).

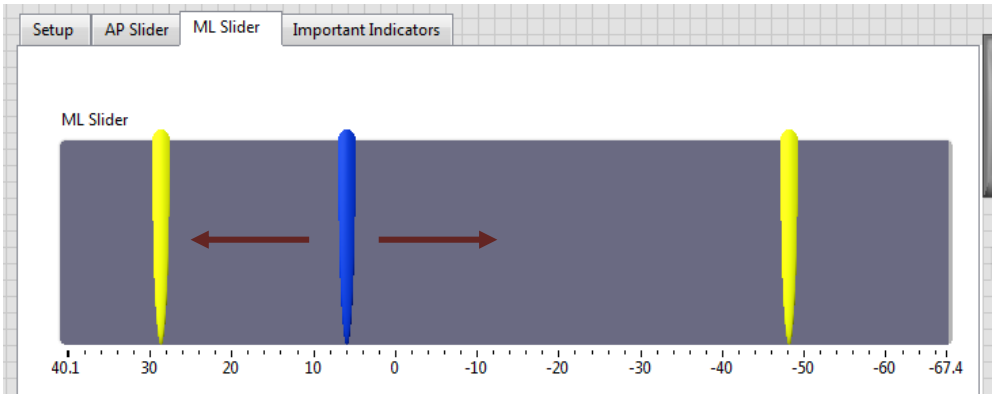


Figure 2.1: Example of visual feedback provided to study participants during the ML force control task. Arrows indicate the direction the mobile cursor moves.

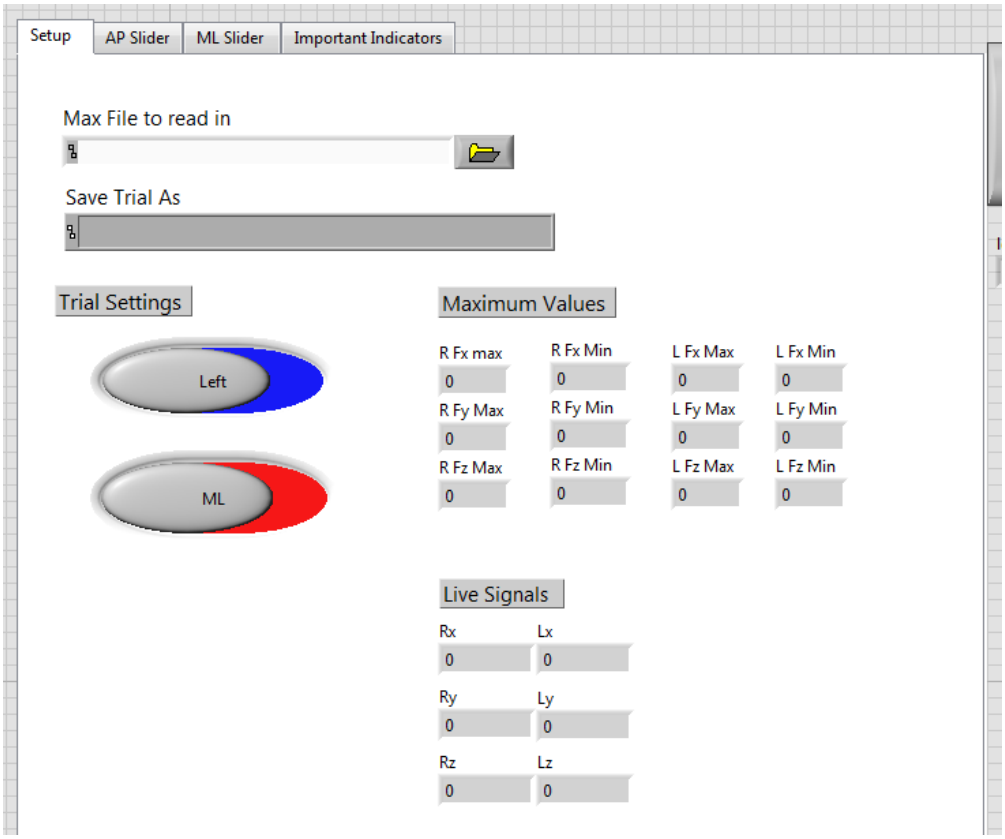


Figure 2.2: Example of visual feedback graphical interface. Test will be performed for each limb in both the anterior/posterior and medial/lateral directions.

To calibrate the force control task to each subjects' strength, subjects performed maximal force production trials. In these trials subjects pushed maximally on the force plates in the anterior, posterior, medial, and lateral directions with both feet. After calibration, subjects were instructed to control the moving cursor and align it with the two stationary sliders. They were instructed to generate force bi-directionally (anterior/posterior or medial/lateral) and continuously to the beat of a metronome set at 60 beats per minute, and to alternatively align the movable slider with each stationary slider. The force control task was two minutes in duration. The force control task was conducted for both the right and left limbs and both the AP and ML directions. During the force control task motion capture data was collected. For motion capture we used an 8 camera system (Qualisys Motion Capture Systems, Gothenburg, Sweden), retro-reflective markers were placed on bony landmarks to define the lower limbs. We also collected surface electromyography (Motion Lab Systems, Baton Rouge, LA) of eight lower limb muscles for each limb. Motion capture data was collected at 50 Hz while EMG was collected at 1000 Hz.

2.2.3 Data Analysis

The Lyapunov exponent was calculated for data collected during both the AP and ML force control tasks. For the force control task we calculated the Lyapunov exponent using the AP and ML force profiles generated during their respective tasks to determine AP force control variability and ML force control variability. In all cases unfiltered data was used as filtering may remove subtle changes within the signal. LyE is defined as the rate of divergence of infinitesimally close trajectories, and is determined through a multi-step process. Calculating LyE requires two input

parameters time lag (τ) and embedding dimension (m). Both of these parameters convert our signals of interest into state space. To calculate τ we used the Average Mutual Information (AMI) function. This function determines the percentage of information shared between two signals, for LyE purposes the two signals of interest were the same and were either AP force profile or ML force profile. *Tau* (τ) is defined as the first local minima of the AMI function. To determine m the Global False Nearest Neighbor function (GFNN) was used. Using time delayed copies (as specified by τ) of a signal GFNN measures the distance between trajectories in one dimension then increases the number of dimensions and measures again. A false nearest neighbor is defined by any significant changes in distance between trajectories when increasing the number of dimensions. As the dimensions are increased the number of false neighbors is determined. The first local minima of the GFNN function was then used as m. GFNN is important to 'unfold' the data as any folding would lead to misinterpreted LyE. Once τ and m are determined the data is transformed in state space; in state space LyE can then be calculated. LyE was calculated using the Wolf algorithm (Wolf et al. 1985). This algorithm calculates the Euclidean distance between trajectories and tracks the trajectories forward in time to determine the rate of divergence or convergence of the trajectories. The rate of divergence is determined for multiple trajectories and the largest, which is our value of interest, was reported.

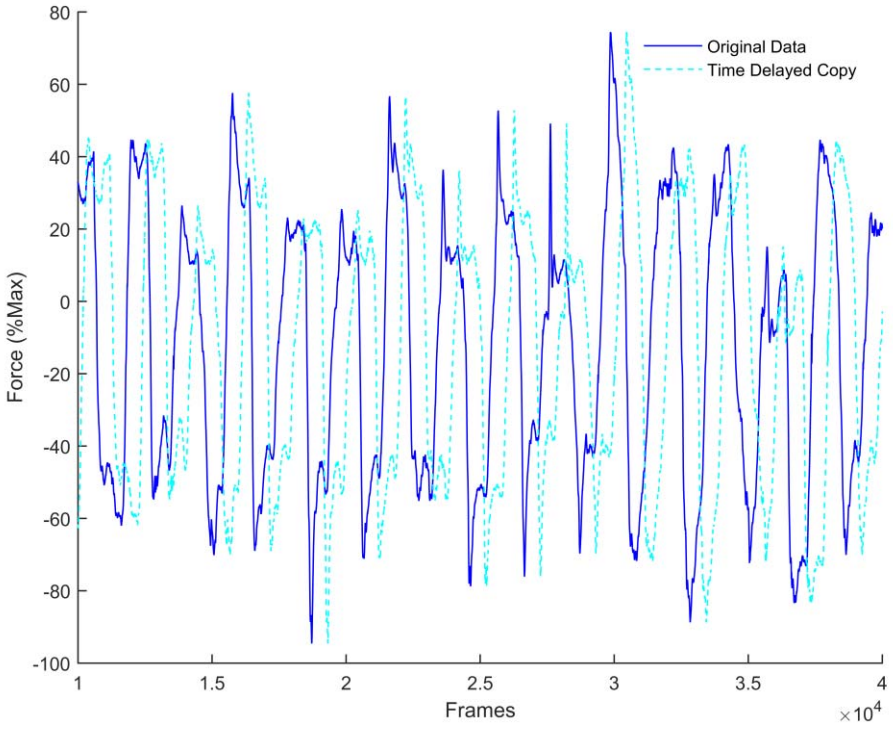


Figure 2.3: Original force profile (solid line) with a time delayed copy (dashed line).

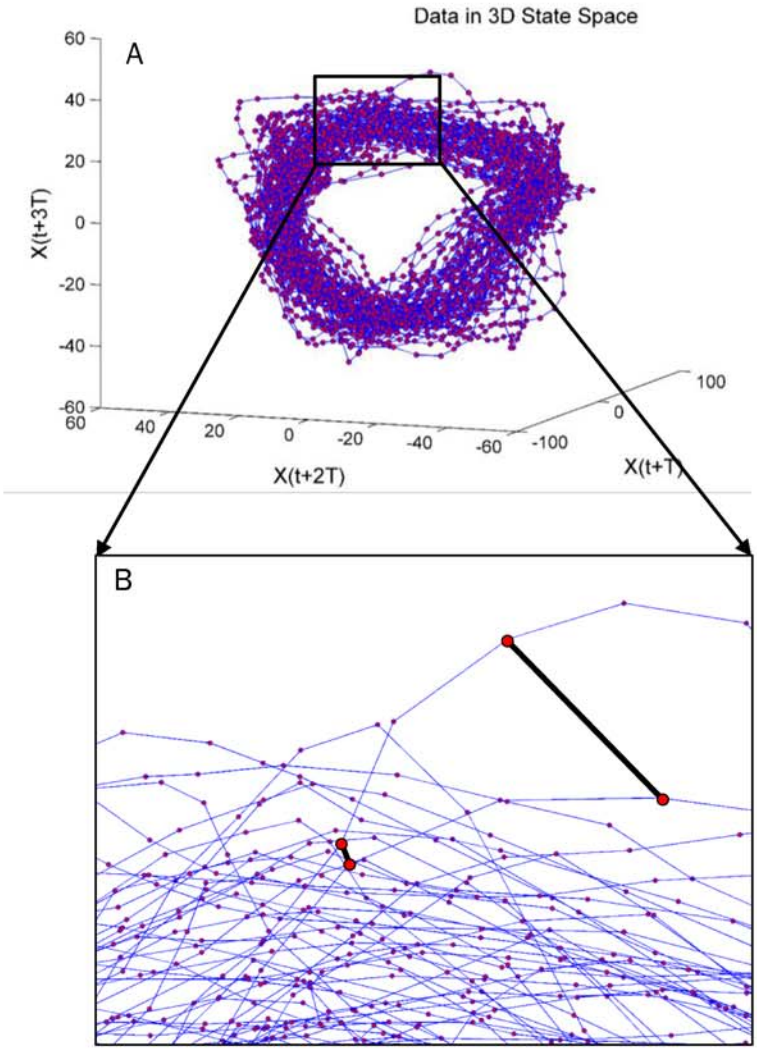


Figure 2.4: Force profile data transformed into 3D state space (A). Zoomed in view (B) with the distance between trajectories highlighted in black.

2.2.4 Statistical Analysis

For all statistical tests LyE calculated for each trial of both the AP and ML was averaged for each subject providing two measures AP LyE and ML LyE for each subject. A 2x2 repeated measures analysis of variance, ANOVA, comparing limbs (right/left) x direction (AP/ML) was used to determine significant differences in force control variability. For both tests significance was defined by $p<0.05$ and conducted using SPSS software (IBM Armonk, NY).

2.3 Results

Our data indicates there is no statistically significant difference in LyE values of healthy uninjured subjects when comparing the right and left limb. In the right limb of healthy uninjured participants we calculated LyE values to be 3.39 ± 0.73 bits/sec and 2.96 ± 0.47 bits/sec in the AP and ML directions respectively (Figure 2.5). While in the left limb we calculated LyE to be 3.24 ± 0.52 bits/sec and 3.19 ± 0.57 bits/sec in the AP and ML directions (Figure 2.5).

Additionally, our data indicates there is no statistically significant difference in LyE values generated in the ML versus AP direction. P-values from the ANOVA approach significance ($p = 0.054$) but effect size calculated using Cohen's d was moderate. On average we calculated LyE in the AP direction to be 3.32 ± 0.63 bits/sec. In the ML direction LyE was calculated to be 3.10 ± 0.53 bits/sec.

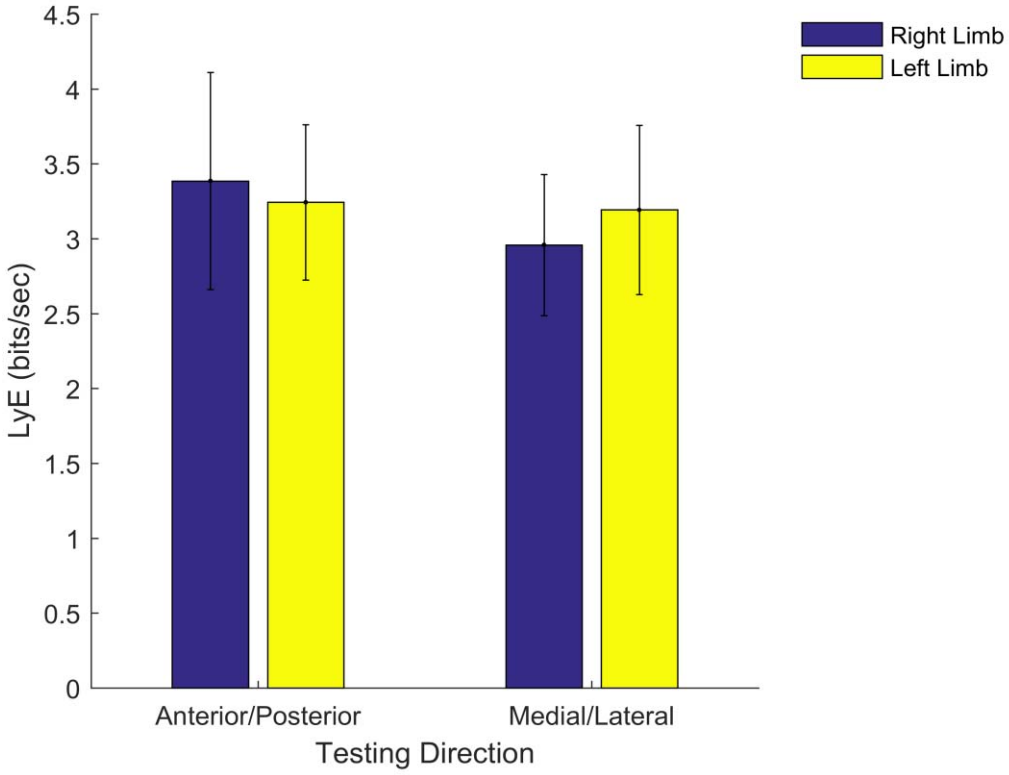


Figure 2.5: Average maximal Lyapunov exponent (bits/second) ± standard deviation during the force control task. Right limb indicated in blue while the left limb is indicated in yellow. Data is reported for both the medial/lateral and anterior/posterior directions.

2.4 Discussion

The aim of this study was to evaluate mGRF control by utilizing a nonlinear analysis called the Lyapunov exponent. Through this methodology we hoped to establish a standard of force control variability with an uninjured and active population. We found no statistically significant difference in force control variability between limbs when performing the force control task. These results support our first hypothesis and was to be expected as our subjects endured no major injuries to their lower limbs. While there was no statistically significant differences in force control variability (LyE) between directions, contrary to our second hypothesis, the p-value from our analysis approach significance ($p = 0.054$). Subjects exhibited reduced LyE values when performing the force control task in the ML direction which may indicate better control in that direction as a lower LyE value indicates less divergence. Overall our results support our first hypothesis that there is no difference in force control variability between limbs, but not our second hypothesis, that subjects would exhibit superior force control variability in the AP direction compared to the ML direction.

Healthy uninjured subjects active in running and cutting sports may have improved control in the ML direction, noted by reduced divergence, due to influence frontal plane loads have on the knee health. ACL injury occurs during cutting and jumping tasks when the knee undergoes a combination of valgus collapse and rotation (Boden, Griffin, and Garrett Jr. 2000; Olsen 2004). With valgus collapse being a major contributor to ACL tears control of ML ground reaction forces is crucial. When looking at cutting maneuvers specifically we find that frontal plane and rotational moments at the knee increase dramatically during cutting tasks as compared to straight running (Besier and Lloyd 2001) and cadaver models indicate valgus and rotational loading significant increases ACL strain (Shin, Chaudhari, and Andriacchi 2011).

Research that focuses on jumping tasks indicates that frontal plane knee kinetics and kinematics were able to prospectively predict ACL injuries in high school female athletes highlighting the importance of adequate neuromuscular control in the ML direction (Hewett et al. 2005). In addition to ACL injuries, frontal plane control has been indicated as an important factor in patellofemoral pain (Ireland et al. 2003; Powers 2003). Significant hip weakness in the frontal and transverse planes has been correlated to the presence of patellofemoral pain (Ireland et al. 2003). Treatment options both proximal and distal to the knee have been identified and encouraged for patellofemoral pain highlighting the need to understand motor control outcomes at the foot which can include mGRF force control variability. In addition to prevent sport related injuries reduced divergence in ML force control may be a protective measure for degenerative joint disease. Frontal plane knee moments have been linked to the progression and severity of knee osteoarthritis (Baliunas et al. 2002; Mündermann et al. 2004). Reduced divergence in ML force control variability may lead to reduced variability in frontal plane knee moments creating a better environment for the knee.

While evident at the joint level, the importance of reduced divergence in ML force control is also noted when looking at muscle activity. In a study utilizing a force matching and control task similar to the one used in this study, Flaxman and colleagues captured EMG of lower limb muscles and classified them into general joint stabilizers, specific joint stabilizers, and moment actuators (Flaxman, Speirs, and Benoit 2012). Based on the muscle's functional role these results may apply to the differences we see AP and ML force control variability. Flaxman classified vastus lateralis (VL), vastus medialis (VM), and biceps femoris (BF) as joint stabilizers while classifying only the semitendinosus (ST) and rectus femoris (RF) as moment

actuators. Relying on the VL and VM as joint stabilizers may lead to reduced significant ML perturbations, again creating a safer environment for the knee joint and leading to less divergence of ML force control variability. Additionally, research on unilateral weight bearing found that the gluteus medius (GM) and VM muscles were most active during the task as it required frontal plane control (Bolgla et al. 2016). As was noted through outcomes seen in both kinematics and kinetics, specified muscle control may be better equipped to control ML force control variability rather than AP force control variability.

There were some limitations in the implementation of this study. We focused on including participants that were active in running and cutting sports which unfortunately excluded potential participants with no history of knee injuries however this provided a cohort more closely linked to potential ACL injury. During the force control task no fail criteria was established to ensure subjects completed the task to a certain level of accuracy. However, during data collection the variables of interest were being monitored and in subsequent data analysis all trials were inspected for both magnitude of each subjects force production and appropriate number of cycles. By monitoring number of cycles we were able to ensure subjects' maintained appropriate timing. None of these limitations would grossly affect our findings.

This study and accompanying data set establishes a standard for force control variability as subjects generate mGRFs. Other studies measuring changes to Lyapunov exponent find a range of values making it difficult to make comparisons between research groups based on magnitude and researchers must utilize a healthy population first to identify trends. While a number of studies have explored the changes to the structure of variability in kinematic measures no work has been done utilizing kinetic

measures. Overall, we found that in a healthy, uninjured, and active cohort there was no difference in LyE values measured from the right and left limb but there may be a difference in LyE measured from the AP and ML direction. Biomechanical data exploring both joint level kinematics and kinetics and muscular control indicate the importance on frontal plane control as it relates to knee injury; smaller LyE values measured in the ML direction may be a beneficial adaptation as it may protect the knee from injury. Lastly, these results have established a standard for healthy and active young adults that will help understand changes to force control variability that may occur as a result of anterior cruciate ligament injury and reconstruction or participation in high performance athletics which can inform both rehabilitation practices and sports training.

Chapter 3

EXPLORING CHANGES TO FORCE CONTROL VARIABILITY AS IT RELATES TO ACL INJURY, ACL RECONSTRUCTION, AND HIGH PERFOMANCE ATHLETICS

3.1 Introduction

Anterior cruciate ligament (ACL) tears are a common sports related injury that affect both recreational athletes and elite athletes. Approximately 3 in 1000 people will experience an ACL tear each year which adds up to a total of 250,000 annually (Bates et al. 2015, Miyasaka et al. 1991). Primarily, those who experience ACL injury and intend to return to sport like activities undergo ACL reconstruction. Conservative estimates of surgery costs indicate a 1 billion annual price tag for treating this injury. Unfortunately, 25% of patients will experience a secondary ACL tear in the 1st year after returning to sports which then funnels them back into surgery and rehabilitation.

After ACL injury and reconstruction there are a number of biomechanical impairments. Of particular interest are neuromuscular control deficits. Post ACL injury there are significant losses to proprioception in both the sagittal and transverse planes; the inability to sense joint position may prevent protective neuromuscular control strategies (Muaidi et al. 2009). Electromyography has highlighted increased hamstring co-contraction and altered hamstrings timing during gait in those classified as non-copers (Rudolph et al. 2001). While a number of clinicians have developed neuromuscular rehabilitation protocols that focus on improving biomechanics during jump landings (Hewett, Ford, and Myer 2006) or developing personalized motor control strategies (Chmielewski et al. 2005; Hartigan, Axe, and Snyder-Mackler 2009; Hurd, Chmielewski, and Snyder-Mackler 2006; Di Stasi and Snyder-Mackler 2011)

there is still a high prevalence of re-injury as patients return to sport like activities. While neuromuscular control may be vital to proper biomechanics during cutting tasks there is still information to be gained that can improve current rehabilitation methods.

ACL injuries occur during running/cutting tasks and to perform a cut athletes initially decelerate prior to changing direction they must generate significant anterior/posterior (AP) ground reaction forces (Besier and Lloyd 2001). Once deceleration is complete they must then redirect the center of mass hence producing greater medial/lateral (ML) ground reaction forces (Besier and Lloyd 2001). As speed and power increases, which increases with skill level, modulation of these multidirectional ground reaction forces (mGRFs) which are seen during cutting tasks are crucial in providing proper body support and positioning to protect against injury. However, performance of these tasks after ACL injury are dangerous and so safe methods that can elicit dynamic loads similar to cutting are crucial.

Sports that require running and cutting tasks see the highest number of ACL tears. This includes sports like basketball, handball, football, and soccer. Internationally, soccer is a very popular sport and when considering that elite soccer athletes experience approximately 700 changes of direction during a single game proper execution of these task is important not only to success as a player but reducing injuries. In addressing this need there are numerous training protocols focused on speed, power, and agility. These training protocols include activities like resisted sprinting and plyometrics which themselves include different activities that focus on straight running and changes of direction. While these programs show improvement in jumping height and sprinting times there are no clear improvements to change of direction performance (de Hoyo et al. 2016). These results indicate strength, power,

and agility may not be major contributing factors to cutting performance. And so understanding cutting mechanics in a high performance cohort utilizing innovative analysis techniques may identify optimal control strategies and potentially better inform ACL injured and reconstructed patients.

Nonlinear analysis techniques may be a pathway to gain additional insight that previous studies have missed. Recently, there has been an increase in the use of nonlinear tools including the Lyapunov Exponent (LyE) and Approximate Entropy (ApEn). These analyses have been shown to capture small time varying changes previously unseen in linear analyses and may prove valuable in understanding injury, recovery, and performance (Decker et al. 2011; Stergiou and Decker 2011). LyE and ApEn are currently being used in a number of different biomechanical applications including postural control (Dingwell and Cusumano 2000), amputee gait (Wurdeman, Myers, and Stergiou 2013; Wurdeman et al. 2014), ACL injury (Stergiou et al. 2004a) and ACL reconstruction (Moraiti et al. 2010b). For this study we focued on the Lyapunov exponent which measures the divergence of infinitesimally close trajectories, by measuring the change in distance between trajectories over time. Utilizing LyE analysis in combination with gait analysis, Stergiou and colleagues found larger kinematic variability (larger LyE) at the knee during gait after initial injury and after reconstruction (Stergiou et al. 2004a). After reconstructive surgery larger kinematic variability persists and is common to all reconstruction types (Moraiti et al. 2010b). While successful in exploring kinematic variability, Lyapunov exponent analyses have yet to be applied to kinetics. Understanding changes to subjects ability to control mGRFs, as measured by LyE, after ACL injury, ACL

reconstruction, and in high performance athletes may provide additional and valuable insight.

Therefore the purpose of this study was to safely quantify changes to force control variability (LyE) that are relatable to mGRF productionafter ACL injury, ACL reconstruction, and participation in high performance athletics. This study quantified both AP force control variability and ML force control variability for each limb with a force control task that utilizes real time visual feedback of an individual's force production. With a large study cohort we have multiple hypothesis, both within and across groups. First, we hypothesized that ACL injured (ACL-d) and ACL reconstructed (ACL-r) would have diminished mGRF control, noted by larger LyE values, in the injured limb when compared to the uninjured limb. Second, we hypothesized the high performance athletes would have similar mGRF control across both limbs and directions (equivalent LyE for both limbs in both directions). Across groups we hypothesized that high performance athletes would exhibit superior mGRF control noted by reduced LyE values when compared to healthy active recreational athletes (uninjured controls). Lastly, we hypothesized that the involved limb of both ACL-d and ACL-r subjects would show diminished mGRF control (larger LyE) when compared to both uninjured recreational athletes and high performance athletes based on neuromuscular control deficits seen post-injury and post-surgery.

3.2 Methods

3.2.1 Subjects

In this study there was a total of 48 participants. Ten subjects between the ages of 14 and 55 years of age who were active in >50 hrs/yr of level I and II sports and no

history of ACL injury or other major lower limb injuries were recruited from the local community. Twenty subjects between the ages of 14 and 55 years of age also active in >50 hrs/yr of level I and II sports were recruited from the University of Delaware physical therapy clinic . Ten subjects who have experienced an ACL rupture and eleven subjects who have received an ACL reconstruction were included in this study. ACL deficient (ACL-d) subjects included those who have experienced an isolated unilateral ACL rupture. ACL reconstructed (ACL-r) subjects included those who have undergone ACL reconstructive surgery in the last 3-9 months. Additionally, we recruited seventeen (8M/ 9W) athletes from the University of Delaware men's and women's soccer team to serve as high performance athletes. This study was approved by internal review board at the University of Delaware and all subjects were provided informed consent.

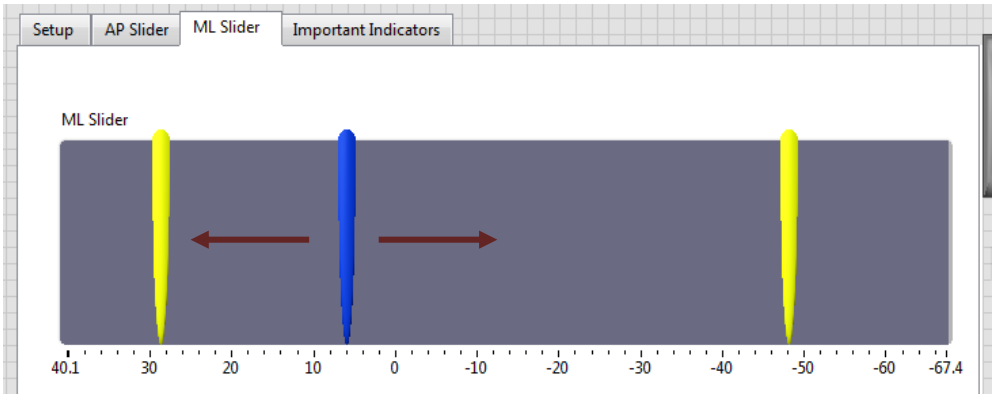
Table 3.1: Subject demographic data (average ± standard deviation) for healthy recreational athletes, ACL-d, ACL-r and collegiate soccer players.

	Age (Years)	BMI (kg/m^2)	Time since injury (Months)	Time since surgery (Months)
Healthy (3M/7F)	22 ± 0.5	22.2 ± 2.0	--	--
ACL-d (7M/3F)	24 ± 8.2	24.5 ± 2.5	3.2 ± 1.8	--
ACL-r (3M/8F)	21 ± 7.8	23.5 ± 4.1	9.8 ± 2.6	8.0 ± 1.8
Soccer (8M/9F)	21 ± 1.4	23.8 ± 2.1	--	--

3.2.2 Force Control Task

All subjects completed a force control task. For the force control task subjects stood on two separate force platforms (AMTI OR-6, Watertown, MA) with a foot on each force plate and data sampled at 1000 Hz. Subjects received real time visual

feedback of their AP or ML force production via custom written Labview code. Visual feedback was presented on a screen in front of subjects and included one slider that responded to force production and two stationary sliders that served as goals for the subjects (Figure 3.1 and 3.2). The two stationary sliders represented 50% of the subjects' maximum strength in that direction, based on a calibration trial (Figure 3.1).



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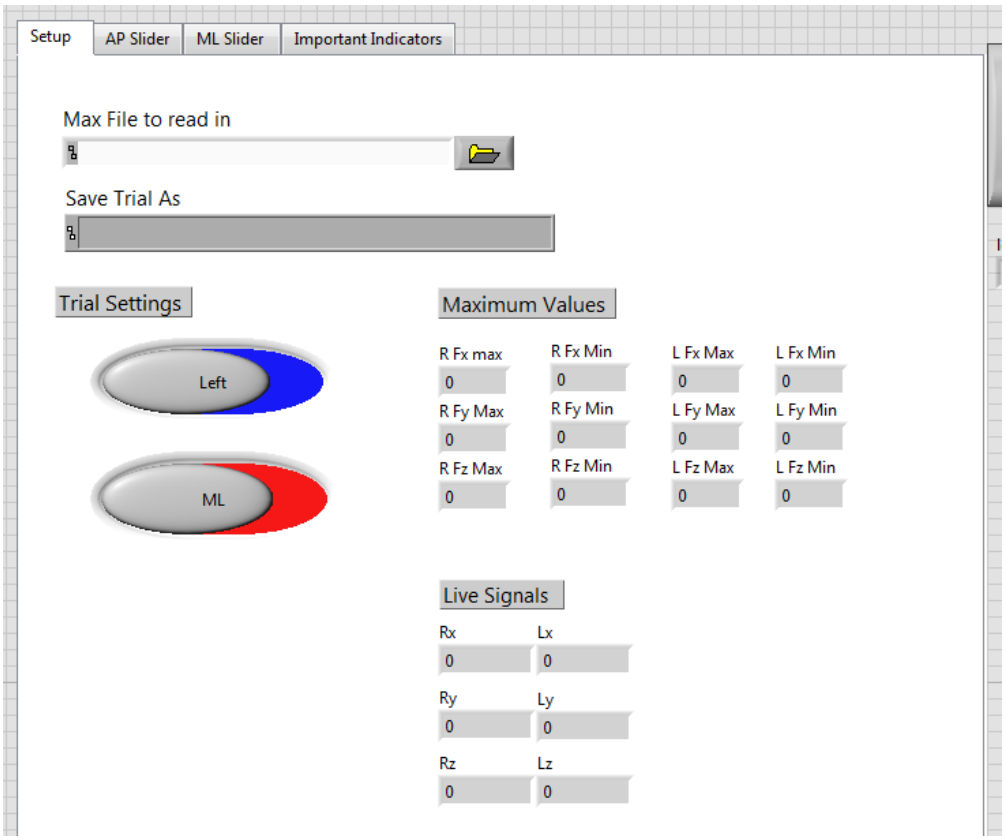

Figure 3.2: Example of visual feedback graphical interface. Test will be performed for each limb in both the anterior/posterior and medial/lateral directions.

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3.2.3 Data Processing

The Lyapunov exponent was calculated for data collected during both the AP and ML force control tasks. For the force control task we calculated the Lyapunov exponent using the AP and ML force profiles generated during their respective tasks to determine AP force control variability and ML force control variability. In all cases unfiltered data was used as filtering may remove subtle changes within the signal. LyE is defined as the rate of divergence of infinitesimally close trajectories, and is determined through a multi-step process. Calculating LyE requires two input parameters time lag (τ) and

embedding dimension (*m*). Both of these parameters convert our signals of interest into state space. To calculate τ we used the Average Mutual Information (AMI) function. This function determines the percentage of information shared between two signals, for LyE purposes the two signals of interest were the same and were either AP force profile or ML force profile. *Tau* (τ) is defined as the first local minima of the AMI function. To determine *m* the Global False Nearest Neighbor function (GFNN) was used. Using time delayed copies (as specified by τ) of a signal GFNN measures the distance between trajectories in one dimension then increases the number of dimensions and measures again. A false nearest neighbor is defined by any significant changes in distance when increasing the number of dimensions. As the dimensions are increased the number of false neighbors is determined. The first local minima of the GFNN function was then used as *m*. GFNN is important to 'unfold' the data as any folding would lead to misinterpreted LyE. Once τ and *m* are determined the data is transformed in state space; in state space LyE can then be calculated. LyE was calculated using the Wolf algorithm. This algorithm calculates the Euclidean distance between trajectories and tracks the trajectories forward in time to determine the rate of divergence or convergence of the trajectories. The rate of divergence is determined for multiple trajectories and the largest, which is our value of interest, was reported.

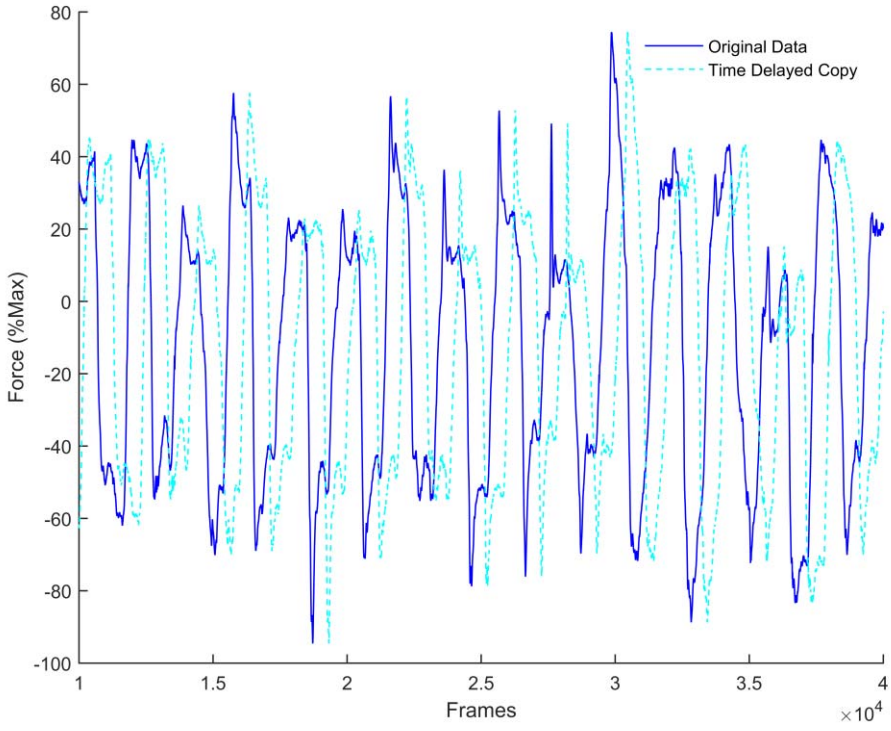

Figure 3.3: Original force profile (solid line) with a time delayed copy (dashed line).

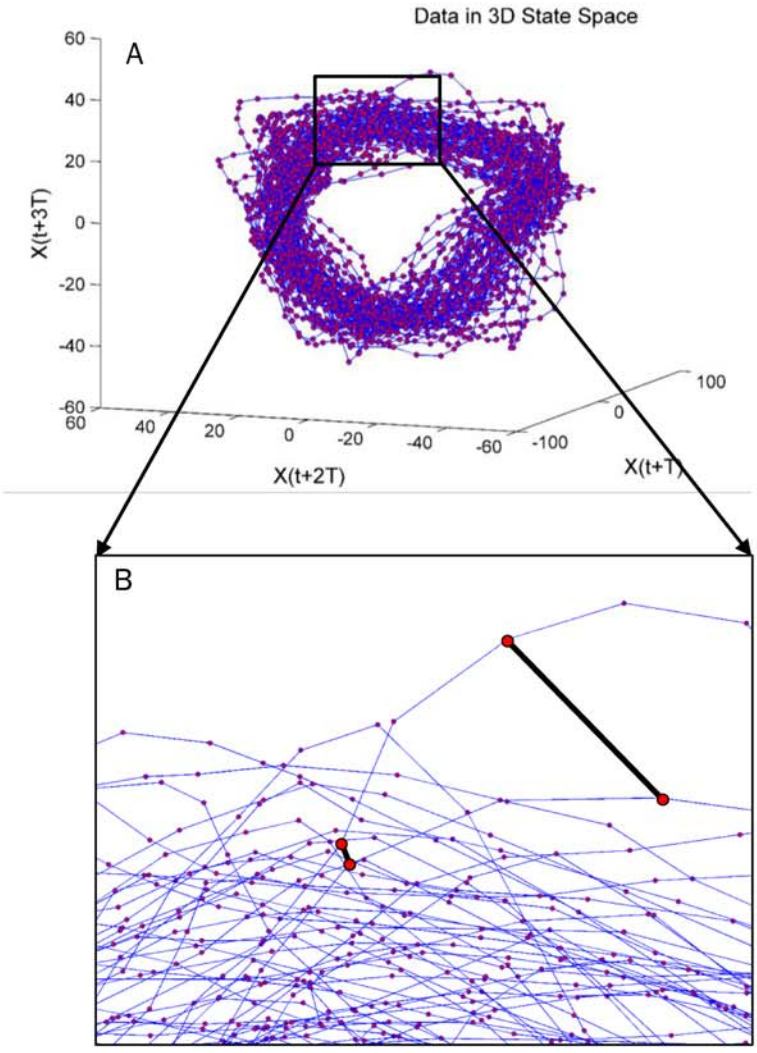

Figure 3.4: Force profile data transformed into 3D state space (A). Zoomed in view (B) with the distance between trajectories highlighted in black.

3.2.4 Statistical Analysis

For this study the statistical analysis was a multistep process. First, limb differences and direction differences during the force control task was determined for the high performance athletes using a 2x2 (limb x direction) repeated measures ANOVA. We also determined limb differences for the ACL-d and ACL-r subjects using a student's paired t-test (involved limb vs. uninvolved limb). To determine significant differences in force control variability (LyE) between groups and direction a 4x2x2 (group x direction x limb) mixed design ANOVA was used. Because there was no difference between limb in healthy controls and high performance athletes, we randomized limb assignment and compared to the involved limb of ACL-d and ACL-r subjects for the mixed design ANOVA. Significanct Based on the results of our initial testing post-hoc testing was used to determine specific differences in force control variability.

3.3 Results

3.3.1 Within ACL-d and ACL-r subjects

Paired t-tests revealed that for ACL-d subjects there is no statistically significant difference between the involved and uninvolved limbs in both the AP and ML directions (Figure 3.5, $p > 0.05$). For these subjects we calculated average LyE values for the AP direction to be 7.33 ± 1.94 and 7.68 ± 1.97 bits/sec of the involved and uninvolved limb, respectively. LyE values calculated for the ACL-d group in the ML direction were 6.90 ± 1.90 and 7.03 ± 1.81 bits/sec in the involved and uninvolved limb, respectively. In ACL-r subjects our analysis revealed a statistically significant difference between the involved and uninvolved limb for LyE calculated from the AP direction (Figure 3.5, $p = 0.016$) but not the ML direction (Figure 3.5, $p > 0.05$).

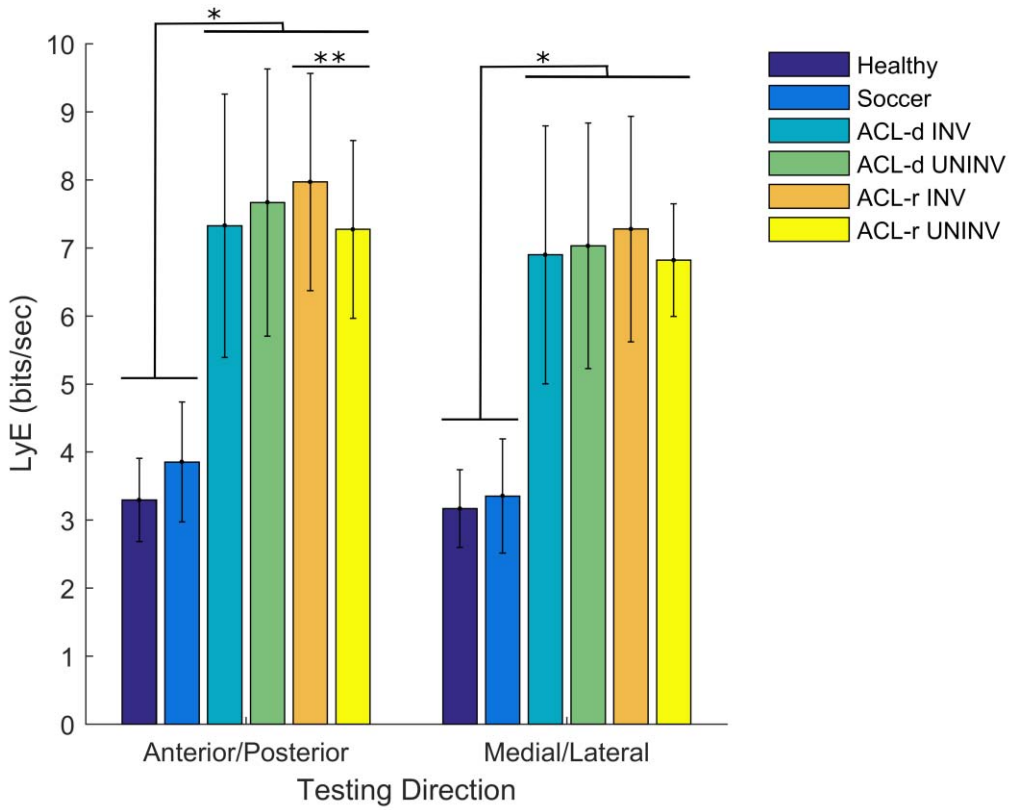

Figure 3.5: Average maximal Lyapunov exponent (bits/second) ± standard deviation during the force control task for healthy control subjects (blue), high performance athletes (red), ACL deficient (ACL-d) patients (green), and ACL reconstructed (ACL-r) patients (orange). For ACL-d and ACL-r subjects the involved limb is solid while the uninvolved limb is striped. Data is reported for both the medial/lateral and anterior/posterior directions.* $p<0.05$ ANOVA, **$p<0.05$ paired t-test

3.3.2 Within high performance athletes

For the high performance group, we found that there is no difference in LyE between the right versus left limb (Figure 3.6, $p > 0.05$) but there is a statistically significant difference in LyE between the AP and ML directions (Figure 3.6, $p = 0.050$). In the AP direction collegiate soccer players exhibited LyE values calculated at 3.98 ± 0.99 and 3.82 ± 1.04 bits/sec for the right and left limb respectively. While in the ML directions we calculated LyE values for high performance athletes averaged at 3.31 ± 0.76 and 3.49 ± 0.98 bits/sec for the right and left limb respectively.

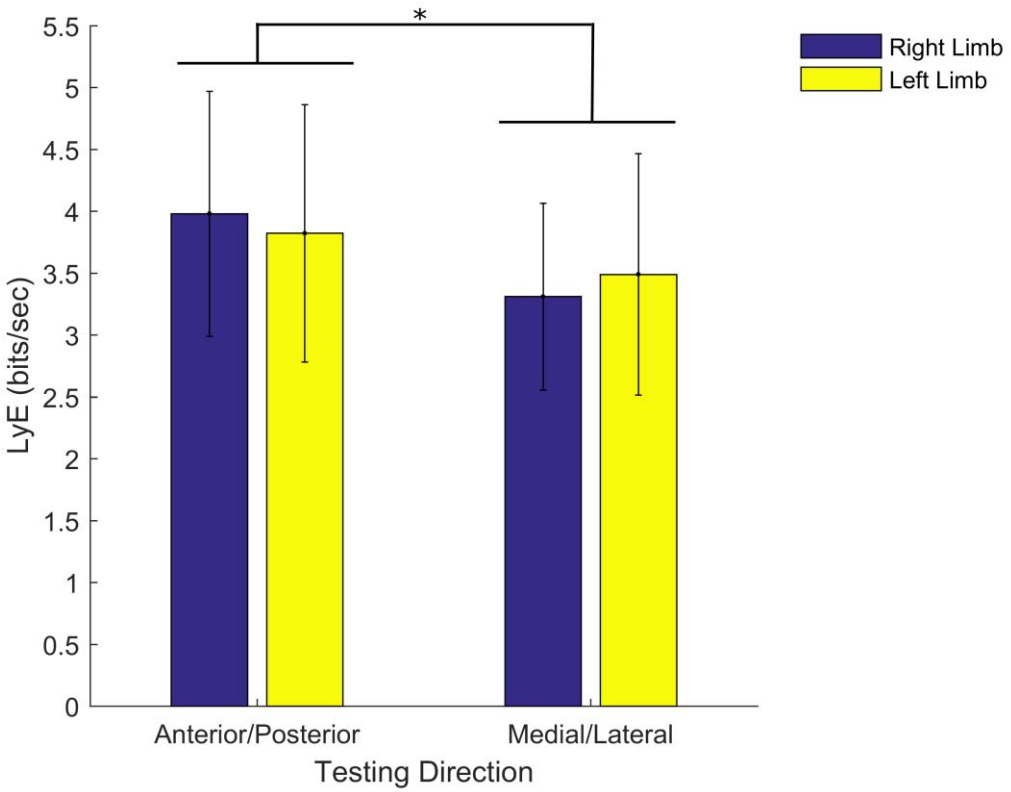

Figure 3.6: Average maximal Lyapunov exponent (bits/second) ± standard deviation during the force control task for high performance athletes. Right limb in blue; left limb in red. Data is reported for both the medial/lateral and anterior/posterior directions. *$p \leq 0.05$

3.3.3 Across all subjects groups

When comparing the involved limb of those who have experienced an ACL tear or undergone ACL reconstruction (ACL-d and ACL-r groups) to both healthy uninjured controls and high performance athletes LyE was significantly larger (Figure 3.5, $p < 0.001$) in both the AP and ML directions. There is no significant difference in LyE values when comparing the involved limb of ACL-d and ACL-r subjects (Figure 3.5, $p > 0.05$). We found no statistically significant difference in LyE measured between healthy uninjured individuals and collegiate soccer players (Figure 3.5, $p > 0.05$).

3.4 Discussion

The purpose of this study was to quantify changes to force control variability in healthy uninjured young adults, high performance athletes, those who have experienced an ACL tear, and those who have undergone ACL reconstruction. Through the utilization of an innovative force control task we were able to determine the Lyapunov exponent as subjects' generated mGRFs in both the AP and ML direction. Our study revealed statistically significant differences in LyE values based on task direction and significant changes to LyE values that occur after ACL injury and ACL reconstruction.

In the ACL-d group we found no difference in force control variability when comparing the involved and uninvolved limbs in both the AP and ML direction. For the ACL-r group the involved limb exhibits significantly diminished mGRF control, or larger LyE values, when compared to the uninvolved limb in the AP direction, but not the ML direction. These results partially support our first hypothesis. While there was a difference in LyE between the involved and uninvolved limb it is only present in

ACL-r subjects and only in the AP direction. High performance athletes exhibited no difference in force control variability between limbs but they exhibited lower LyE values in the ML direction when compared the AP direction. These results also partially support our second hypothesis as there was no limb effect but there was a direction effect. When looking at differences across group we found no statistically significant differences between healthy uninjured subjects and collegiate soccer players contrary to our third hypothesis. Lastly, there is significantly larger divergence of force control variability in both the AP and ML directions in the ACL-d and ACL-r groups when compared to our healthy and high performance cohorts which supports our final hypothesis. In summary, after ACL injury there is significantly more force control variability indicating more divergence and poor mGRF control which is not resolved through ACL reconstructive surgery; ACL-r subjects exhibit differences in control between involved and uninvolved limbs while high performance athletes exhibit differences in force control based on direction.

High performance athletes exhibit similar force control variability limb to limb but when comparing directions we found less divergence or lower LyE values in the ML direction when compared to the AP direction. All participants in the high performance group were members of the University of Delaware men's or women's soccer team; soccer is a unique sport in that it requires multiple changes of speed and direction while players see very little time in contact with the ball. Studies estimate that elite soccer players may experience over 700 changes of direction in a single match (Bloomfield et al. 2007). This play style requires athletes to be agile, and to accelerate quickly and with great power. In an effort to meet these requirements soccer training includes many drills that include sprinting, plyometrics, and skill drills.

Research on these training regimes indicate resistance sprinting, limb strengthening, and plyometrics can improve jumping and sprinting performance but not change of direction performance (de Hoyo et al. 2016). In addition to protocols focused on improved player skill there are also specific training protocols designed to prevent ACL injury like the FIFA 11+ protocol. This training regime includes two sessions of running exercises that bookend plyometric and balance exercises in an effort to improve not only running performance but proprioception, balance, and core strength (Bizzini and Dvorak 2015; Silvers-Granelli et al. 2015). FIFA 11+ has shown to reduce the number of injuries and time lost to injury in a number of different populations but also lead to improved motor control as measured by muscle activation, dynamic stability, and time to stabilization (Bizzini and Dvorak 2015). Specialized training, which focuses particularly on change of direction tasks, including injury prevention and agility protocols may influence the directional differences we see in force control variability rather than the other measures of change of direction performance measured in other studies.

While not statistically significant our data indicates LyE calculated from high performance athletes was greater than that of healthy uninjured recreational athletes this may identify optimal control. In sports athletes must be consistent yet flexible in their movements to avoid injury in such a fast paced sport, hence the potential need for larger LyE values, or more divergence, when compared to uninjured controls. The theory of optimal movement variability posits that there exists a preferred bandwidth of variability for healthy and mature motor skills (Stergiou, Harbourne, and Cavanaugh 2006). Movements should not be too rigid to limit adaptability but not too unstable to limit predictability. This theory has been supported through case studies

that explore postural control in children with cerebral palsy and athletes who have experienced a concussion (Stergiou, Harbourne, and Cavanaugh 2006). Because high performance athletes participate in intense training and high skill play they are ideal candidates to maintain optimal movement variability. While we only see a marginally larger LyE values this may be enough to reach preferred levels of divergence.

We found no difference between involved and uninvolved limbs in a majority of comparisons within ACL-d and ACL-d subjects. Deficits to mGRF control, as noted through larger LyE values were consistent in both the involved and uninvolved limb. This global pattern of poor mGRF control indicates an upper level neuromuscular error may be at play. Animal models of ACL injury indicate deficits to fine motor control and altered regulation of motor reflexes which highlights both motor and spinal dysfunction resulting from ACL injury (Laurin et al. 2015). Additionally, in vivo studies found that post ACL injury there is a reorganization of the central nervous system measured by functional MRI (Kapreli et al. 2009), and diminished corticospinal excitability measured by trans-cranial magnetic stimulation (A. S. Lepley et al. 2015; Adam S. Lepley et al. 2014). Disruption to the sensory inputs and alterations to the central nervous system caused by ACL injury may influence force control variability in both the involved and uninvolved limb of patients.

Our results identified significantly larger LyE values in both the involved and uninvolved limb of the ACL-d and ACL-r subjects when compared to our uninjured cohort (healthy uninjured subjects, and high performance athletes). One quarter of those who experience an ACL injury will experience a 2nd injury in the first year they return to sports. Contralateral re-injury is equally if not more likely than an ipsilateral

re-injury (Wiggins et al. 2016). Biomechanical changes occurring bilaterally, to both the involved and uninvolved, may be contributing to this phenomena. Current research highlights bilateral changes to both dynamic and static balance after ACL injury (Clagg et al. 2015; Culvenor et al. 2016). Importantly, postural stability is predictive of second ACL injury (M. V. Paterno et al. 2010). Significantly larger force control variability between the injured (ACL-d and ACL-r) and uninjured (Healthy Controls and High Performance Athletes) cohorts that is present in both the involved and uninvolved limbs may identify this measure as a contributing factor to re-injury seen in both ipsilateral and contralateral limbs.

We found a statistically significant difference between involved and uninvolved limbs when generating force in the AP direction for ACL-r subjects. In the AP direction ACL-r subjects had larger LyE values in the involved limb when compared to the uninvolved limb. Reduced mGRF control in the reconstructed limb is not surprising as surgery is very chaotic for the knee joint. Because the ACL is unable to heal itself it must be completely replaced in the event of a tear. Tibial tunneling during reconstructive surgery significantly reduces meniscal attachment area and ultimate strength which risks further injury (LaPrade et al. 2015; Watson et al. 2014). Additionally, removal of the ligament remnant may eliminate important mechanoreceptors (Georgoulis et al. 2001). After surgery research indicates reduced proprioception (Muaidi et al. 2009), reduced coordination variability (Pollard et al. 2014), altered muscle activation (Macleod 2012), and altered joint kinematics (Georgoulis and Papadonikolakis 2003; Hofbauer et al. 2014). The combination of these deficits may translate to poor control of mGRFs at the foot, noted by larger LyE values for the involved limb.

This is the first study to evaluate dynamic ML force production3 in ACL-d and ACL-r patients prior to returning to sports. Generating this type of loading has been previously unattainable as cutting maneuvers can risk further injury. Our results demonstrated a large significant difference in force control variability, as measured by LyE, between our injured and uninjured cohorts. LyE values in the ACL-d and ACL-r subjects was larger than healthy controls and high performance athletes by almost two fold. This large difference may be indicative of large variation in the magnitude of forces generated at the foot which may not only affect a persons' ability to successfully complete cutting and running maneuvers but also increase the likelihood of enduring a second ACL injury. This may be crucial in understanding injury recovery as those who are injured begin to return to sports.

There are a few limitations to the implementation of this study. This study was limited in the number of participants because we focused on including healthy participants that were active in running and cutting sports, injured and reconstructed participants that intended to return to running and cutting sports, and high performance athletes. While these specifications limited the number of eligible participants it does provide a cohort more related to and at risk for ACL injury. We were unable to control for type and frequency of rehabilitation provided to those subjects who had experienced an ACL tear or undergone ACL reconstruction. Evaluating, rehabilitation was beyond the scope of this study however we hope that by including participants acutely following injury and reconstructive surgery we were able to capture force control variability at appropriate time points reflecting different levels of recovery. Lastly, no fail criteria was established for the force control task to ensure collection and data analysis the variables of interest were being monitored and all trials were

inspected for both magnitude of each subjects force production and appropriate number of cycles. By monitoring number of cycles we were able to ensure subjects' maintained appropriate timing. Overall, we believe that none of these limitations grossly affect our results.

In conclusion, those who have experienced an ACL injury and subsequent reconstructive surgery exhibit poor mGRF control when compared to both uninjured controls and collegiate soccer players by a large margin. This is noted by significantly larger LyE calculated during a force control task. Poor control or more divergence in the ACL-d and ACL-r groups occurs in both the AP and ML directions. For the ACL-r group we found that the involved limb exhibited significantly greater LyE values when compared to uninvolved limb in the AP but not the ML direction. High performance athletes control mGRFs better in the ML direction than the AP direction which may be an effect of their specialized training and unique skill set. Most importantly, significantly larger force control variability by almost two-fold after ACL injury and reconstruction may identify an aspect of recovery missed by current rehabilitation protocols that could contribute to the high rates of re-injury in both ipsilateral and contralateral limbs. Addressing this deficit both prospectively and during rehabilitation would be extremely beneficial to those at risk for ACL tears.

Chapter 4

HOW DOES INJURY MECHANISM IMPACT FORCE CONTROL VARIABILITY MEASURED THROUGH LYAPUNOV EXPONENTS?

4.1 Introduction

Anterior cruciate ligament tears are a common sports injury prevalent in young adults. Over the past two decades the number of ACL ruptures in those 18 years of age and younger has been increasing (LaBella, Hennrikus, and Hewett 2014). There are approximately 100,000 ruptures annually in the United States (Frank and Jackson 2005; Prodromos et al. 2007). The most common form of treatment is reconstructive surgery which replaces the torn ligament, approximately 60-90% of those injured undergo surgery (Linko et al. 2005). ACL reconstruction is the most common treatment and has been shown to reduce knee instability and reduce the likelihood of meniscal injuries (Meunier, Odensten, and Good 2007). Unfortunately, about 25% of those injured, regardless of treatment course, will experience a re-tear in the first year after meeting return to sport criteria and returning to play (Mark V. Paterno et al. 2012).

ACL injuries can be classified as contact or non-contact owing to the manner in which the player is injured with the majority of injuries being caused by a non-contact mechanism (Boden, Griffin, and Garrett Jr. 2000). Contact injuries are a result of direct contact to the knee itself or player's body as they maneuver while non-contact injuries typically occur as the participant lands from a jump or performs a cutting maneuver. While there is extensive study on the mechanism and recovery of non-contact ACL injury, little research has explored those with ACL injuries as a result of contact (Bates and Hewett 2015; Hewett, Di Stasi, and Myer 2013; Negrete, Schick, and Cooper 2007). Contact injuries are of increasing importance in full-

contact sports like rugby and American football where a majority of injuries occurred in the lower extremities through contact (Willigenburg et al. 2016). More importantly, risk of ACL graft rupture increases 3-fold for patients that initially experienced an ACL tear through a contact mechanism (Salmon et al. 2005) While contact injuries comprise a minority of total injuries understanding recovery after such injuries may assist in reducing the high number of secondary injuries.

A majority of ACL injuries occur through a noncontact mechanism, where there is a combination of full knee extension, internal or external tibial rotation, and knee valgus (Boden, Griffin, and Garrett Jr. 2000; Rochcongar et al. 2009; Serpell et al. 2011). The noncontact injury mechanism of ACL rupture indicates a motor control deficit may be a contributing factor. Current research indicates significant increases to electromechanical delay induced by fatigue in female soccer players (De Ste Croix et al. 2015). These results suggest athletes may not be able to activate muscles quick enough to stabilize the knee joint to prevent injury. Additionally, motor control deficits are present after ACL injury and reconstruction. Post ACL injury there are significant losses to proprioception in both the sagittal and transverse planes; the inability to sense joint position may prevent protective neuromuscular control strategies (Muaidi et al. 2009). While a number of neuromuscular deficits have been identified the addition of new measurement tools may be beneficial as the number of re-tears has yet to decline.

Nonlinear analyses can be useful to understand function and recovery after ACL injury and subsequent reconstructive surgery. These analyses are valuable in their ability to capture subtle changes in biomechanical measures and have additionally been identified as a measure of neuromuscular deficits (Decker et al. 2011). One such

measure, the Lyapunov exponent (LyE), which measures divergence of small trajectories describes the predictability of dynamic systems. Reduced LyE indicates reduced divergence or a more rigid system while larger LyE values indicates more divergence and a more unstable system. Calculating LyE from knee flexion angles during gait Stergiou and colleagues found significantly larger LyE indicating more kinematic variability (Stergiou et al. 2004b). Additionally, there is significantly more force control variability (measured using LyE) after ACL injury and reconstruction as found in previous chapters. These results highlight deficits to both kinematic and kinetic control as a result of ACL injury and reconstruction.

Research indicates there is a high prevalence of ACL re-tears which can be caused through contact or non-contact mechanisms. Contact injuries may put patients at serious risk for re-injury while the many neuromuscular deficits seen after non-contact injury may also contribute to re-injuries. Non-linear analyses may be able to shed additional light on neuromuscular control and injury recovery. However, the relationship between LyE and injury mechanism has yet to be explored. This relationship may potentially be useful in understanding ACL tears and identifying factors contributing to secondary ACL injury. Therefore, the purpose of this study was to determine the relationship between force control variability, measured using LyE, and injury mechanism, contact versus noncontact, in both ACL deficient and ACL reconstructed populations. We hypothesized that, in ACL-d subjects, those who experienced a non-contact injury would have larger force control variability (LyE) when compared to those who have experienced a contact injury in both the AP and ML directions. We also hypothesized that, in ACL-r subjects, those who experienced

non-contact injuries would exhibit larger LyE values when compared to contact injuries in both the AP and ML directions.

4.2 Methods

4.2.1 Subjects

All subjects included in this study were between the ages of 14 and 55 years of age also active in >50 hrs/yr of level I and II sports were recruited from the University of Delaware physical therapy clinic. Nine subjects who have experienced an ACL rupture and eleven subjects who have received an ACL reconstruction were included in this study. ACL deficient (ACL-d) subjects included those who have experienced an isolated unilateral ACL rupture. ACL reconstructed (ACL-r) subjects included those who have undergone ACL reconstructive surgery in the last 3-9 months. This study was approved by the internal review board at the University of Delaware and all subjects were provided informed consent.

Table 4.1: Demographic data (average ± standard deviation) for ACL-d subjects.

	Age (Years)	BMI (kg/m^2)	Time since injury (Months)
Contact (6)	21 ± 4.3	24.9 ± 2.5	2.8 ± 1.3
Non-Contact (3)	29 ± 11.9	24.2 ± 2.2	4.5 ± 2.2

Table 4.2: Demographic data (average ± standard deviation) for ACL-r subjects.

	Age (Years)	BMI (kg/m^2)	Time since injury (Months)	Time since surgery (Months)
Contact (5)	25 ± 8.3	24.7 ± 4.3	11.1 ± 3.0	8.6 ± 1.8
Non-Contact (6)	18 ± 5.4	22.4 ± 3.7	8.8 ± 1.5	7.4 ± 1.5

4.2.2 Data Collection

For this study all subjects completed a force control task. This tasks utilizes real time visual feedback to understand force control. For the force control task subjects stood on two separate force platforms (AMTI OR-6, Watertown, MA) with a foot on each force plate and data sampled at 1000 Hz. Subjects received real time visual feedback of their AP or ML force production via custom written Labview code. Visual feedback was presented on a screen in front of subjects and included one slider that responded to force production and two stationary sliders that served as goals for the subjects. The two stationary sliders represented 50% of the subjects' maximum strength in that direction, based on a calibration trial.

To calibrate the force control task to each subjects' strength, subjects performed maximal force production trials. In these trials subjects pushed maximally on the force plates in the anterior, posterior, medial, and lateral directions with both feet. Subjects were instructed to control the moving cursor and align it with the two stationary sliders. They were instructed to generate force bi-directionally and continuously to the beat of a metronome set at 60 beats per minute, and to alternatively align the movable slider with each stationary slider. The force control task was two minutes in duration. The force control task was conducted for both the right and left limbs and both the AP and ML directions. During the force control task motion capture data was collected. For motion capture we used an 8 camera system (Qualisys Motion Capture Systems, Gothenburg, Sweden), retro-reflective markers were placed on bony landmarks to define the lower limbs. We also collected surface electromyography (Motion Lab Systems, Baton Rouge, LA) of eight lower limb muscles for each limb. Motion capture data was collected at 50 Hz while EMG was collected at 1000 Hz.

Additionally, all subjects for this study were participants in other studies at the University of Delaware. As part of those study clinical and functional data was collected which included injury mechanism. This data was used to dichotomize the subjects (contact vs. non-contact) within each participant cohort (ACL-d and ACL-r).

4.2.3 Data Analysis

The Lyapunov exponent was calculated for data collected during both the AP and ML force control tasks. For the force control task we calculated the Lyapunov exponent using the AP and ML force profiles generated during their respective tasks to determine AP force control variability and ML force control variability. In all cases unfiltered data was used as filtering may remove subtle changes within the signal. LyE is defined as the rate of divergence of infinitesimally close trajectories, and is determined through a multi-step process. Calculating LyE requires two input parameters time lag (τ) and embedding dimension (m). Both of these parameters convert our signals of interest into state space. To calculate τ we used the Average Mutual Information (AMI) function. This function determines the percentage of information shared between two signals, for LyE purposes the two signals of interest were the same and were either AP force profile or ML force profile. *Tau* (τ) is defined as the first local minima of the AMI function. To determine m the Global False Nearest Neighbor function (GFNN) was used. Using time delayed copies (as specified by τ) of a signal GFNN measures the distance between trajectories in one dimension then increases the number of dimensions and measures again. A false nearest neighbor is defined by any significant changes in distance when increasing the number of dimensions. As the dimensions are increased the number of false neighbors is determined. The first local minima of the GFNN function was then used as m. GFNN

is important to 'unfold' the data as any folding would lead to misinterpreted LyE. Once τ and m are determined the data is transformed in state space; in state space LyE can then be calculated. LyE was calculated using the Wolf algorithm. This algorithm calculates the Euclidean distance between trajectories and tracks the trajectories forward in time to determine the rate of divergence or convergence of the trajectories. The rate of divergence is determined for multiple trajectories and the largest, which is our value of interest, was reported.

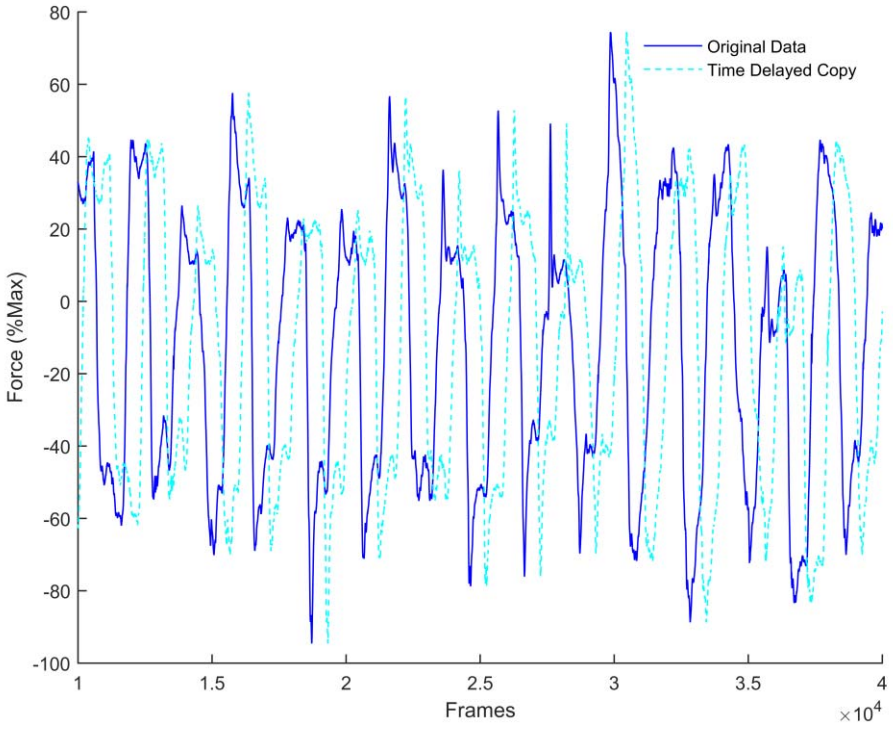

Figure 4.3: Original force profile (solid line) with a time delayed copy (dashed line).

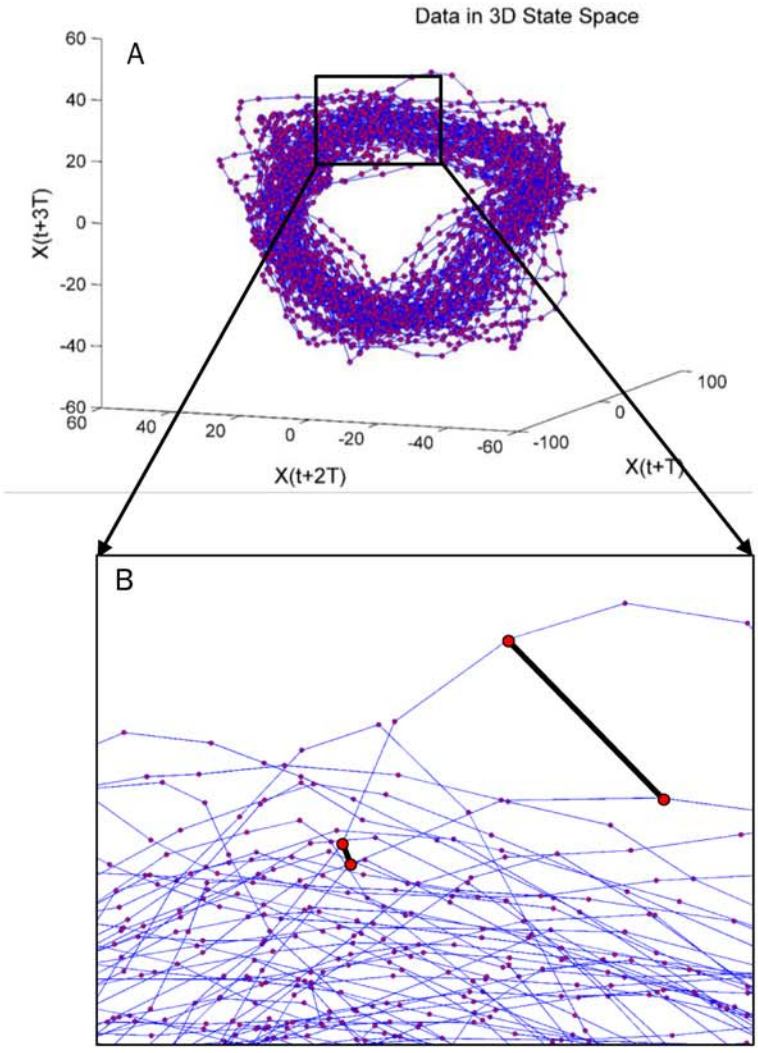

Figure 4.4: Force profile data transformed into 3D state space (A). Zoomed in view (B) with the distance between trajectories highlighted in black.

4.2.4 Statistical Analysis

For each patient population, ACL-d and ACL-r, a 2x2 (injury mechanism x direction) mixed design ANOVA was used to detect differences in force control variability. Our variable of interest was LyE calculated from the involved limb in both the AP and ML directions. A statistically significant difference was defined using an alpha level of 0.05. Additionally, effect sizes were calculated for all measures using Cohen's d with $d > 0.08$ defining a large effect size.

4.3 Results

For ACL-d subjects we found significant differences in LyE values between those who experienced a contact injury and those who experienced a non-contact injury based on direction as indicated by a significant interaction term, limb x direction (Figure 4.5, $p=0.022$). However, post-hoc t-tests indicated no significant difference between LyE of contact and noncontact injuries in either the AP or ML directions. Cohen's d values revealed a large effect size ($d=1.34$) when comparing LyE calculated in the ML direction where LyE from those with contact injuries being greater than those with non-contact injuries. In the AP direction we found a small effect size when comparing LyE of the contact and noncontact groups ($d=0.032$). On average, those who experienced a contact injury had LyE values of 7.60±2.27 bits/sec and 7.85±1.92 bits/sec for the AP and ML directions, respectively. Those who experienced a non-contact injury had LyE values of 7.54±0.55 bits/sec and 5.67±0.05 bits/sec for the AP and ML directions, respectively.

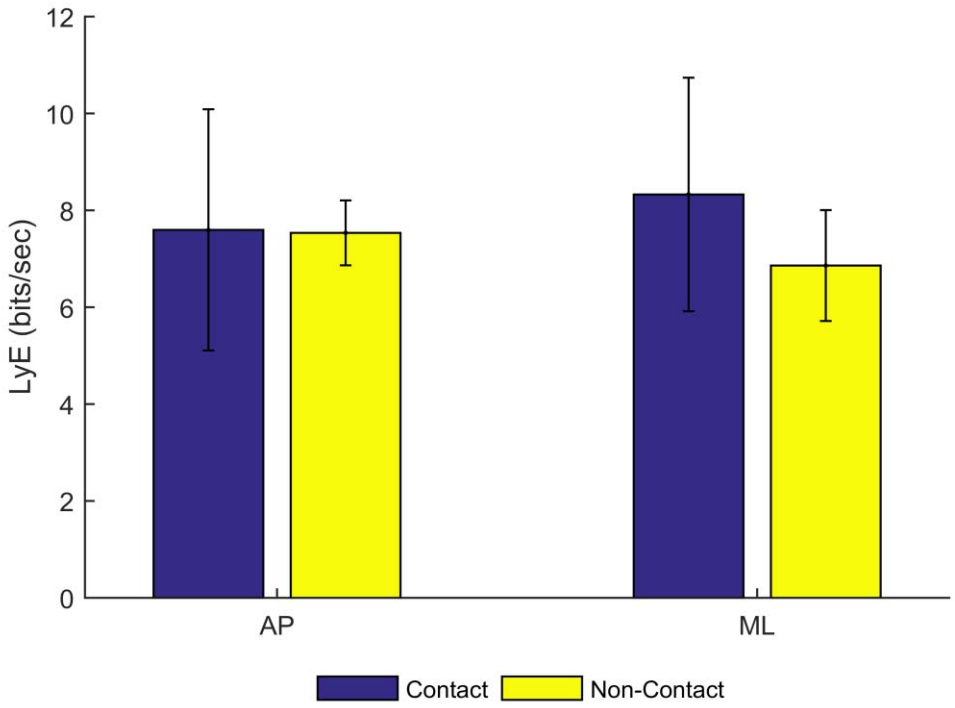

Figure 4.5: Average maximal Lyapunov Exponent (bits/second) ± standard deviation calculated during the force control task for ACL-d subjects. Contact injuries are in blue while noncontact injuries are in yellow.

We found no statistically significant difference in LyE between contact and non-contact injuries in ACL-r subjects based on ANOVA results (Figure 4.2). Cohen's d values revealed a large effect size in both the AP and ML directions, 1.14 and 0.998, respectively. For those who experienced a contact injury we calculated the LyE to be 8.87±1.79 bits/sec and 8.12±2.21 bits/sec in the AP and ML directions, respectively. Within the non-contact injury cohort of ACL-r subjects we calculated LyE to be 7.22±1.09 bits/sec and 6.58±0.61 bits/sec in the AP and ML directions respectively.

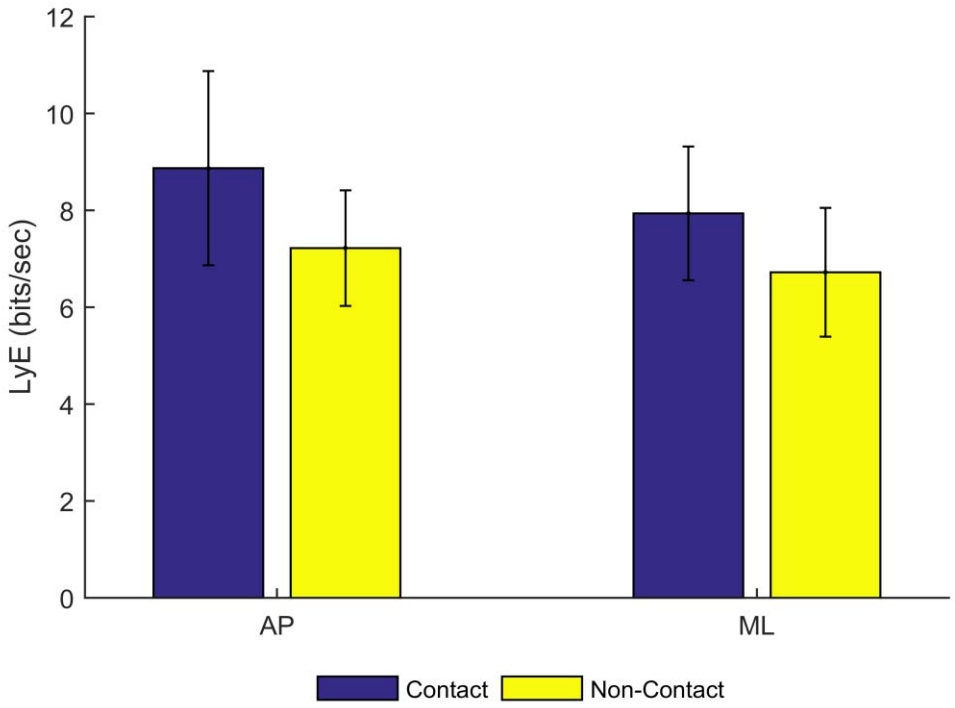

Figure 4.6: Average maximal Lyapunov Exponent (bits/second) ± standard deviation calculated during the force control task for ACL-r subjects. Contact injuries are in blue while noncontact injuries are in yellow.

4.4 Discussion

The purpose of this study was to determine the relationship between injury mechanisms and force control variability, as measured by LyE. We found that LyE values can differentiate ACL injury mechanism based on effect sizes. However, the strength of this relationship is dependent on injury status, deficient or reconstructed. Acutely following injury (ACL-d) LyE measured in the ML direction was able to differentiate contact and non-contact injuries with a large effect size, however, this was not the case for the AP direction. These results are contrary to our first hypothesis that those with noncontact injuries would exhibit larger LyE values than those who experienced a contact injury. After ACL reconstruction LyE measured from both the AP and ML directions was able to identify contact and non-contact injuries with large effect sizes. These results do not support our second hypothesis as those with contact injuries exhibited larger LyE values than those with a non-contact injury. Differences in LyE values based on injury mechanisms may identify a subset of patients with significant neuromuscular deficits who may be at higher risk for a second ACL injury.

Regardless of injury mechanism subjects generated similar LyE values in the AP direction of the ACL-d subjects. In the ML direction ACL-d subjects who experienced a contact injury generated larger LyE values than those who experienced a non-contact injury. Larger LyE values indicates more divergence or diminished mGRF control. These results are contrary to our first hypothesis, while there is a difference in LyE based on injury mechanism with those with contact injuries having larger force control variability than those with a non-contact injury. In ACL-r subjects we found that in both the AP and ML directions those who experienced a contact injury had more divergence or larger LyE values when compared to those with a non-contact injury. As with our first hypothesis, results seen in the ACL-r subjects are

contrary to our second hypothesis. After reconstructive surgery, those with contact injuries exhibited larger LyE or more divergence when compared to those with a non-contact injury.

Larger LyE values, or more force control variability in those with a contact injury may be a result of secondary tissue damage caused by a contact ACL tear. Contact ACL injuries are classified by contact directly to the knee or a collision to other parts of the players body (Krosshaug et al. 2007). In ACL-d subjects we found no difference in AP LyE based on injury mechanism, however there was a difference in ML LyE based on injury mechanism. Bone bruise patterns indicate valgus loads contribute more than anterior translation in a contact ACL injury (Viskontas et al. 2008). Larger ML LyE values, which indicates more force control variability, in our contact ACL-d group may be a result of medially directed force that initially caused injury. Sensory structures oriented in the medial/lateral direction may potentially be damaged in the initial injury leading to reduced control in that direction.

Secondary tissue damage that is more commonly seen in contact ACL injuries may contribute to larger LyE values in both AP and ML directions measured in our contact ACL-r cohort. A number of studies identify the contact injury mechanism as risk factor for lateral meniscal tears (Feucht et al. 2015) while a case study of a professional soccer player describes acute avulsion of both posterior meniscal roots concomitant with a contact ACL tear (Mariani et al. 2014). Additionally, reconstructive surgery of the ACL removes important neural mechanoreceptors (Georgoulis et al. 2001) and tibial tunnel reaming significantly reduces meniscal attachment area and ultimate strength. The global pattern of more force control variability, noted by larger LyE values, seen in both directions may be a result of the

traumatic injury at the tissue level or tissue damage incurred through surgery. Both of these factors could be a contributing factor to increased injury risk identified by Salmon and colleagues.

In execution of this study there are a few limitations that need to be addressed. For this study we focused on including subjects that participated in >50 hours per year of running and cutting sports and that intended to return to these sports after recovery. While these specifications limited the number of eligible participants it did provide a cohort most at risk for a secondary ACL injury, making this research more valuable. Unfortunately, we were unable to control for type and frequency of rehabilitation provided to subjects. Evaluating rehabilitation was beyond the scope of this study however we hope that by including participants acutely following injury and reconstructive surgery we were able to capture force control variability at appropriate time points reflecting different levels of recovery. In an effort to overcome these limitations we focused on statistical measures that can be informative without requiring large samples sizes (Cohen's d).

Overall, we found differences in force control variability based on injury mechanism in both ACL-d and ACL-r subjects. For the ACL-d subjects differences in force control variability by injury mechanism were present only in the medial direction. For ACL-r subjects we found more force control variability in those with contact ACL tears when compared to those with non-contact injury in both directions (AP, ML). Larger LyE which indicates neuromuscular deficits may be a contributing factor to the high incidence of ACL re-tears as patients return to sports. Based on these results understanding biomechanical changes that can occur as a result of contact ACL injuries may be more important than previously though. Rehabilitation

addressing these deficits, particularly in those who experienced a contact ACL tear, is crucial in achieving full recovery.

Chapter 5

DETERMINING HOW FORCE CONTROL VARIABILITY RELATES TO CUTTING PERFORMANCE

5.1 Introduction

Anterior cruciate ligament tears are a common sports related injury (Rosa et al. 2014). This injury occurs primarily during running, cutting, and jumping tasks (Boden, Griffin, and Garrett Jr. 2000; Olsen 2004). Hence, sports requiring these maneuvers see the highest number of ACL tears. Soccer players can experience up to 700 changes of direction (COD) during a single match (Bloomfield, Polman, and O'Donoghue 2007). With the high exposure to CODs and increased incidence of ACL tears sports, biomechanics research has focused on understanding cutting maneuvers. Many studies explore biomechanics of cutting while a secondary cohort utilizes agility and cutting training to prevent initial ACL injury.

Cutting maneuvers are defined as any instance where an athlete quickly and suddenly changes direction, and can be both anticipated and unanticipated, with the latter being more dangerous. As athletes initially decelerate prior to changing direction they must generate significant anterior/poster (AP) ground reaction forces (Besier and Lloyd 2001). Once deceleration is complete they must then redirect the center of mass hence producing greater ML ground reaction forces (Besier and Lloyd 2001). As speed and power increases, modulation of these multidirectional ground reaction forces (mGRFs) are crucial in providing proper body support and positioning to protect against injury. Kinetic measures provide insight into the speed and power generated during cutting, and have been shown to be related to cutting performance. Specifically, greater ML GRF impulse is significantly correlated to greater performance during side step cutting maneuvers (Havens and Sigward 2014a).

Elucidating the impact of mGRF control is important in treating injury and preventing re-injury as athletes are required to generate these forces as they return to sports and return to running and cutting.

A significant number of training programs exist that aim to reduce ACL tears in soccer athletes. One such program is called FIFA 11+. This program is a combination of 15 warm-up exercises with particular attention on posture, body control, leg alignment, knee-over-toe position, and soft landings. Some of the exercises include forwards and backwards sprints, plant and cut running, and jumping with shoulder contact (Soligard et al. 2008). FIFA 11+ has shown to reduce the number of injuries and time lost to injury in a number of different populations and improved motor control as measured by muscle activation, dynamic stability, and time to stabilization (Bizzini and Dvorak 2015). Understanding biomechanical outcomes of maneuvers included in the FIFA 11+ training program may be able to indicate the mechanisms making this training program successful.

Nonlinear analysis techniques have the potential to help understand mGRF control and lend additional insight into performance and control. In contrast to linear analysis techniques, nonlinear analysis techniques can capture subtle changes which may be valuable to understanding mGRF control. These analysis techniques have been used in a number of biomechanical applications including postural control and prosthetic design (Dingwell and Cusumano 2000; Stergiou and Decker 2011; Wurdeman, Myers, and Stergiou 2013; Wurdeman et al. 2014). One such analysis, maximal Lyapunov Exponents, indicate significant changes for kinematic variability which indicates poor kinematic control after ACL injury and reconstruction (Moraiti et al. 2010a; Stergiou et al. 2004a). Collegiate soccer players exhibit slightly larger force

control variability, as measured by LyE, when compared to healthy recreational athletes. These results may indicate those with optimal control. Further research with collegiate soccer players, specifically the relationship between control and performance, can be beneficial in designing rehabilitation and injury prevention programs.

The purpose of this study was to evaluate the relationship between force control variability and performance of sport-like tasks including a 90° cut maneuver, deceleration task, and lateral shuffle, in high performance athletes. This subject group was chosen because by other measures they have higher force control variability than non-athletes and, hence, may best demonstrate the type of variability we wish to examine. Because LyE captures force control variability through measuring divergence, our performance measures include both average and the standard deviation of our performance metrics: ML ground reaction impulse (GRI), AP ground reaction impulse, and stance time. For the lateral shuffle and cutting task, we hypothesized that average ML GRI, standard deviation ML GRI, average stance time, and standard deviation of stance time would correlate with ML force control variability. We additionally hypothesized that average AP GRI, standard deviation of AP GRI, average stance time, and standard deviation of stance time would correlate with AP force control variability in the deceleration task.

5.2 Methods

5.2.1 Subjects

Seventeen athletes from the University of Delaware men's and women's soccer team were recruited for this study (Table 5.1). This study was approved by the internal

review board at the University of Delaware and all subjects were provided informed consent.

Table 5.1: Demographic data (average ± standard deviation) for collegiate soccer players.

	n	Age (Years)	BMI (kg/m^2)
Male Athletes	9	21 ± 1.5	24.1 ± 1.2
Female Athletes	8	20 ± 0.9	21.5 ± 1.8

5.2.2 Force Control Task

For this study all subjects completed a force control task. This tasks utilizes real time visual feedback to understand force control. For the force control task subjects stood on two separate force platforms (AMTI OR-6, Watertown, MA) with a foot on each force plate and data sampled at 1000 Hz. Subjects received real time visual feedback of their AP or ML force production via custom written Labview code. Visual feedback was presented on a screen in front of subjects and included one slider that responded to force production and two stationary sliders that served as goals for the subjects. The two stationary sliders represented 50% of the subjects' maximum strength in that direction, based on a calibration trial.

To calibrate the force control task to each subjects' strength, subjects performed maximal force production trials. In these trials subjects pushed maximally on the force plates in the anterior, posterior, medial, and lateral directions with both feet. Subjects were instructed to control the moving cursor and align it with the two stationary sliders. They were instructed to generate force bi-directionally and continuously to the beat of a metronome set at 60 beats per minute, and to

alternatively align the movable slider with each stationary slider. The force control task was two minutes in duration. The force control task was conducted for both the right and left limbs and both the AP and ML directions. During the force control task motion capture data was collected. For motion capture we used an 8-camera system (Qualisys Motion Capture Systems, Gothenburg, Sweden), retro-reflective markers were placed on bony landmarks to define the lower limbs. We also collected surface electromyography (Motion Lab Systems, Baton Rouge, LA) of eight lower limb muscles for each limb. Motion capture data was collected at 50 Hz while EMG was collected at 1000 Hz.

5.2.3 LyE Data Analysis

The Lyapunov exponent was calculated for data collected during both the AP and ML force control tasks. For the force control task we calculated the Lyapunov exponent using the AP and ML force profiles generated during their respective tasks to determine AP force control variability and ML force control variability. In all cases unfiltered data was used as filtering may remove subtle changes within the signal. LyE is defined as the rate of divergence of infinitesimally close trajectories, and is determined through a multi-step process. Calculating LyE requires two input parameters time lag (τ) and embedding dimension (m). Both of these parameters convert our signals of interest into state space. To calculate τ we used the Average Mutual Information (AMI) function. This function determines the percentage of information shared between two signals, for LyE purposes the two signals of interest were the same and were either AP force profile or ML force profile. *Tau* (τ) is defined as the first local minima of the AMI function which indicates the amount of time delay required for the two signals to be considered independent (Figure 5.1). To determine

m the Global False Nearest Neighbor function (GFNN) was used. Using time delayed copies (as specified by τ) of a signal GFNN measures the distance between trajectories in one dimension then increases the number of dimensions and measures again. A false nearest neighbor is defined by any significant changes in distance when increasing the number of dimensions. As the dimensions are increased the number of false neighbors is determined. The first local minima of the GFNN function was then used as *m*. GFNN is important to 'unfold' the data as any folding would lead to misinterpreted LyE. Once τ and *m* are determined, the data are transformed in state space; in state space LyE can then be calculated (Figure 5.2). LyE was calculated using the Wolf algorithm. This algorithm calculates the Euclidean distance between trajectories and tracks the trajectories forward in time to determine the rate of divergence or convergence of the trajectories. The rate of divergence is determined for multiple trajectories and the largest, which is our value of interest, was reported.

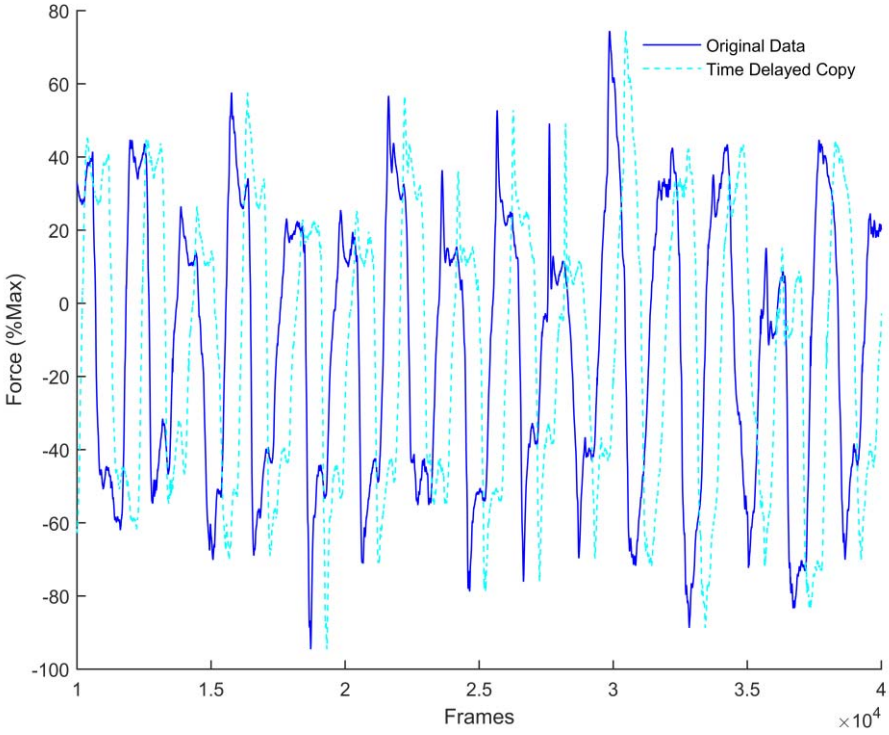

Figure 5.1: Original force profile (solid line) with a time delayed copy (dashed line).

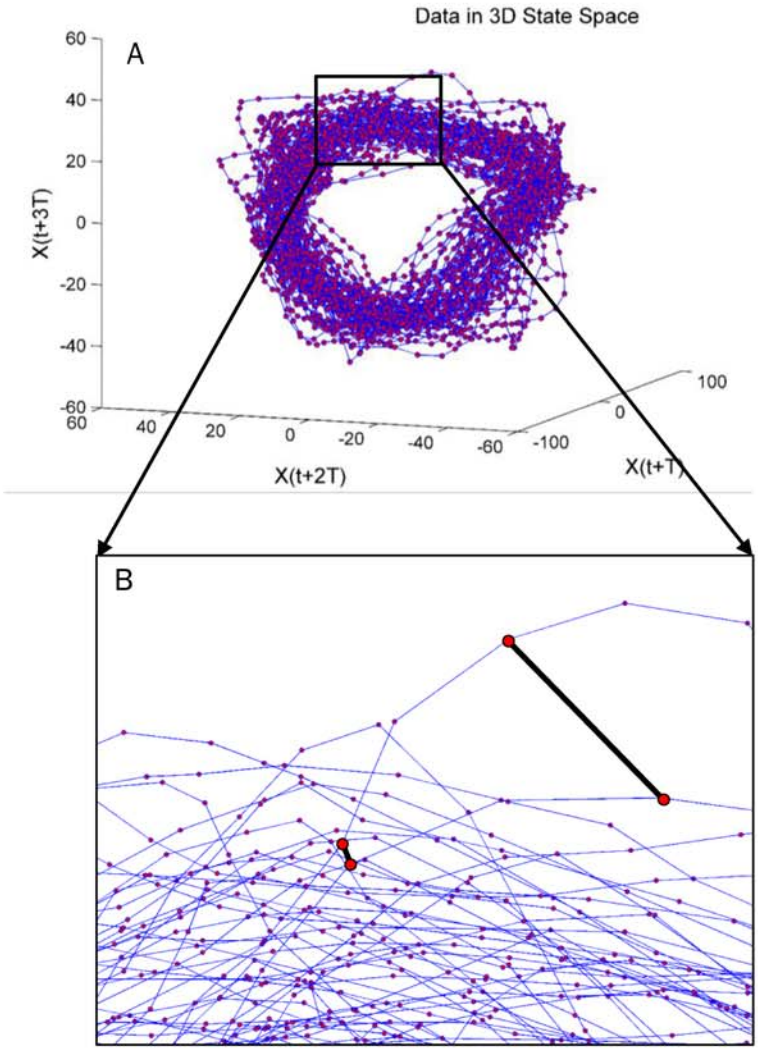

Figure 5.2: Force profile data transformed into 3D state space (A). Zoomed in view (B) with the distance between trajectories highlighted in black.

5.2.4 Motion Capture

Subjects were asked to complete three maneuvers for this study. The tasks included a lateral shuffle, a 90° cut, and a deceleration maneuver. For the lateral cut subjects were instructed to run forward, plant their foot on the force plate, complete a 90° turn, and continue running. Subjects were encouraged to perform all maneuvers as if they were in a competitive setting. The lateral shuffle maneuver was 7m and subjects were instructed to shuffle back and forth while planting a single limb at the end of each shuffle on the force platform before changing directions. For the deceleration task subjects were asked to run towards the force platform, plant a single limb on the platform and reverse direction completing the task by running backwards to the location where they started. Each limb was tested for all maneuvers. During these maneuvers both kinematic and kinetic data will be collected. Kinetic data was collected via six component embedded force platforms (Bertec Corp, Worthington, OH) sampled at 1080 Hz. Kinematic data will be collected using an eight camera motion capture system (VICON, Oxford Metrics Ltd., London, England) sampled at 120 Hz. Our variables of interest were ML and AP ground reaction force impulse (ML GRI, AP GRI) and stance time. These measures were calculated using Visual 3D software (C-Motion Inc., Germantown, MD). AP and ML GRI was defined as the integral of the AP and ML ground reaction force during the stance phase of each task. Stance phase was calculated using a force threshold of 50 N and stance time was calculated as the duration of the stance phase.

5.2.5 Statistical Analysis

The relationship between force control variability and performance (average and the standard deviation: ML GRI, AP GRI, and stance time) during sport-like

maneuvers was determined by calculating the Pearson correlation coefficient (r). A moderate correlation was defined by those having an r value between 0.3 and 0.5, while strong correlations were defined by r values greater than 0.5.

5.3 Results

Based on our analysis, there were no strong correlations between force control variability and our task performance measures for all tasks. There were a few moderate correlations based on Pearson correlation coefficients. However, no variables exhibited significant correlations.

When looking at average values of our performance metrics, Pearson correlation coefficients identify moderate correlations between metrics and LyE for each task. For the cutting and lateral shuffle tasks, our analysis focused on the relationship between performance metrics and ML LyE. In the cutting task, we found moderate correlations between average stance time in both limbs and right limb ML LyE (Table 5.2). In both cases, we found a negative correlation which indicates that as LyE increases stance time decreases (Figure 5.3). For the lateral shuffle, there was a moderate correlation between average left limb stance time and right limb ML LyE (Table 5.2). As seen in the cutting task this correlation is also negative. Analysis of the deceleration task focused on the relationship between performance metrics and AP LyE. Average right limb stance time was moderately correlated with both right limb and left limb AP LyE; both correlation coefficients were negative (Figure 5.4, Table 5.2). For all tasks, we found no moderate or strong correlations for both ML and AP GRI.

Table 5.2: Pearson correlation coefficients and corresponding p-values for average stance time (seconds), average medial/lateral ground reaction impulse (Ns/N), average anterior/posterior ground reaction impulse (Ns/N), and average maximal Lyapunov exponents (bits/sec). (DC= Deceleration Task, LS= Lateral Shuffle Task)*r $\geq$ 0.3

		Avg. Stance Time						Avg. ML GRI				Avg. AP GRI	
		Right Cut	Left Cut	Right DC	Left DC	Right LS	Left LS	Right Cut	Left Cut	Right LS	Left LS	Right DC	Left DC
R AP LyE	r	--	--	-0.467*	-.201	--	--	--	--	--	--	-.279	-.130
	p-value	--	--	.059	.439	--	--	--	--	--	--	.279	.620
L AP LyE	r	--	--	-0.378*	-.265	--	--	--	--	--	--	-.092	.181
	p-value	--	--	.149	.320	--	--	--	--	--	--	.736	.502
R ML LyE	r	-0.442*	-0.413*	--	--	.102	-0.345*	-.220	-.231	-.059	-.184	--	--
	p-value	.076	.100	--	--	.697	.174	.395	.373	.822	.480	--	--
L ML LyE	r	.012	-.091	--	--	.047	-.158	-.085	-.206	-.011	.041	--	--
	p-value	.964	.737	--	--	.864	.559	.756	.445	.967	.879	--	--

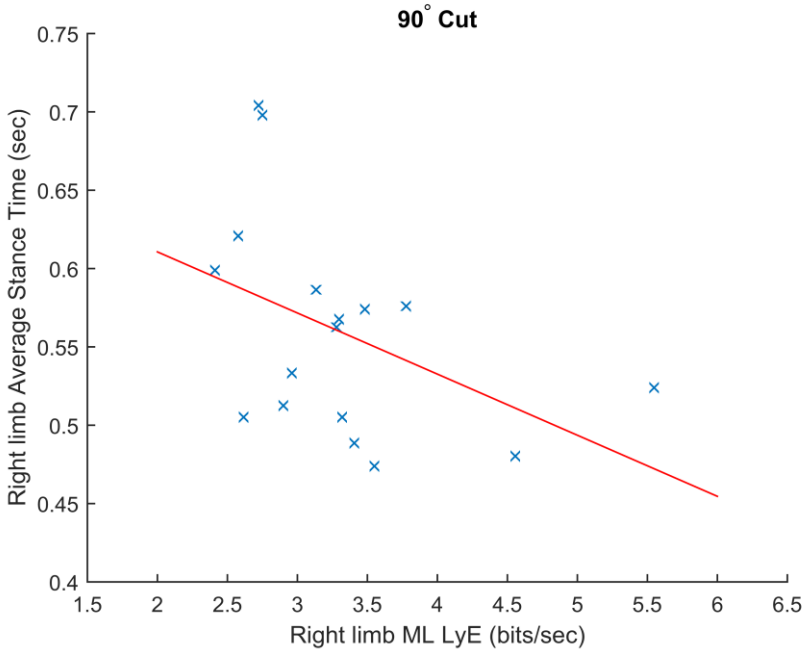

Figure 5.3: Average right limb stance time (seconds) during a 90° cutting task plotted as a function of average right limb ML LyE (bits/sec) for collegiate soccer players. Linear fit indicated in red.

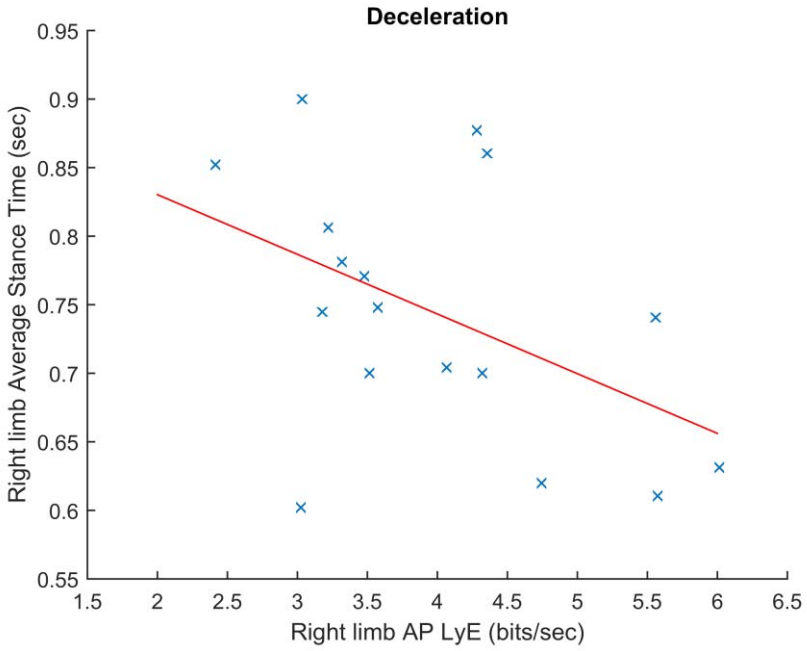

Figure 5.4: Average right limb stance time (seconds) during a deceleration task plotted as a function of average right limb AP LyE (bits/sec) for collegiate soccer players. Linear fit indicated in red.

Correlation coefficients calculated from AP and ML LyE and the standard deviation of performance metrics identify moderate correlations in at least one measure for each task. During cutting maneuvers, results indicate a moderate correlation between the standard deviation of right limb stance time and right limb ML LyE (Figure 5.5, Table 5.3). In contrast to the average stance time this relationships exhibits a positive relationship. As right limb ML LyE increases the standard deviation of stance time also increases (Figure 5.5). Evaluation of the lateral shuffle reveals a moderate correlation between left limb stance time standard deviation and right limb ML LyE (Table 5.3). Again, the correlation coefficient is positive; as right limb ML LyE increases, left limb stance time standard deviation also increases (Figure 5.6). For the deceleration task there were numerous moderate correlations. First, right limb stance time standard deviation was moderately positively correlated with both right and left limb AP LyE (Table 5.3). Second, left limb stance standard deviation and left limb AP LyE were moderately and positively correlated (Table 5.3). Lastly, right limb AP GRI standard deviation had a moderate negative correlation with left limb AP LyE (Figure 5.7, Table 5.3).

Table 5.3: Pearson correlation coefficients and corresponding p-values for stance time standard deviation (seconds), medial/lateral ground reaction impulse standard deviation (Ns/N), anterior/posterior ground reaction impulse standard deviation (Ns/N), and average maximal Lyapunov exponents (bits/sec). (DC= Deceleration Task, LS= Lateral Shuffle Task)*r ≥ 0.3

		Stance Time Std.						ML GRI Std.				AP GRI Std.	
		Right Cut	Left Cut	Right DC	Left DC	Right LS	Left LS	Right Cut	Left Cut	Right LS	Left LS	Right DC	Left DC
R AP LyE	r	--	--	0.347*	0.202	--	--	--	--	--	--	-0.057	0.177
	p-value	--	--	0.173	0.437	--	--	--	--	--	--	0.828	0.496
L AP LyE	r	--	--	0.432*	0.465*	--	--	--	--	--	--	-0.378*	-0.098
	p-value	--	--	0.084	0.060	--	--	--	--	--	--	0.133	0.708
R ML LyE	r	0.327*	0.266	--	--	0.158	0.336*	-0.144	0.288	-0.167	0.035	--	--
	p-value	0.216	0.319	--	--	0.559	0.203	0.594	0.279	0.537	0.896	--	--
L ML LyE	r	-0.014	0.087	--	--	0.090	0.165	-0.049	-0.231	-0.091	-0.236	--	--
	p-value	0.959	0.750	--	--	0.739	0.541	0.858	0.389	0.738	0.378	--	--

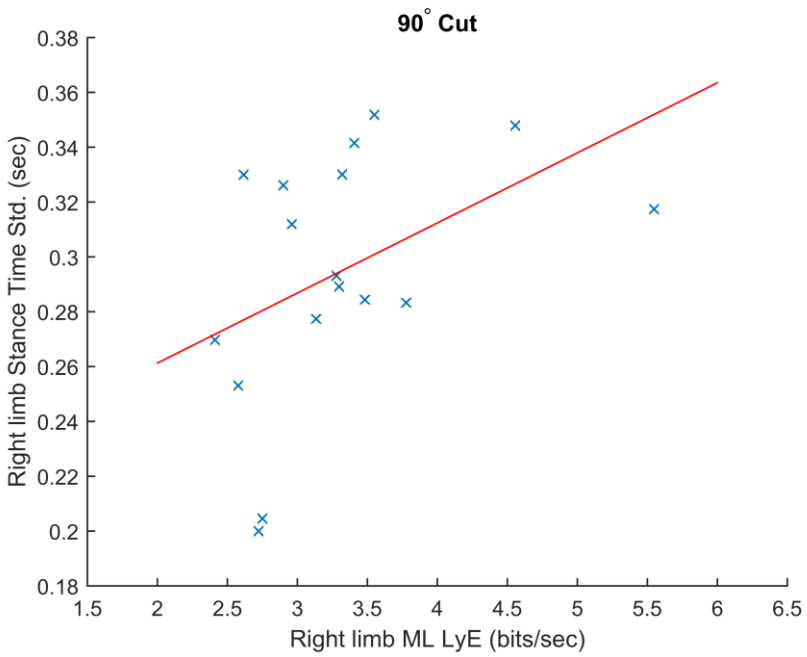

Figure 5.5: Right limb stance time standard deviation (seconds) during a 90° cutting task plotted as a function of average right limb ML LyE (bits/sec) for collegiate soccer players. Linear fit indicated in red.

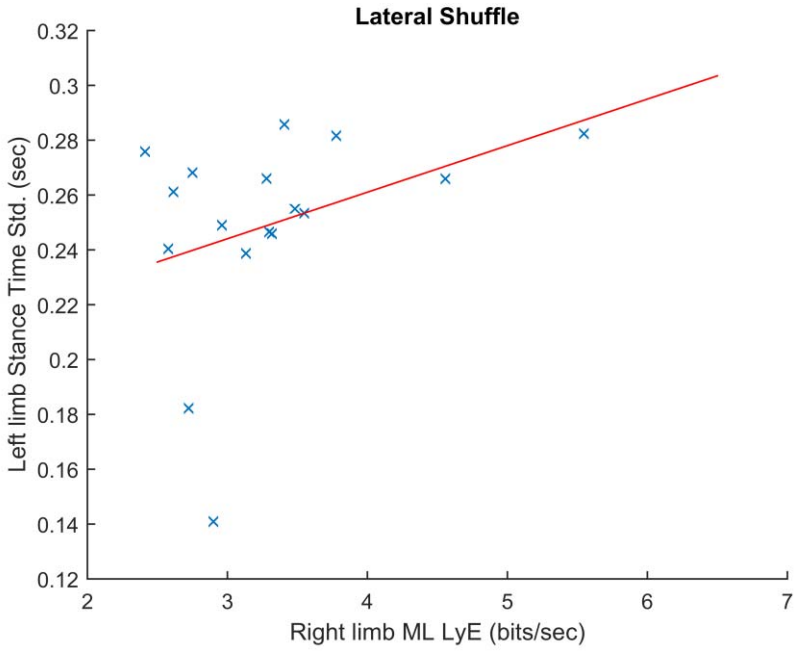

Figure 5.6 Left limb stance time standard deviation (seconds) during a lateral shuffle task plotted as a function of average right limb ML LyE (bits/sec) for collegiate soccer players. Linear fit indicated in red.

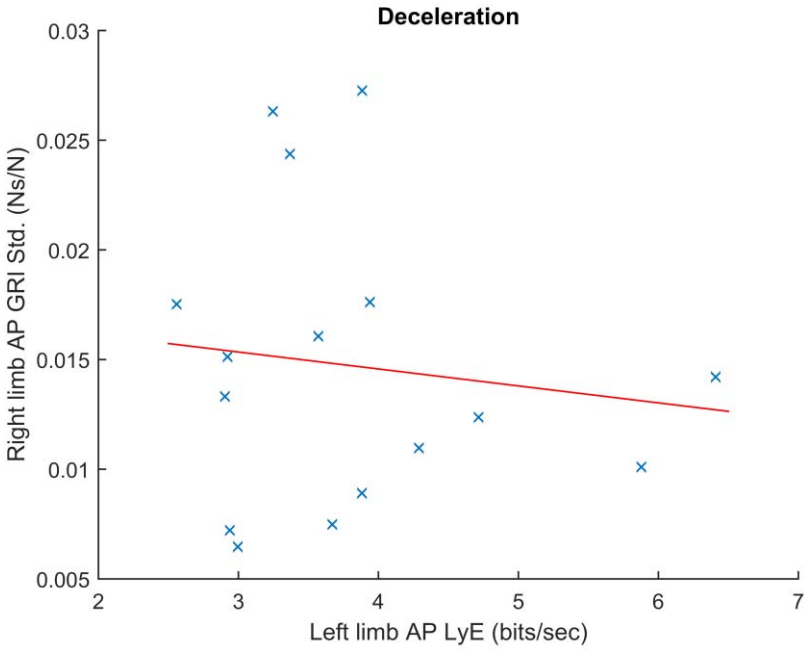

Figure 5.7: Right limb AP GRI standard deviation (Ns/N) during a deceleration task plotted as a function of average left limb AP LyE (bits/sec) for collegiate soccer players. Linear fit indicated in red.

5.4 Discussion

The goal of this study was to determine the relationship between force control variability, measured by LyE, and performance of sport-like tasks in collegiate soccer players. We found that there were no statistically significant relationships between our performance metrics (average stance time, stance time standard deviation, average AP GRI, AP GRI standard deviation, average ML GRI, and ML GRI standard deviation) and force control variability for all tasks included in this study. However, a number of moderate correlations were identified.

Our analysis focused on both averages and standard deviations of the three performance metrics (AP GRI, ML GRI, and stance time). We will first focus on the cutting and lateral shuffle tasks. During both the cutting and lateral shuffle tasks, only average stance time was correlated with ML LyE while there was no correlation found between average ML GRI and ML LyE. In both tasks, these relationships had negative correlations. As force control variability increased, stance time decreased. In our analysis of stance time and ML GRI variability, moderate correlations were also identified during the cutting and lateral shuffle tasks. As with average stance time, ML LyE during the cutting task moderately correlated to stance time standard deviations. During the lateral shuffle, stance time standard deviation also correlated with ML LyE. In both the cutting and lateral shuffle tasks, the performance metric (stance time standard deviation) was positively with force control variability (ML LyE). As stance time variability increased ML LyE also increased. Interestingly, for both tasks average stance time and stance time standard deviation only correlated with right limb ML LyE. In our analysis of the cutting and lateral shuffle task, results partially support our hypothesis, as force control variability correlated with stance time but not kinetic measures (ML GRI).

In contrast to the lateral shuffle and cutting tasks, we focused on identifying relationships between stance time, AP GRI, and AP LyE. For the deceleration task average stance time, stance time standard deviation, and AP GRI correlated with AP force control variability (LyE). Average stance time and AP GRI standard deviation were negatively correlated with AP LyE; as LyE increased these two measures decreased. Contrastingly, stance time standard deviation was positively correlated with AP LyE. Again, our results partially supported our second hypothesis regarding the deceleration task.

In all tasks, there was a moderate correlation between average stance time and force control variability and also between stance time standard deviation and force control variability. Our correlations indicate that as force control variability increased stance time decreased. Larger LyE values indicate more divergence or force control variability. Impact loading of the knee has been identified as a risk factor for ACL injuries (Chappell et al. 2002; Devita and Skelly 1992; Yu, Lin, and Garrett 2006). In all tasks performed in this study subjects must decelerate their body mass and change direction whether that be medially, laterally, or posteriorly. With this in mind it can be assumed that all tasks had similar loading and so a reduction in stance time may lead to an increase in impact loading. As impact loading increases in an effort to protect the limb, subjects may be focusing on stabilizing their body and lower limb rather than providing accurate mGRF control. Athletes must make a decision between limb preservation and force control and so mGRF control suffers.

While both average and standard deviation of ML GRI and AP GRI were considered as performance metrics for the sport-like tasks conducted for this study only AP GRI standard deviation was moderately correlated with AP LyE during the

deceleration task. The moderate negative correlation specifies that as LyE increased AP GRI standard deviation decreased, so as force control variability increased GRI variability decreased. Unfortunately, there is limited research on impulse loading variability during different sport-like tasks and current research evaluating ground reaction impulse magnitudes during cutting provides mixed results. In evaluating cutting performance, Havens and Sigward, found that increases in ML GRI correlated with better performance as measured by performance time (Havens and Sigward 2014a). However, vertical ground reaction impulse asymmetries predicted asymmetries at the knee during both stop-jump and side-cutting tasks (Dai et al. 2013). While anterior-posterior ground reaction impulse was strongly positively correlated with knee joint kinetics during a stop-jump task (Dai et al. 2013). Studies by Havens and Sigward indicate increases to ML GRI improve performance while Dai and colleagues found that increases to impulse increase knee joint loading implying increased injury risk. The fact that many studies find strong relationships between impulse loading and cutting performance highlights the importance of these measures, unfortunately, the functional impact of this relationship has yet to be seen and requires further exploration.

While a majority of moderate correlations identified by this study were within the same limb there were a few correlations that were across limb. For example, left limb average stance time during the cutting task moderately and negatively correlated with right limb ML LyE. Across limb correlations were identified in each task. Earlier work found no significant difference in LyE generated by the right or left limb, but differences in task performance between limb was not evaluated. These correlations may be an effect of the bilateral nature of the force control task.

Additionally, while upper extremity limb differences can be significant this is not the case for the lower limb. In soccer players and sedentary participants Cug and colleagues found no limb dominance effect on knee proprioception (Cug et al. 2016). Additionally, change of direction ability evaluated for both dominant and non-dominant limbs showed mixed performance; in some cases the dominant limb performed better and vice versa (Rouissi et al. 2016). For our study the right and left limbs may be completing each task (force control and sport-like) similarly with each limb which influences correlations across limbs.

A lack of strong correlations between force control variability and our performance metrics may be a consequence of non-linear analysis demands. Lyapunov exponents describe the predictability of a dynamic system and in doing so require large amounts of data for valid calculations. The force control testing for this study required subjects to complete at least 60 cycles of each test to meet the data requirements for LyE analysis. However, for the task performance component subjects only completed three trials of each maneuver. This many not have been enough data to fully describe task performance. However, it would be difficult to have subjects complete a large number of dynamic tasks without inducing fatigue.

There are a few limitations to the implementation of this study. The study participants included athletes that played a variety of positions and with varying levels of play-time. While it would be ideal to have subjects that had the most game-time and played positions that experienced the highest number of cutting maneuvers all participants were included in team training. So it can be assumed they were exposed to an extensive amount of agility training. In regards to the force control task, no fail criteria were established to ensure subjects completed achieved a certain level of

accuracy. However, during data collection and data analysis the variables of interest were being monitored and all trials were inspected for both magnitude of each subjects force production and appropriate number of cycles. By monitoring number of cycles, we were able to ensure subjects' maintained appropriate timing. Overall, we believe that none of these limitations grossly affect our results.

In conclusion, the study found that as force control variability increases there are reductions to stance time which may be a result of a biomechanical re-prioritization to provide lower limb stability over mGRF control. Additionally, force control variability increased as AP GRI variability decreased during a deceleration task. While we found no statistically significant correlations results, our results may have been limited by the performance metrics used. Significant changes to force control variability are seen after ACL injury and reconstruction and so focusing on biomechanical measures more closely related to this injury may provide stronger relationship. Some of these measures may include knee joint loading and lower extremity muscle activity. Exploring the relationship between force control variability and muscle activity may lend additional insight into the relationship seen between stance time and ML LyE values which may in turn identify modifiable factors than can be targeted for rehabilitation.

Chapter 6

CONCLUSIONS

This dissertation utilized nonlinear analysis techniques to quantify force control variability in uninjured recreational athletes, those who experienced an anterior cruciate ligament injury, those who underwent anterior cruciate ligament reconstruction, and collegiate soccer players. A novel task utilizing real time visual feedback of a person's force production was used to measure maximal Lyapunov exponents. The force control task was able to help quantify force control variability for each limb in the anterior/posterior and medial/lateral directions which may be relatable to mGRF production during running and cutting. From this task we were able to identify changes to force control variability after ACL injury and reconstruction. We then explored the connection between force control variability and injury mechanism. Finally, the relationship between force control variability and sports performance was evaluated in collegiate soccer players.

6.1 Major Findings

6.1.1 Force control variability in recreational athletes and high performance athletes

In the first step to understand changes to force control variability as a result of ACL injury and reconstruction, force control variability was evaluated in both uninjured recreational athletes (healthy controls) and high performance athletes. These studies were conducted to evaluate limb differences (right versus left) and directional differences (AP versus ML) in force control variability. For both groups there were no significant differences in LyE when comparing right versus left limb. In the healthy uninjured recreational athletes, we found no statistically significant

differences in LyE calculated for the ML versus AP directions. In the high performance athletes, data analysis revealed a statistically significant difference in LyE based on directions; LyE calculated in the ML direction was significantly lower when compared to that of the AP direction.

Collegiate soccer players exhibited lower LyE values in the ML direction when compared to the AP direction indicating better mGRF control in that direction. This trend may be a protective measure to prevent ACL injury, a result of the different roles of lower limb muscles, or a result of strenuous change of direction focused training. Frontal plane loading at the knee has been identified as an inciting factor in ACL injury and contributing factor in OA progression. Superior ML mGRF control noted by reduced ML force control variability may lead to reduced variability in frontal plane knee moments creating a better environment for the knee. Additionally, specified muscle control may be better equipped to control ML force control variability rather than AP force control variability as indicated in a similar force control task. Lastly, controlling superior ML mGRF control may be an effect of specialized training and unique skill set of collegiate soccer players.

Evaluating LyE in recreational and high performance atheletes establishes a standard for force control variability as subjects generate mGRFs. While a number of studies have explored the changes to the structure of variability in kinematic measures no work has been done utilizing kinetic measures. These results have established a standard for healthy and active young adults that will help understand changes to force control variability that may occur as a result of anterior cruciate ligament injury and reconstruction or participation in high performance athletics which can inform both rehabilitation practices and sports training.

6.1.2 Force control variability after ACL injury and reconstruction

After establishing a standard for force control variability in healthy uninjured participants and high performance athletes, our study was expanded to include ACL deficient (ACL-d) individuals and ACL reconstructed (ACL-r) individuals. This study found that acutely following ACL injury and reconstruction there are significantly larger LyE values when compared to healthy uninjured subjects and collegiate soccer players. In ACL-d subjects larger force control variability measurements were present in both the involved and uninvolved limbs with no significant differences between directions. While ACL-r subjects exhibited no differences in LyE values when comparing directions we did identify differences between LyE of the involved and uninvolved limbs in the AP direction.

Reduced mGRF control in the reconstructed limb is not surprising as surgery is very chaotic for the knee joint causing meniscal destabilization and removal of mechanoreceptors which commonly leads to reduced proprioception (Muaidi et al. 2009), reduced coordination variability (Pollard et al. 2014), altered muscle activation (Macleod 2012), and altered joint kinematics (Georgoulis and Papadonikolakis 2003; Hofbauer et al. 2014). This combination of these deficits may translate to poor control of mGRFs at the foot. In the ACL-d cohort, overall larger LyE values which indicates globally poor motor control may be a result of upper level motor dysfunction. Seen both in animal models and in vivo (Laurin et al. 2015; A. S. Lepley et al. 2015).

Those who have experienced an ACL injury and subsequent reconstructive surgery exhibit poor mGRF control when compared to both uninjured controls and collegiate soccer players that is present in both the AP and ML directions by a large margin. Significantly more force control variability after ACL injury and reconstruction may identify an aspect of recovery missed by current rehabilitation

protocols that could contribute to the high rates of re-injury. Addressing this deficit both prospectively and during rehabilitation would be extremely beneficial to those at risk for ACL tears.

6.1.3 ACL injury mechanism and force control variability

In an effort to explore further changes in force control variability and ACL injury and reconstruction, clinical data were used to identify sub-groups of the ACL-d and ACL-r groups and evaluate LyE differences. For this study we dichotomized both ACL-d and ACL-r subjects by injury mechanism (contact versus non-contact). Due to small sample size both ANOVA and effect size calculations were used to compare LyE values for our groups. Using effect size(Cohen's d) as an indicator, we found differences in force control variability based on injury mechanism in both ACL-d and ACL-r subjects. For the ACL-d subjects differences in force control variability by injury mechanism were present only in the ML direction. For ACL-r subjects larger force control variability measurements in those with contact ACL tears were present in both directions. Overall, those who experienced a contact ACL injury exhibited larger LyE values than those who experienced a noncontact injury. This trend may be a consequence of the traumatic nature and secondary tissue damage typically seen in a contact injury. The neuromuscular deficits identified in this study may be a contributing factor to the high incidence of ACL re-tears as patients return to sports. Rehabilitation addressing these deficits, particularly in those who experienced a contact ACL tear, is crucial in achieving full recovery.

6.1.4 Sports performance and force control variability

The goal of this study was to evaluate performance of sport-like tasks as it relates to changes in force control variability. In this study, collegiate soccer players performed cutting, lateral shuffle, and deceleration tasks in addition to completing the force control task as described in Aim 2. Performance was evaluated using average stance time, stance time standard deviation, average AP GRI, AP GRI standard deviation, average ML GRI, ML GRI standard deviation as metrics. Pearson correlation coefficients were calculated to determine relationships between the performance measures and LyE values.

This study found that as force control variability increases there are reductions to stance time which may be a result of a biomechanical re-prioritization to provide lower limb stability over mGRF control. Additionally, force control variability increases as AP GRI variability decreases during a deceleration task. Significant changes to force control variability are seen after ACL injury and reconstruction and so focusing on biomechanical measures more closely related to this injury may provide stronger relationship. Some of these measures may include knee joint loading and lower extremity muscle activity. Exploring the relationship between force control variability and muscle activity may lend additional insight into the relationship seen between stance time and ML LyE values which may in turn identify modifiable factors than can be targeted for rehabilitation.

6.2 Future Work

6.2.1 Evaluating Drop Vertical jump

From this work we have gained an understanding of how force control variability influences sport-like task performance. In this study only a limited number

of tasks were used with cutting and cutting related tasks being the most important. However, in addition to cutting maneuvers jumping tasks have been highly researched in relation to ACL injuries. Mechanics of jumping tasks have been able to predict ACL injury in female athletes (Hewett et al. 2005) and have been identified as a major contributor to noncontact ACL injuries (Olsen 2004). To gain further understanding of LyE changes after ACL injury and reconstruction the relationship between jumping performance and LyE values would be beneficial. In addition to adding jumping tasks to force control variability studies

6.2.2 Evaluating clinical impact

Understanding the relationship between force control variability and injury mechanism is the first step to elucidating the role nonlinear analysis techniques play in injury rehabilitation. Over the course of rehabilitation a number of performance based and patient reported outcomes are used to evaluate patient health (Adams et al. 2012). These evaluation tools not only allow patients to progress in rehabilitation and return to sport like activities but also capture a more complete picture of the patient. Understanding how force control variability relates to these measures may be able to identify unexplored pathways in current rehabilitation, provide a better more comprehensive understanding of neuromuscular control, and highlight those at a greater risk for re-injury.

6.2.3 Modifying LyE post ACL injury

Since significantly larger force control variability values persist after ACL injury and reconstruction current rehabilitation is unable to address this alteration to neuromuscular control. Future work should focus on identifying modifiable factors

which influence force control variability. This information would be highly valuable in a clinical setting. In conjunction with this work the development of clinical tools to address force control variability would be highly valued. The force control task was a useful research condition, however, its clinical implementation would be difficult at its current state. Simplifying this task for use in a clinic or even at home could provide superior rehabilitation.

6.3 Conclusions

This body of work has established a standard for force control variability, identified changes to force control variability after ACL injury and reconstruction, and explored the relationships between force control variability and sport performance and injury mechanism. While alterations to kinematic variability have been identified in ACL-d and ACL-r subjects using Lyapunov exponents, this was the first study to explore force control variability. We found there are significantly larger LyE values after ACL injury and reconstruction in both the involved and uninvolved limbs. Current rehabilitation protocols are not addressing this deficit which may be contributing to the high re-injury rates after patients return to sport. From this initial finding we explored the connection between LyE and contact versus noncontact injuries. Surprisingly, those who experienced contact injuries had significantly larger LyE values when compared to those with noncontact injuries. These mGRF control deficits may be a result of secondary tissue damage seen more commonly after contact ACL tears. Lastly, we showed that both magnitude and variability of performance metrics influence force control variability during sport-like tasks. Overall, this work highlights the importance of utilizing nonlinear analysis techniques in biomechanics; we have been able to identify phenomena that may contribute to injury, re-injury, and

improve sports performance. Further exploration in these areas will ultimately better inform physicians, physical therapists, and coaches which will lead to better overall understanding of biomechanics and neuromuscular control.

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Appendix

IRB APPROVAL DOCUMENTATION

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	June 1, 2015
TO:	Amelia Lanier
FROM:	University of Delaware IRB
STUDY TITLE:	[617232-4] Directional and Dynamic Stability after ACL Injury and Reconstruction
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	June 1, 2015
EXPIRATION DATE:	June 17, 2016
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(9)*

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Generated on IRBNet

University of Delaware
Informed Consent Form
-Dynamic and Directional Stability-

Title of Project: Directional and Dynamic Stability after ACL injury and reconstruction

Principal Investigator (s): Amelia S. Lanier, Thomas S. Buchanan

Other Investigators: Kurt Manal

You are being asked to participate in a research study. This form tells you about the study including its purpose, what you will be asked to do if you decide to participate, and any risks and benefits of being in the study. Please read the information below and ask the research team questions about anything we have not made clear before you decide whether to participate. Your participation is voluntary and you can refuse to participate or withdraw at any time without penalty or loss of benefits to which you are otherwise entitled. If you decide to participate, you will be asked to sign this form and a copy will be given to you to keep for your reference.

STUDY DESCRIPTION

You are invited to participate in a research study conducted by Dr. Thomas S. Buchanan which will compare the stability of force production and treadmill walking of healthy young adults, those who have experienced an anterior cruciate ligament (ACL) injury and those who have undergone ACL reconstructive surgery. The purpose of this study is to evaluate knee joint stability when generating specified forces in multiple directions and when walking on a treadmill in these groups. This project is part of a doctoral dissertation.

Participants

You are being asked to take part in this study because you are either a healthy active, young adult with no history of knee injury, have experienced an ACL injury, or have undergone ACL reconstructive surgery. You must also participate in jumping, pivoting, and cutting activities greater than 50 hours per year. You are being asked to take part in a study that will have 85 total participants with 35 uninjured subjects, 25 ACL injured subjects, and 25 ACL reconstructed subjects..

You should not participate in this project if you are currently pregnant or have a muscle, bone or nervous system disorder. Additionally, you should not participate in this study if you have a previous injury to your other knee (i.e. meniscal tears, ligament tears).

Protocol

For this study there will be two portions, treadmill walking and force control. This study will assess your maximum force generating stability and walking stability at the biomechanics lab

Participant Initials _____

located at the University of Delaware. For both portions of this study you will be asked to wear shorts, t-shirt, but no footwear. While you are seated we will attach reflective markers to your legs using Velcro straps. We will also attach individual reflective markers to the bony parts of your ankle, knee, and pelvis. We will measure your height and weight. Surface electrodes will be placed on the skin of your lower leg and thigh. All electrodes will be taped to your skin to prevent movement and then plugged into a small (6" x 4" x 3") transmitter box that will be attached to the back of a vest, which you will be wearing, with Velcro. Maximum muscle activity will be collected so we can determine when the muscles are working during the activities.

For the force control portion you will first be asked generate maximum force in four different directions (forward, backward, left, and right) while standing on two force platforms attached to the floor. You will then be instructed to move a cursor using force measured by the force platform in a single direction to a target projected on a screen in front of you. You will perform three trials of the cursor control task, moving the cursor back and forth on the screen in front of you. You will perform the cursor control task in each direction (forward/backward and left/right) for each leg. The movement of your legs and upper body will be recorded using special video cameras which detect the reflective objects attached to your body segments. We may also collect video and take photographs for presentation purposes. You will have at least one minute of rest between each trial. Setup will take approximately 1 hour, the testing session will take approximately 45 minutes, and the total time for your participation for this portion will be no longer than 2 hours.

For the treadmill walking portion you will be asked to walk on a treadmill at your self-selected speed for five trials each lasting 2 minutes. The movement of your legs and upper body will be recorded using special video cameras which detect the reflective objects attached to your body segments. We may also collect video and take photographs for presentation purposes. You will have at least one minute of rest between each trial. Setup will take approximately 1 hour, the testing session will take approximately 30 minutes, and the total time for your participation will be no longer than 2 hours.

In total this study should take no longer than 3-4 hours.

For those subjects who have experienced and ACL injury but has not yet undergone reconstructive surgery, you may be asked to return for an additional collection approximately 6 months after your surgery.

RISKS & BENEFITS

As with any physical activity, risks during walking include dizziness, discomfort in breathing and heart problems. While walking on the treadmill, a handrail will be within reach. An emergency stop is available on the treadmill and at the computer within reach of the subject and experimenter, respectively. You may experience muscle soreness the day following the force

Participant Initials _____

control study similar to vigorous exercise. Additionally, you may experience mild discomfort when removing the reflective markers and surface electrodes adhered to your body. If you have an allergy to any adhesives, please let the researcher know. This may exclude you from the trial.

While there is no direct benefit to the participant, it is hoped that your participation in this project will improve our understanding of knee joint stability after knee injuries and subsequent reconstructive surgery. This project aims to better inform physical therapists in designing effective and optimal treatment protocols.

DATA CONFIDENTIALITY

Personal information about you will remain confidential and will not be released (including any publication) without written consent. You will be de-identified using code numbers. Any file containing a link between your subject number and your personal information will be stored on a password protected computer in an encrypted file. Only the investigators of this study will have access to this link. Data obtained from this study will be recorded on a password protected computer and the data will be archived indefinitely. Any paper research records containing personal information will be stored in a locked filing cabinet. If you consent, video or photography acquired during this study may be used as a part of educational presentation which may be viewed by researchers and professional audiences. We will obscure your face so that your identity will not be revealed during these presentations. Any and all data, research records, and videos/pictures will be stored at the University of Delaware indefinitely. Your research records may be viewed by the University of Delaware Institutional Review Board, but the confidentiality of your records will be protected to the extent permitted by law.

COST & COMPENSATION

There will be no cost to you for this study. However you will be compensated $50 for the force control and treadmill walking motion analysis to compensate for travel expenses and time involved via honoraria. If you wish to no provide your social security number you will be compensated with a $50 gift card for travel expenses and time involved.

INJURY

If you are injured during research procedures, you will be offered first aid at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

PARTICIPATION

Taking part in this research study is entirely voluntary. You do not have to participate in this research. If you choose to take part, you have the right to stop at any time. If you decide not to

Participant Initials _____

participate or if you decide to stop taking part in the research at a later date, there will be no penalty or loss of benefits to which you are otherwise entitled. Your refusal will not influence current or future relationships with the University of Delaware. During the course of this study we may find more information that could be important to you. This may include information that may cause you to change your mind about participating in the study. We will notify you as soon as possible if any new information becomes available.

As a student, if you decide not to take part in this research, your choice will have no effect on your academic status or your grade in the class.

CONTACT INFORMATION

If you have any questions about this study, please contact the Principal Investigator, Amelia S. Lanier at alanier@udel.edu .

If you have any questions or concerns about your rights as a research participant, you may contact the University of Delaware Institutional Review Board at 302-831-2137.

Your signature below indicates that you are voluntarily agreeing to take part in this research study. You have been informed about the study's purpose, procedures, possible risks and benefits. You have been given the opportunity to ask questions about the research and those questions have been answered. You will be given a copy of this consent form to keep.

_____ _____

Signature of Participant Date

_____

Printed Name of Participant

Participant Initials _____

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE: July 8, 2015

TO: Holly Silvers, MPT
FROM: University of Delaware IRB

STUDY TITLE: [617078-3] Biomechanical analysis of the FIFA 11+ injury prevention program

SUBMISSION TYPE: Continuing Review/Progress Report

ACTION: APPROVED
APPROVAL DATE: July 8, 2015
EXPIRATION DATE: July 6, 2016
REVIEW TYPE: Expedited Review

REVIEW CATEGORY: Expedited review category # *(4,6)*

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.

Generated on IRBNet

University of Delaware
Informed Consent Form

Title of Project: *Biomechanical Analysis of the FIFA 11+ Injury Prevention Program*

Principal Investigator (s): Holly J. Silvers, PT

Other Investigators: Amelia Arundale, PT, Ryan Zarzycki, PT, Adam Marmon, PhD, Lynn Snyder-Mackler, PT, ScD, Adam Marmon, PhD

You are being asked to participate in a research study. This form tells you about the study including its purpose, what you will be asked to do if you decide to participate, and any risks and benefits of being in the study. Please read the information below and ask the research team questions about anything we have not made clear before you decide whether to participate. Your participation is voluntary and you can refuse to participate or withdraw at any time without penalty or loss of benefits to which you are otherwise entitled. If you decide to participate, you will be asked to sign this form and a copy will be given to you to keep for your reference.

WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to learn more about the effectiveness of an injury prevention program (The FIFA 11+ program) developed specifically for competitive soccer players. The program has been studied in females and in male NCAA soccer players and has been shown to reduce injury by 52%. Your coach, has agreed to utilize the FIFA 11+ program as your warm-up throughout the course of the season. We now would like to understand how it works by analyzing it biomechanically.

- *The study will be conducted in the NCAA using men's and women's soccer players.*
- *We will analyze your movements in the motion analysis lab (Star Campus, University of Delaware). These movements are soccer specific and will include cutting, pivoting, deceleration, a lateral shuffle, a triple hop and a single leg squat.*
- *Your team will utilize the FIFA 11+ warm-up program twice a week throughout the course of the 2014 fall, 2015 spring and fall seasons.*
- *Your injuries will be tracked throughout the course of the season. This information will be kept confidential and your name and any unique identifying information will never be disclosed.*
- *This study will be utilized towards completion of a PhD dissertation.*

You are being asked to take part in this study because…

- *You are a soccer player competing as a member of an NCAA Men's or Women's soccer team.*

Participant Initials _____

- *Every player on your team will be invited to participate. You have the option not to participate in the study.*
- *The testing will take approximately 2 hours. The testing will occur once during preseason and again at the conclusion of the fall (men) and spring seasons (mend and women) for a total of 4-6 hours. In addition, your team will utilize the FIFA 11+ Program as a dynamic warm-up twice a week. This will replace your traditional warm-up and will require no additional time from you as a player.*

WHAT WILL YOU BE ASKED TO DO?

- *Upon your approval, a member of the research team will schedule you for a session in the motion analysis lab to be tested at your convenience. A follow-up testing will be scheduled at the end of the fall and spring seasons.*
- *During the season, you will perform the FIFA 11+ warm-up program twice a week. This will replace your traditional warm-up program and will require no additional time from you. You can view the program in its' entirety here: http://f-marc.com/11plus/home/*
- *Any injury that you incur during the course of the season will be collected.*
- *The number of completed FIFA 11+ warm-ups throughout the course of the season will also be collected.*
- *The movement testing will take place in the motion analysis laboratory on the Star Campus, University of Delaware. The testing will occur once in August, 2014 (men), once during December, 2014 – February, 2015 (men and women), in May, 2015 (men and women), August, 2015 and December, 2015 (for men and women) . Each testing session will take approximately 2 hours.*

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

- *There are no foreseeable risks from participating in the study. The FIFA 11+ warm-up is a well-researched program and has been shown to decrease the number of soccer related injuries and time loss associated with injury.*
- *There is always a small risk associated with movement analysis or participating in an injury prevention program. However, we expect these risks to be minimal or non-existent.*
- *There are no financial risks associated with your participation. The only foreseeable discomfort may be associated with the time devoted to the actual testing process (2 hours for 3 sessions – one in August, one in December, 2014 – February, 2015, and once in May, 2015)*

WHAT ARE THE POTENTIAL BENEFITS?

- *The potential benefit for participating in the research project is that your risk of soccer related injury may potentially be decreased.*

Participant Initials _____

- *The other potential benefits include a more thorough understanding of how the FIFA 11+ program helps to reduce injury in soccer players.*
- *The future benefits include a better scientific understanding of how the FIFA 11+ program helps to decrease injury by slight changes to movement patterns in competitive soccer players. This knowledge is very important to the medical community that works directly with soccer players.*

HOW WILL CONFIDENTIALITY BE MAINTAINED?

- *Your identity will be protected during the course of the study. You will be assigned a unique identifying code number and your name will not be associated with any data that the research team collects. A list will be created to link the code number to the participant. Only the primary investigator (Holly Silvers) will have access to this and this will be kept in a secured file (password protected).*
- *The data that we collect in the lab and the injury data collected throughout the course of the season will be stored in an excel worksheet and will be password secured. Any data that we collect on paper will be stored in a locked filing cabinet in the BIOMS office on the Star Campus, University of Delaware. The records will be stored for three years.*
- *Only the research team will have access to the video recordings of your movement analysis. This data will be stored on a secured hard drive and will be password protected. Your file will be saved with a unique identifier and will not include your name or date of birth. This data will be stored for seven years and will only be used for research purposes. Recordings will be permanently destroyed at this time.*
- *The data will be reported in a manuscript form and will not include your name, age, date of birth or any unique identifier. The data will be grouped together and analyzed. Upon completion, the data will be submitted to a scientific journal for publication.*
- *There is always a risk that confidentiality may be breached. However, the research team will do everything possible to ensure that this in minimized; including the removal of your name, date of birth, position and jersey number from every file, assigning you with a unique identifying code, keeping all of the files password secured, and keeping any written documentation in a locked-secured cabinet.*

"We will make every effort to keep all research records that identify you confidential to the extent permitted by law. In the event of any publication or presentation resulting from the research, no personally identifiable information will be shared."

Your research records may be viewed by the University of Delaware Institutional Review Board, but the confidentiality of your records will be protected to the extent permitted by law.

- *No other research organization will have access to these records.*

WILL THERE BE ANY COSTS RELATED TO THE RESEARCH?

- *There are no financial costs associated with your participation*

Participant Initials _____

WILL THERE BE ANY COMPENSATION FOR PARTICIPATION?

- *There is no compensation associated with your participation, in accordance with NCAA rules and regulations.*

WHAT IF YOU ARE INJURED DURING YOUR PARTICIPATION IN THE STUDY?

If you are injured during research procedures, you will be offered first aid and medical treatment at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

DO YOU HAVE TO TAKE PART IN THIS STUDY?

Taking part in this research study is entirely voluntary. You do not have to participate in this research. If you choose to take part, you have the right to stop at any time. If you decide not to participate or if you decide to stop taking part in the research at a later date, there will be no penalty or loss of benefits to which you are otherwise entitled. Your refusal will not influence current or future relationships with the institution that you attend or are employed by, or the Women's or Men's soccer program at the NCAA member institution.

As a student-athlete, if you decide not to take part in this research, your choice will have no effect on your academic status or with your status as a player on the Men's soccer team.

- *If you are indeed injured during the course of the study, you have the option to have your medical care provided by the University of Delaware's Athletic Department. You may also see private treatment through your private insurance.*
- *In the event you incur a season ending injury during the course of the season, your participation in the study will end.*

WHO SHOULD YOU CALL IF YOU HAVE QUESTIONS OR CONCERNS?

If you have any questions about this study, please contact the Principal Investigator, Holly Silvers at 310 871-2823 or via email: hollys@udel.edu.

If you have any questions or concerns about your rights as a research participant, you may contact the University of Delaware Institutional Review Board at 302-831-2137.

Participant Initials _____

Your signature below indicates that you are voluntarily agreeing to take part in this research study. You have been informed about the study's purpose, procedures, possible risks and benefits. You have been given the opportunity to ask questions about the research and those questions have been answered. You will be given a copy of this consent form to keep.

_____ _____

Signature of Participant Date

_____

Printed Name of Participant

Participant Initials _____

UNIVERSITY OF DELAWARE
RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	November 19, 2015
TO:	Lynn Snyder-Mackler, PT, ScD, FAPTA
FROM:	University of Delaware IRB
STUDY TITLE:	[535813-8] Reactive Agility System for ACL Rehabilitation
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	Approved for Data Analysis Only
APPROVAL DATE:	November 19, 2015
EXPIRATION DATE:	December 17, 2016
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(8)*

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

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Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.

Generated on IRBNet

University of Delaware
Informed Consent Form

Title of Project: Reactive Agility System for ACL Rehabilitation

Principal Investigator (s): Lynn Snyder-Mackler PT, ScD, FAPTA
David Logerstedt PT, PhD, SCS
Other Investigators: Mathew Failla PT, MSPT, SCS
Zakariya Nawasreh, BPT, MS
Elizabeth Wellsandt PT, DPT
Kathleen White PT, DPT
Amy Arundale PT, DPT, SCS

BACKGROUND

You are being asked to take part in this study because of your recent knee injury. After anterior cruciate ligament (ACL) injuries, patients tend to change the way they walk and run. We have developed a balance training program called perturbation training that can help patients return to sports without surgery. In the past, a physical therapist has done this by hand pushing wooden platforms on wheels to challenge the patient's balance. Currently, perturbation training is not often used in most physical therapy clinics because of the one on one time required for the treating therapist. A company called "Simbex LLC" along with Dr. Snyder-Mackler through a National Institutes of Health Small Business Innovation Research (NIH-SBIR) grant, have now developed a device called "Reactive Agility System". This device provides perturbation training based on computer controls and settings set by the physical therapist. The purpose of this study is to see if the perturbation training on the new device is similar to performing the training by hand.

DESCRIPTION

Twenty-two subjects will be enrolled into this study. You must be between the ages of 14 and 55, and participate in jumping, pivoting, and cutting activities greater than 50 hours per year prior to your injury. You will be not be included if you have other serious lower extremity injuries that may prevent you from safely performing testing procedures.

In this study, you will complete functional testing and motion analysis. After the first testing, you will have 10 perturbation training sessions. After the 1st, 5th, and 10th session, a brief, optional addition perturbation session using motion analysis will be performed. A final testing and motion analysis will be completed within two weeks following completion of the 10th training session.

CLINICAL PROCEDURES

The following procedures, which are also common physical therapy treatment and testing after ACL injury in our clinic, are the clinical tests we are studying.

Parent/Guardian's Initials_____ Subject's Initials_____

Functional testing –About 60 minutes
Functional testing will take place in the Health Sciences Complex building on the STAR campus at the University of Delaware, Newark, DE 19716 and will last about 60 minutes.

Strength Testing
The test will measure the strength of the quadriceps muscle (the muscle on the front of your thigh). You will be seated in a device that resists your kicking motion, and measures how much force your muscle can create. Sticky electrodes will be attached to the front of your thigh, and you will be asked to kick as hard as possible by straightening your knee from a bent position. An electrical stimulus will be activated while you are kicking, to fully contract your muscle. During the electrical stimulus you may feel a cramp in your muscles, like a "Charlie Horse", lasting less than a second. Each test will require a series of practice and recorded contractions. Trials will be repeated (up to a maximum of 4 trials) until a maximum contraction is achieved for both legs.

Unilateral Hop Testing
You will perform 2 trials each of the following single-legged hop tests (Diagram 1) in this order: single hop for distance; triple cross-over hop for distance in which you hop three times in a row, with each hop crossing over a 15 cm wide piece of tape; a triple hop for distance in a straight line; and 6 meter timed-hop. Testing for each hop test will begin on your healthy leg, followed by the injured side. You will be allowed 2 practice trials followed by two scoring trials per leg.

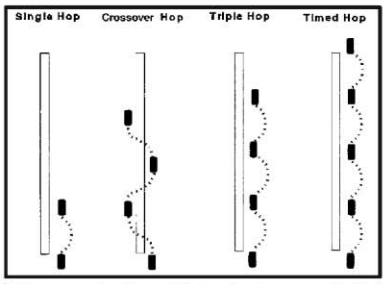

Diagram 1. Four (4) hop tests as part of the functional test protocol.

Self-Report of Function Questionnaires
After hop testing, you will complete a packet of questionnaires. These questionnaires will ask you to rate how well your knee is doing currently compared to your knee prior to your injury.

Motion Analysis Testing – about 2.5 hours
You will be asked to perform motion analysis testing, which will take place in the Motion Analysis Lab in the Health Sciences Complex building on the STAR campus at the University of Delaware. Motion analysis is a set of cameras that will watch your movement. You will come in for optional motion analysis testing after the 1st, 5th, and 10th sessions of perturbation training. In addition, subjects will also ask to come in for motion analysis testing 2 weeks after the completion of perturbation training.

Motion Analysis
We will attach small shiny ball markers on your legs and shoes with tape. Plastic with balls on them will be placed on your lower back, thighs and calves and will be held in

place with elastic wraps. These ball markers will allow the cameras to see your leg positions.

Muscle Activity
Plastic electrodes placed on your skin will be used to record the electrical activity of muscle fibers in your legs. After all electrodes have been placed, you will perform a maximum contraction of each muscle, with straps applied to your ankles to provide resistance. Nine electrodes will be put on each leg with tape and then plugged into a small (6" x 4" x 3") transmitter box that will be attached to the back of a vest with Velcro. The transmitter sends the signal to the computer so we can see when the muscles are contracting during the activities. These measurements will be taken at the same time as the walking and standing during motion analysis testing. The electrodes will be removed at the end of the testing session.

Motion Analysis Activities
There are two parts of the motion analysis testing. Gait or walking analysis will be completed within 2 weeks before the first testing session and within 2 weeks after the 10 treatment sessions. Walking analysis will give us information about the way your hips, knees, and ankles move while you walk. You will be asked to perform 5 trials of walking at a comfortable, self-selected speed, although you may need to walk many times to get 5 good trials for each leg. While you are walking, a computer records the motions of your hips, knees, and ankles. Walking analysis may take up to two hours to complete. The other part of motion analysis will be during perturbation testing, which will be the same as the training sessions but shorter.

Perturbation Training
After completing the First testing, if you agree to be in the study, you will complete 10 sessions of perturbation training using the Reactive Agility System developed by Simbex, LLC (Diagram 2). You will be assigned into one of two different treatment groups, one that has higher level, challenging activities on 3 degrees of freedom platform (3DOF) and one that incorporates higher level, challenging activities on 1 degree of freedom platform (1DOF). Training on the 3DOF platform including single leg balance and standing with one limb on the 3 degrees of freedom plate and one on a fixed surface similar to manual perturbation training currently performed by physical therapists in our clinic. Training on the 1DOF platform will consist of activities involving double and single leg balancing of the plate where the plate can suddenly drop a maximum of 2 inches. Training can be progressed to include higher level activities, such as hopping/jumping-type activities including drop jumps onto the plate. Sessions will be scheduled depending on your time constraints and your ability to improve in rehabilitation.

Diagram 2. Reactive Agility System.

Risks/Discomfort

Subjects with knee injuries could experience a loss of balance during hop testing, however your other leg is free to touch down to provide support and prevent loss of balance. The strength testing can cause muscle soreness and fatigue. Following the testing, your muscles may feel as if you have worked out very hard.

Benefits

The benefits include comprehensive testing sessions and pre-operative physical therapy. You will have no out-of-pocket expenses for the physical therapy. The validation of this device may help improve future treatment after ACL injury and make perturbation training more readily available for clinicians around the world. Medications, medical devices (e.g. braces) and other non-physical therapy expenses are not covered.

Compensation

You will be paid a two-time payment of $50 for participation and completion in this study.

Confidentiality and records

Only the investigators, you and your doctor will have access to the data. All of your data will be de-identified for the purposes of data management and processing. Simbex LLC will also have access to the data from this study but will not be able to access any identifying information. Neither your name nor any identifying information will be used in publication or presentation resulting from this study. A statistical report, which may include slides or photographs which will not identify you, may be disclosed in a scientific paper. Data will be archived indefinitely and may be used for secondary analysis of scientific and clinical questions that arise from this research.

DO YOU HAVE TO PARTICIPATE TAKE PART IN THIS STUDY?

Taking part in this research study is entirely voluntary. You do not have to participate in this research. If you choose to take part, you have the right to stop at any time. If you decide not to participate or if you decide to stop taking part in the research at a later date, there will be no penalty or loss of benefits to which you are otherwise entitled. Your decision to stop participation, or not to participate, will not influence current or future relationships with the University of Delaware

Parent/Guardian's Initials_____ Subject's Initials_____

Videos/photographs

For the purposes of presenting these data in academic papers and presentations, we may also want to use photographs and videos of you performing testing and training procedures. In order for us to use photographs or videos of you, you must agree. Your identity will be hidden. Please circle "Agree" and print, sign, and date if you consent to use of photographs or videos of you, or circle "Disagree" and print, sign, and date if you do not consent to use of photographs or videos of you.

Agree

_____
Subject's Name

_____
Subject's Signature Date

_____
Parent/Guardian's Name (if required)

_____
Parent/Guardian's Signature Date

Disagree

_____
Subject's Name

_____
Subject's Signature Date

_____
Parent/Guardian's Name (if required)

_____
Parent/Guardian's Signature Date

Parent/Guardian's Initials_____ Subject's Initials_____

Study Title: Reactive Agility System for ACL Rehabilitation

Principal Investigator (s): Lynn Snyder-Mackler PT, ScD, FAPTA
David Logerstedt PT, PhD, SCS
Other Investigators: Mathew Failla PT, MSPT, SCS
Zakariya Nawasrah, BPT, MS
Elizabeth Wellsandt PT, DPT
Kathleen White PT, DPT
Amy Arundale PT, DPT, SCS

Subject' Statement:

I have read this consent/assent form and have discussed the procedures described above with a principal investigator. I have been given the opportunity to ask questions regarding this study, and they have been answered to my satisfaction.

If you are injured during research procedures, you will be offered first aid at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

Subject's Consent (for subjects 18 years of age and older)

I have been fully informed of the above described procedures, with its possible risks and benefits, and I hereby consent to the procedures set forth above.

_____
Subject's Name

_____
Subject's Signature Date

_____
Witness Date

Parent/Guardian's Initials_____ Subject's Initials_____

Child's Assent (For subjects under 18 years of age)

I have been fully informed of the above described procedures, with its possible risks and benefits, and I hereby assent to the procedures set forth above. As I am under 18 years of age, I understand that parental or guardian consent is required. My parent or guardian has printed and signed his/her name below.

_____
Subject's Name

_____
Subject's Signature Date

_____
Witness Date

_____
Parent/Guardian's Name (if required)

_____
Parent/Guardian's Signature Date

If you have any questions concerning the rights of individuals who agree to participate in research, you may contact the Chair, Institutional Review Board (302-831-2137). The Institutional Review Board is created for the protection of human subjects involved in research conducted at the University of Delaware.

Further questions regarding this study may be addressed to:
Lynn Snyder-Mackler, PT, ScD, FAPTA
Physical Therapy Department, (302) 831-3613

Parent/Guardian's Initials_____ Subject's Initials_____

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	February 17, 2016
TO:	Lynn Snyder-Mackler, PT, ScD, FAPTA
FROM:	University of Delaware IRB
STUDY TITLE:	[225014-14] Can Neuromuscular Training Alter Movement Patterns? (Renewal Period)
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	February 17, 2016
EXPIRATION DATE:	March 14, 2017
REVIEW TYPE:	Full Committee Review

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Full Committee Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.

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UNIVERSITY OF DELAWARE
DEPARTMENT OF PHYSICAL THERAPY
INFORMED CONSENT FORM

Study Title: Can Neuromuscular Training Alter Movement Patterns? (Renewal Period), Experiment 2 (Aim 3), new randomized controlled trial.

Principal Investigators: Lynn Snyder-Mackler, ScD, PT

Co-investigators: Thomas Buchanan, PhD, Kurt Manal, PhD, Gregory Hicks, PT, PhD, Paul Kolm, PhD, Stephanie Di Stasi, PT, MSPT, PhD, David Logerstedt, PT, MPT, PhD, Andrew Lynch, PT,DPT, Michael J. Axe, M.D., Emily Gardinier, MS, Kathleen White, PT, DPT, Amelia Arundale, PT, DPT, Ryan Zarzycki, PT, DPT, Jacob Capin, PT, DPT, MS,

PURPOSE AND BACKGROUND

You are being asked to participate in a study that will investigate the effect of post-operative rehabilitation on the movement patterns and functional abilities in patients who have had a complete anterior cruciate ligament (ACL) rupture. You have been referred to this study because you have completed a functional screening examination and have decided to undergo ACL reconstruction.

Participation in this research study is voluntary. Your treatment will not be affected by whether or not you participate and you may withdraw from the study at any time without penalty. This program will include treatment activities we currently use in our clinic to treat patients with ACL injury. Your surgeon and physical therapist have agreed that this treatment regimen and all of the testing procedures included in the study are acceptable. Your surgeon has agreed that the tests are being conducted at appropriate intervals following your surgery. In addition to the physical therapy treatment, the study includes strength testing and analysis of your knee movement during walking. There will be a total of four (4) testing sessions: 1) pre-intervention, 2) immediately after completing intervention, 3) 12 months after surgery, and 4) 24 months after surgery. This research study will involve approximately eighty (80) subjects with ACL injury between the ages of 13-55 years. Persons of all sexes, races, and ethnic origins may serve as subjects for this study.

A description of each procedure and the approximate time it takes for each test and the study procedure are outlined below.

Subject's Initials_____

PROCEDURES

ACL Functional Test

Functional testing will take place in the Physical Therapy Clinic at the University of Delaware, 540 S. College Ave, Newark, DE, 19713 and will last approximately 1 hour. Testing will be performed pre-intervention, immediately after intervention and 12 and 24 months after surgery. This is commonly done at the University of Delaware Physical Therapy Clinic as part of the post-operative ACL rehabilitation protocol.

Strength Testing

The test will measure the strength of the quadriceps muscle on the front of your thigh. You will be seated in a dynamometer, a device that resists your kicking motion, and measures how much force your muscle can exert. Self adhesive electrodes will be attached to the front of your thigh, and you will be asked to kick as hard as you can against the arm of the dynamometer. An electrical stimulus will be activated while you are kicking, to fully contract your muscle. During the electrical stimulus you may feel a cramp in your muscles, like a "Charlie Horse", lasting less than a second. Each test will require a series of practice and recorded contractions. Trials will be repeated (up to a maximum of 4 trials) until a maximum contraction is achieved for both legs.

Hop Testing

A series of four (4) single leg hop tests (Diagram 1) will be performed once the swelling in your knee has resolved and you demonstrate good thigh muscle strength. The tests are performed in the order seen in Diagram 1. You are required to wear a standard off-the-shelf knee brace on your injured knee during this portion of the testing.

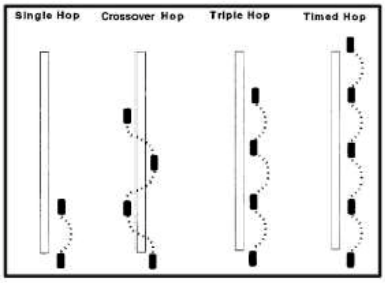

Diagram 1. Four (4) hop tests as part of the functional test protocol.

Two practice trials will precede each of the hop tests before the recorded testing begins. You can put your other leg down at any time to prevent yourself from losing your balance. However, only the two trials in which you are able to 'stick the landing' on one foot will be counted towards your scores. This series of hop tests will be performed on both legs.

Questionnaires

Following strength and hop testing, you will be asked to complete a test packet which includes questions about your injury, past and current functional status, and perceived functional capabilities. If you are unable to complete the strength and hop testing for any reason, you will still be asked to complete the test packet so that we may get as much information about the current status of your knee as possible.

Motion Analysis Testing

All subjects will be asked to perform motion analysis testing, which will take place in the Motion Analysis Laboratory at the University of Delaware, Department of Physical Therapy, 540 S. College Ave, Newark, DE, 19713. Motion analysis testing will take place pre-intervention, immediately after intervention, 12 and 24 months after your ACL surgery.

Motion Analysis

Markers will be affixed to your skin and sneakers on both legs using adhesive skin tape. Shells with markers on them will be placed on your pelvis, thighs and calves and will be held in place with elastic wraps. These markers will allow the cameras to track your leg positions.

Muscle Activity

Electrodes, taped to your skin, will be used to record the electrical activity of your muscles. After all electrodes have been placed, you will perform a maximum contraction of each muscle, with straps applied to your ankles to provide resistance. Nine electrodes will be secured to each leg and then plugged into a small (6" x 4" x 3") transmitter box that will be attached to the back of a vest with Velcro. The transmitter sends the signal to the computer so we can determine when the muscles are contracting during the activities. These measurements will also be taken during the walking trials of the motion analysis testing. The electrodes will be removed at the conclusion of the testing session.

Walking Trials

Immediately following the initial muscle activity testing, you will be asked to perform several walking trials in our laboratory. Walking trials will give us information about the way your hips, knees, and ankles move while you walk. You will be asked to perform 7 trials of walking at a comfortable, self-selected speed, although additional trials may be required to obtain enough data. While you are walking, a computer records the 3 dimensional motions of your hips, knees, and ankles. The entire motion analysis session will last approximately two (2) hours.

Subject's Initials_____

Physical Therapy

Twelve weeks after surgery, and when you have sufficient thigh muscle strength, you will be randomized into one of two different treatment groups, both of which incorporate higher level, progressive activities, including running and agility training. Ten sessions will be scheduled two to five times weekly, depending on your time constraints and your ability to progress with therapy.

Risks/Discomfort

You may experience discomfort from the removal of tape holding markers and EMG electrodes in place. Subjects with ACL injury could experience a loss of balance during testing, however your other leg is free to touch down to provide support and prevent loss of balance. The strength testing can be associated with local muscle soreness and fatigue. Following the testing, your muscles may feel as if you have exercised vigorously.

Benefits

The benefits include comprehensive testing sessions and post-operative physical therapy. All physical therapy sessions will be administered by a licensed physical therapist. The results of this study may help us improve the way we treat patients with ACL injury. Out-of-pocket expenses related to postoperative physical therapy treatment sessions, specifically your co-pay, will be covered by this grant. Medications, medical devices (e.g. braces) and other nonphysical therapy expenses are not covered.

Compensation

You will be paid an honorarium of $100 for the motion analysis testing and functional testing to compensate you for travel expenses and the time involved.

Confidentiality and records

Only the investigators, you and your physician will have access to the data. All of your data will be de-identified for the purposes of data management and processing. Neither your name nor any identifying information will be used in publication or presentation resulting from this study. A statistical report, which may include slides or photographs which will not identify you, may be disclosed in a scientific paper. Data will be archived indefinitely and may be used for secondary analysis of scientific and clinical questions that arise from this research.

Subject's Initials_____

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Subject's Statement:

I have read this consent/assent form and have discussed the procedure described above with a principal investigator. I have been given the opportunity to ask questions regarding this study, and they have been answered to my satisfaction.

If you are injured during research procedures, you will be offered first aid at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

I have been fully informed of the above described procedures, with its possible risks and benefits, and I hereby consent/assent (for those under 18 years of age) to the procedures set forth above.

If I am under 18 years of age, I understand that parental or guardian consent is required. My parent or guardian has printed and signed his/her name below.

_____	_____	_____
Subject's Name	Subject's Signature	Date
_____	_____	_____
Parent/Guardian's Name	Parent/Guardian's Signature	Date
_____		_____
Lynn Snyder-Mackler, Principal Investigator		Date

If you have any questions concerning the rights of individuals who agree to participate in research, you may contact the Institutional Review Board (302-8312137). The Institutional Review Board is created for the protection of human subjects involved in research conducted at the University of Delaware.
Further questions regarding this study may be addressed to:
Lynn Snyder-Mackler, ScD, PT, Physical Therapy Department, (302) 831-3613