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Evaluation of portable accelerometers and force platforms as clinically feasible instrumented outcome measures

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EVALUATION OF PORTABLE ACCELEROMETERS AND FORCE PLATFORMS AS CLINICALLY FEASIBLE INSTRUMENTED OUTCOME MEASURES

by

David Paul Robbins

A thesis submitted in partial fulfillment
of the requirements for the Doctor of Philosophy
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CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D thesis of

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has been approved by the Examining Committee for
the thesis requirement for the Doctor of Philosophy degree
in Physical Rehabilitation Science at the December 2015 graduation.

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To my wife Karen, and children Josh, Hannah, and Sarah for your tireless support, patience, and unconditional love

The hill, though high, I covet to ascend,
The difficulty will not me offend;
For I perceive the way to life lies here.
Come, pluck up heart, let's neither faint nor fear;
Better, though difficult, the right way to go,
Than wrong, though easy, where the end is Woe.

John Bunyan
The Pilgrim's Progress

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ABSTRACT

The use of wearable sensors in consumer health and medicine is a rapidly developing topic of interest. The main purpose of the series of studies in this thesis is to identify novel uses of technology that can provide clinicians and scientists clinically feasible, low cost approaches to obtain meaningful information about functional limb symmetry in patients with knee injuries.

In Study 1, individuals undergoing knee surgery were evaluated as they walked and stepped down onto a force platform in a manner similar to how one would step off a curb to cross a street. When subjects stepped onto their uninvolved leg, peak vertical ground reaction force was greater and occurred earlier than when stepping onto their involved leg. Asymmetries were greater in those with higher quadriceps neuromuscular impairment.

In Study 2, the reliability and validity of using wearable accelerometer sensors was evaluated for estimating single leg vertical hop height in healthy people and individuals after ACL reconstruction surgery. The reliability and concurrent validity of using accelerometers to estimate single leg hop height were excellent, and were similar for healthy and ACL-reconstructed subjects. Error for this method was low, in particular when the accelerometer was worn at the lower leg. Asymmetry in hop height was greater in those with higher quadriceps neuromuscular impairment.

In Study 3, wearable accelerometers were compared to a system of motion capture cameras and force platforms as a method to assess functional movement asymmetry in healthy people and individuals after ACL reconstruction. While walking and stepping down, accelerometers worn at the waist were able to detect underlying movement asymmetry when it exists in people after ACL reconstruction. Acceleration at the waist was strongly associated with vertical ground reaction force and moderately associated with knee extension moments. Collectively, these studies provide evidence

that functional movement symmetry can be measured with simple, inexpensive methods that can be used in a variety of clinical or field-based settings.

PUBLIC ABSTRACT

People move differently after they have recovered from knee surgery. Most equipment that measures the way people move is very expensive and not practical for health care providers or patients. Sensors like fitness bands and activity monitors have become increasingly popular. Small monitors like these may offer a less expensive alternative to expensive lab equipment to measure how people move.

In the first study of this thesis, we measured how hard people stepped down from a raised platform similar to stepping off a curb to cross a street. This movement has not been studied much, but is performed frequently in daily life. People after knee surgery stepped down harder onto their uninjured leg, especially when their quadriceps muscles were weak. In the second study, we used monitors similar to activity monitors to measure how high people could hop off one leg after recovery from knee surgery. When we compared the height from the monitors to the height measured by laboratory equipment, the two measurements strongly agreed. In the third study, we used monitors similar to activity monitors to measure how hard people stepped down from a raised platform. The monitors were able to measure differences between legs that were similar to expensive equipment used in the laboratory. Together, these studies show that we can measure differences in how people move with inexpensive portable monitors. This may be an additional tool can be used by health care providers like physical therapists to identify abnormal movement and help patients.

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CHAPTER 1

INTRODUCTION

Instrumented Outcome Measures in Rehabilitation

Personal monitoring technologies like fitness trackers and mobile applications have exploded in popularity and are driving a revolution in healthcare. Consumer-oriented monitors are designed to measure and motivate habitual behavior that extends beyond formal bouts of exercise. Within healthcare and research, similar technology can provide practitioners, patients, and scientists with a rich and comprehensive set of data not typically gleaned during routine visits to a medical provider. This mobile health (mHealth) information permits objective insight into how patterns of habitual behavior affect patient outcome and prevention of disease. Furthermore, wearable sensors empower patients by providing real-time feedback to the user.

Despite a wide range of potential uses for wearable sensors in healthcare, wearable sensors have been used primarily over a period days to quantify habitual behavior that occurs outside of formal rehabilitation. Accelerometer sensors have gained notoriety for measuring habitual behavior like daily physical activity and patterns of sleep. When used in this fashion, accelerometers are useful in assessing patient outcome, aiding in diagnosis, and providing evidence for therapeutic intervention (http://www.bu.edu/bostonroc, 2015). This highlights the profound benefit of wearable sensors used in mHealth. They provide unique and complementary information to other types of outcomes that are more commonly used in healthcare. Thus, they play a key role in comprehensively understanding the complex nature of patient outcome (http://www.bu.edu/bostonroc, 2015). But, wearable accelerometer sensors also hold exceptional potential to enable clinicians and scientists to understand specific movement biomechanics in ways not currently possible in non-laboratory situations. The inquiry and application of wearable accelerometer sensors for biomechanical monitoring is in its infancy, and is the primary focus of the series of studies in this thesis. The long term goal of this work is to advance wearable accelerometers as detectors of biomechanical events in order to permit insight into movement asymmetry for clinicians and scientists without practical access to a biomechanical laboratory.

The wearable sensors used in mHealth are known as instrumented outcome measures (IOMs). Instrumented outcomes measures are one of three categories of outcomes defined by the Boston Rehabilitation Outcome Center, part of the Medical Rehabilitation Research Infrastructure Network funded by the National Institutes of Health (http://www.bu.edu/bostonroc, 2015). The other two categories of outcomes, Patient Reported Outcomes (PROs) and Performance Based Measures (PBMs), are commonly used in rehabilitation and research and are familiar to most clinicians. Examples of instruments used as IOMs include accelerometer sensors, gyro sensors, global positioning sensors, and heart rate monitors. These sensors are small and unobtrusive to wear. Recent improvements in instrumentation have provided upgrades in the precision, reliability, and measurement range of these sensors. Many sensors also have enough data storage onboard the device to record and save over a month of continuously-sampled data. Electronic platforms for data management have become highly adaptable and user-friendly. Altogether, implementation of IOMs has become feasible in most clinical and research settings.

Chapter 2, which is entitled “Instrumented Outcome Measures Present New Opportunities in Orthopaedic Rehabilitation”, discusses the unique and exciting role of wearable sensing in orthopaedic rehabilitation. The use of wearable sensors to characterize novel aspects of patient outcomes in orthopaedic rehabilitation is in its early stages and offers tremendous potential. Wearable sensors provide objective insight into patterns of daily behavior that are crucial to health and disease prevention. They also hold great promise for use as clinically feasible surrogates for more complicated and costly methods of assessing biomechanical outcome in rehabilitation. Because wearable sensing is novel to many practicing clinicians, this chapter outlines major categories of wearable sensors and provides a review of the current usage of these sensors with emphasis on orthopaedic and sports rehabilitation. These sensors are an important component of mobile health technology that enable health information to be seen by the user and shared with providers. This information is empowering and can motivate patients and provide clinicians and researchers with important data on movement and health status. Chapter 2 is designed to enhance understanding of how wearable sensors can significantly impact rehabilitation practice. It is my goal that the published manuscript resulting from Chapter 2 not only inform readers, but spur broad use of wearable sensors in clinical practice and clinical research. The topics discussed in Chapter 2 are particularly germane to today's healthcare climate where mobile health, ubiquitous healthcare, and wearable sensing are rapidly developing topics of interest.

The series of studies in Chapters 3 through 5 explore novel uses of technology that have the potential to give clinicians and scientists simple, low cost approaches to obtaining meaningful biomechanical information in patients with knee injuries. Chapter 3 investigates the ability of a single force platform to detect altered performance of a task encountered in everyday life after knee surgery. Chapters 4 and 5 investigate the use of wearable sensors containing tri-axial accelerometers as a clinically feasible approach to identifying biomechanical events and characterizing asymmetry when performing common functional tasks after knee surgery. My work expands the use of portable accelerometers beyond typical gathering of habitual physical activity data to a clinically feasible method of collecting data related to movement biomechanics that are normally collected in biomechanics laboratories. I expect these investigations to foster additional research and development directed at producing a translatable measurement system. The end goal is to enhance the ability of practicing physical therapists to quantitatively evaluate movement and use their findings to prescribe therapies or make decisions that improve patients' quality of life.

Arthroscopic Partial Meniscectomy and Anterior Cruciate Ligament Reconstruction

Knee arthroscopy is one of the most common musculoskeletal surgical procedures in the United States, and the number of procedures appears to be increasing (Kim et al., 2011, Cullen et al., 2009). Of the nearly 1 million estimated knee arthroscopies performed annually in the United States, approximately 50% are performed for meniscal tear (Cullen et al., 2009, Kim et al., 2011). This procedure is most prevalent in individuals over the age of 45, affecting 30.9 per 10,000 people between 45 and 64 years of age (Cullen et al., 2009). Arthroscopic partial meniscectomy is associated with a marked increase in the incidence of post-traumatic osteoarthritis (PTOA), even in otherwise healthy younger adults (Englund and Lohmander, 2004, Roos, 2005).

Anterior cruciate ligament (ACL) injury is also common, especially among young, active people. The yearly incidence of ACL injuries has been estimated as high as 36.9 per 10,000 people (Gianotti et al., 2009), and up to 300,000 ACL reconstruction surgeries are performed annually in the United States (Cohen and Sekiya, 2007). Despite surgery and subsequent rehabilitation, overall outcome is not ideal. Individuals with ACL-reconstructed knees show persistent abnormalities in lower extremity biomechanics, are at increased risk for subsequent ACL injuries, achieve low rates of return to pre-injury levels of activity, and have a marked increase in the incidence of post-traumatic osteoarthritis (OA).

Consequences of Knee Surgery on Lower Extremity Function

Lower extremity impairments commonly occur after a wide variety of knee surgeries (Hart et al., 2010a, Roewer et al., 2011). Not surprisingly, large impairments accompany major surgeries that have long recovery times. In ACL reconstruction and total knee arthroplasty, impairments can be severe and long lasting (Roewer et al., 2011, Mizner et al., 2003, Stevens et al., 2003, Petterson et al., 2011). Similar impairments, though less severe, are common after more basic surgeries like uncomplicated knee arthroscopy. Altered movement and muscle weakness often persist long after patients have been discharged from rehabilitation and returned to full activity (Ericsson et al., 2006, Glatthorn et al., 2010, McLeod et al., 2012). Quadriceps atrophy, weakness, and altered voluntary activation are particularly common (McLeod et al., 2012, Ericsson et al., 2006, Glatthorn et al., 2010, Mizner et al., 2003, Gapeyeva et al., 2000, Matthews and St-Pierre, 1996, Stevens et al., 2003, Petterson et al., 2011, Becker et al., 2004, Mizner et al., 2005, Williams et al., 2005). This quadriceps dysfunction leads to alterations in neuromuscular control (Williams et al., 2004) and mechanical adaptations in gait and other functional activities (Yoshida et al., 2012).

Quadriceps muscle strength plays an important role in determining the long term health of the knee. The quadriceps play a central role in absorbing and dissipating forces across the knee joint during the loading response phase of gait and other functional activity (O'Connor and Brandt, 1993, Hurley, 1999). When quadriceps weakness impairs this function, the knee is subjected to increased axial loads during the initial contact and loading response phases of gait (Mikesky et al., 2000, Jefferson et al., 1990). This pattern is associated with the development and progression of knee OA (Bennell et al., 2008, Segal et al., 2009, Segal et al., 2010, Segal et al., 2012). Quadriceps muscle weakness also contributes to biomechanical adaptations after knee surgery (Lewek et al., 2002).

In addition to being able to attenuate less force across the knee joint during weightbearing activity, quadriceps weakness also exposes patients to other biomechanical risk factors for knee OA. For example, the quadriceps offers primary restraint to external knee adduction moments that compress the medial compartment of the knee where OA most commonly occurs (Shelburne et al., 2006, Bennell et al., 2008). When this function is diminished by quadriceps weakness, patients' knees become subjected to high external knee adduction (varus) moments that contribute to the progression of medial compartment knee OA (Sturnieks et al., 2008a, Butler et al., 2009, Sward et al., 2013). Thus, quadriceps muscle weakness is believed to affect the long term health of the knee by affecting mechanics in the sagittal and frontal planes.

Stepping Down During Ongoing Gait

Stepping down during ongoing gait is a common task in daily activity throughout all stages of life. This is performed routinely when stepping off a curb. People who undergo knee surgery typically begin stepping down while walking around two weeks after surgery. Although performed daily, this movement is challenging in that it places a similar demand on the quadriceps muscles as that experienced when running at 3.2 m/s (Houck and Yack, 2003). In ACL-deficient individuals who were young and otherwise healthy, this task was able to discriminate between normal and reduced values for peak knee flexion angle and peak internal knee extension moment that is common for this population (Houck and Yack, 2003). The ubiquity and challenging nature of stepping down while walking make it attractive as an outcome measure for use in a wide range of rehabilitation populations. Few performance-based measures are appropriate for use in patients with varied tolerances for physical activity. Walking and stepping down is challenging enough to differentiate function in young, active patients with isolated joint involvement, but routine enough to be performed with older people.

Few studies in the literature include stepping down while walking as a task of interest. In the majority of studies where subjects perform this movement task, performance is analyzed using 3D motion capture and force platforms (Houck and Yack, 2003, van Dieen et al., 2008, Barbieri et al., 2014, Dundas et al., 2014). I'm aware of only one study in which stepping down while walking was used to characterize lower extremity biomechanics in people with knee pathology (Houck and Yack, 2003). I could find no published studies using this task to characterize patient outcomes and lower extremity biomechanics after knee surgery.

Chapter 3, which is entitled "Walking and Stepping Down: A Simple and Relevant Functional Outcome Measure," presents a study in which stepping down while walking is evaluated as an outcome measure in subjects after knee surgery. This study is novel in three ways. First, the walk and step down test was evaluated in two cohorts of subjects that typically display quadriceps dysfunction and altered gait mechanics. Patients undergoing arthroscopic partial meniscectomy and ACL reconstruction were selected because they represent different sides of the continuum of severity after knee surgery. Second, ground reaction force data recorded during the test were compared with measures of quadriceps function and commonly-reported patient-reported outcomes. This provides appropriate context for this test as a performance based outcome measure and explores the relationship between quadriceps strength and stepping down. Third, performance of the movement tasks was evaluated using ground reaction force data from a force platform alone. When compared with the status quo of instrumented motion analysis, this is a less complex and more widely feasible for use as a method of assessing between-limb asymmetry in clinical settings. This approach reflects a central principle of the series of studies in this thesis, which is to empower clinicians by employing simple and accessible methods to characterize patient outcome during common functional tasks. Chapter 3 establishes the utility of the walk and step down test as a functionally relevant performance based measure for use in knee injury patients.

Specific questions this study sought to answer include: How does the performance of stepping down while walking differ between limbs and across time before and after knee surgery? What are the relationships between ground reaction force during this task and measures of quadriceps function (strength, voluntary activation, and atrophy) and patient-reported outcome before and after knee surgery? How does the strategy for stepping down differ for subjects with good vs. poor quadriceps strength?

Based on preliminary data, I hypothesized:

Hypothesis 3a

Performance of the walk and step down test will differ significantly between limbs and across time for both cohorts of subjects. When stepping onto the uninvolved limb, higher values will occur for peak vertical ground force during loading response and for the impulse of vertical ground reaction force from initial contact through the peak during loading response. Peak vertical ground reaction force during deceleration will occur earlier in stance when stepping onto the uninvolved limb. Differences will exist at all measurement points, but will be most pronounced early after surgery.

Rationale: When compared to level walking, prior research shows more pronounced asymmetry during activities that place a higher mechanical demand on the knee (Kuenze et al., 2013, Thambyah et al., 2004, Hooper et al., 2002, Gao et al., 2012, Hall et al., 2012, Ernst et al., 2000, Rudolph et al., 2001). Stepping down while walking produces knee extension moments similar to running (Houck and Yack, 2003). Because of this, I expect this task to result in asymmetry between limbs after knee surgery when neuromuscular impairments are common. Lowering the center of mass with the trailing leg is an activity performed predominantly using the quadriceps muscles. Subjects with quadriceps dysfunction are expected to be less capable of lowering their center of mass in a controlled fashion. As a result, higher amplitude peak vertical ground reaction forces and earlier timing within stance are likely.

Hypothesis 3b

Quadriceps strength, atrophy, and quadriceps activation of the trailing limb will be moderately to highly associated with and predictive of performance on the walk and step down test ($r < -0.50$). Patient-reported outcomes pertaining to the trailing limb will demonstrate lower association and be less predictive of changes in ground reaction force ($r < -0.25$).

Rationale: After knee surgery, quadriceps weakness contributes to alterations in sagittal plane knee mechanics. People with strong quadriceps demonstrated less asymmetry in internal knee extension moment and peak knee flexion angle while walking and jogging, whereas those with weaker quadriceps appeared similar to low-functioning ACL-deficient subjects (Lewek et al., 2002). In contrast, patient-reported outcomes tend to reach a ceiling effect before quadriceps strength and knee mechanics normalize (Sturnieks et al., 2008a, Mendias et al., 2013). I expect the relationship between patient-reported outcomes and ground reaction force variables to be lower because significant differences in strength and mechanics may not be mirrored by similar differences in subjective knee rating scores.

Accelerometer Sensors in Musculoskeletal Rehabilitation and Biomechanical Research

Physical activity monitors use accelerometers to measure the duration and intensity of movement in daily life. These portable monitors are popular because of their low cost and their ability to monitor weeks of free-living activity (Mathie et al., 2004, Chen et al., 2012). Within rehabilitation, portable monitors containing accelerometers have primarily been used as physical activity monitors to investigate the relationship between objectively-measured physical activity and chronic pain (Verbunt et al., 2001, Bousema et al., 2007, van Weering et al., 2009, Alschuler et al., 2011a, Alschuler et al., 2011b, Bussmann et al., 1998, de Groot et al., 2008b, Murphy et al., 2008, Holsgaard-Larsen and Roos, 2012, Farr et al., 2008, Farr et al., 2010, White et al., 2013, White et al., 2012, White et al., 2014, Prioreschi et al., 2013, Piva et al., 2010, Lee et al., 2012, Kop et al., 2005). Some researchers have used portable monitors containing accelerometers to study physical activity as an outcome measure following therapeutic intervention (Bleakley et al., 2010, Tully et al., 2012, Farr et al., 2010, Ilich et al., 2013, de Groot et al., 2008a, Kuhn et al., 2013, Brandes et al., 2011). Laboratory-based studies have used accelerometers mounted on the lower trunk to describe general parameters of gait such as cadence, step length, and walking speed, and have also recorded specific gait events (Auvinet et al., 2002, Zijlstra and Hof, 2003, Kose et al., 2012, Kobsar et al., 2013). Lower-trunk-mounted accelerometers have also been able to reliably discriminate between normal and altered walking patterns (Senden et al., 2011). But, the accelerometers used in these biomechanical studies have typically been custom made and unavailable commercially.

Wearable sensors containing accelerometers have gained popularity due to their usefulness in recording general physical activity (Mathie et al., 2004, Chen et al., 2012).

Within the last several years, wearable sensor technology has improved to the point that acceleration can now be measured over a full physiological range and sampled at frequencies high enough to capture all human movement. Large amounts of raw data can be stored on small devices that are unobtrusive to wear. Recent advances may allow these monitors to be used to measure distinct biomechanical events in non-laboratory settings. For example, pelvic acceleration measured with wireless accelerometers is strongly associated with vertical ground reaction force during ambulation and other functional tasks (Rowlands and Stiles, 2012). Biomechanical events typically recorded with research-grade accelerometers tethered to a computer may now be measured with less expensive, commercially available devices. Wearable sensors have the potential to enhance clinicians' and researchers' ability to obtain meaningful data outside typical laboratory settings, but this must be verified.

Hop Testing in Orthopaedic Rehabilitation

Vertical hop testing is a popular functional performance based measure for healthy athletes and patients nearing return to full activity after injury. Because most athletes are accustomed to taking off from one leg for maximal height, this test has good ecological validity. The single leg vertical hop possesses high test-retest reliability (Gustavsson et al., 2006) and is one of the most sensitive tests for detecting between-limb differences in performance after ACL reconstruction (Petschnig et al., 1998, Thomee et al., 2012, Gustavsson et al., 2006). Large differences in internal knee extension moment are observed in the take-off phase of single leg hopping in the involved limbs of ACL reconstructed people (Ernst et al., 2000).

Several groups of researchers have advocated for using hop testing to help quantify patient outcome and determine readiness to return to full activity after injury (Gustavsson et al., 2006, Noyes et al., 1991, Fitzgerald et al., 2000, Grindem et al., 2011, Logerstedt et al., 2012). These groups advocate for a cluster of single leg hop tests, and require the participant to score over 85-90% of the uninvolved limb in order to return to full sports participation (Gustavsson et al., 2006, Fitzgerald et al., 2001, Di Stasi et al., 2013). Most hop tests used by these groups involve horizontal hopping for distance. Correct performance requires that the participant execute a clean landing onto the same limb taken off from (Gustavsson et al., 2006, Noyes et al., 1991). This requires the subject to not only possess adequate strength and power to hop far, but also adequate confidence and neuromotor control to land cleanly. Hops for distance that require clean landings measure a combination of at least three constructs: lower extremity power required while taking off, confidence in the involved limb while landing, and adequate neuromuscular control to land cleanly. Undoubtedly, these are all important considerations in returning injured athletes to their desired sporting activity. But, if a clinician or researcher wishes to quantify lower extremity power, then requiring a clean landing may not be necessary.

Because they are novel tasks, hop testing for distance and hop testing requiring repeated hops may also lack functional relevance for many people. This is especially true for tests that require consecutive hops with or without a cutting component, such as the triple hop for distance, the triple cross-over hop, the 6 meter timed hop (Noyes et al., 1991), and the side hop (Ageberg et al., 2008). Thus, the vertical hop has greater ecological validity when compared to horizontal hopping, especially when consecutive hops are required when hopping for distance or time.

Many methods exist for scoring hop height. Force platforms with or without a system of motion capture cameras are considered to be the gold standard, but are typically found within biomechanical laboratories (Leard et al., 2007, Castagna et al., 2013, Casartelli et al., 2010). Although highly reliable and accurate, these systems have limited use in clinical or field-based measurement because of high cost and lack of portability. Several field-based methods exist for calculating jump or hop height. The jump and reach method requires that subjects reach as high overhead as possible during flight and manually move as many swiveling plastic slats as possible. The Vertec (Sports Imports, Hilliard OH) is a device commonly used in this measurement method. The jump and reach requires that the participant be at the apex of flight while directly underneath the stack of plastic slats. Because its slats are at half inch intervals, this method lacks precision for detecting between-limb differences in height when compared to other methods. This is especially true for detecting differences in single limb hop height, where the height for most individuals is less than 10 inches (Gustavsson et al., 2006, Ageberg et al., 2008, de Fontenay et al., 2014). The jump and reach method used with the Vertec also lacks accuracy compared to other methods like contact mats and force platforms (Leard et al., 2007).

Contact mats like the Just Jump System (Probotics, Huntsville, AL) score hop height via force or pressure sensors that measure the time of flight and then calculate the hop height based on a simple equation. Similarly, optical timing systems like the Optojump (Microgate, Bolzano, Italy) use the interruption of beams from photoelectric cells placed just off the ground to estimate flight times. Contact mats and photoelectric cells compare favorably against force platforms (Leard et al., 2007, Castagna et al., 2013, Glatthorn et al., 2011). Although contact mats and photoelectric cells also may be used to time events, these devices have few other functions. In contrast, wearable accelerometers have the capability of providing meaningful measurements of habitual physical activity and movement asymmetry, and also require little space.

Prior research investigated the use of portable accelerometers mounted at the shank or trunk to determine flight time and calculate jump height from a double-leg take-off and landing. Some studies report moderate to high agreement between jump height measured by accelerometers and force platforms (Picerno et al., 2011, Castagna et al., 2013, Choukou et al., 2014). But, studies also reported systematic bias between accelerometer-based methods and criterion standards, with accelerometers generally over-estimating hop height (Casartelli et al., 2010, Choukou et al., 2014). Other studies report very high correlation between jump heights calculated from accelerometers mounted on the shank and force platforms (Elvin et al., 2007, Palma, 2008). Because several of these studies do not report absolute agreement and rely exclusively on simple correlation for their conclusions, systematic error between methods is unclear (Elvin et al., 2007, Quagliarella et al., 2010). The methods of scoring reported in these studies are variable and in some cases not well defined, especially for determining the moment of take-off (Picerno et al., 2011, Castagna et al., 2013). In addition, subjects were frequently restricted from using an arm swing during the vertical jump (Castagna et al., 2013, Picerno et al., 2011, Choukou et al., 2014). This practice is utilized to simplify the acceleration signal and make data processing easier, but it also decreases the functional relevance of such jumps and may limit subjects' maximal effort by introducing a novel movement strategy. Research that scores single limb hop height from accelerometers in subjects after injury is needed.

Chapter 4, which is entitled "Measurement of Single Leg Vertical Hop Height by Wearable Accelerometer Sensors", investigated the ability of wearable accelerometers worn at the waist and shank to judge flight time and calculate hop height. This study is novel in two primary ways. First, it is the first study to investigate the ability of accelerometers to judge flight time and calculate height from a single-leg hop rather than a double-leg jump. This is noteworthy because the amplitude of height from hopping off one leg is significantly less than that from jumping off two legs. Measurement error relative to single leg hop height has the potential to be comparatively large versus the measurement error relative to double legged jumping. Second, this is the first study to investigate the use of accelerometers to estimate hop height in a cohort of subjects after knee surgery with expected neuromuscular impairments. It is unknown if criteria for scoring flight time relate equally well to injured subjects as for healthy subjects. Because this study included cohorts of healthy and injured subjects, the results of this study permit insight into this question.

Specific questions this study sought to answer include: Is a simple method of detecting takeoff and landing from accelerometers worn at the waist or shank reliable and valid when compared to the criterion standard from a force platform? Are there differences between healthy subjects and subjects after ACL-reconstruction when using accelerometers to score hop height? What is the magnitude of error associated with using accelerometers to estimate hop height, and what are the clinical implications of this error? How strong are the relationships between hop height calculated with accelerometers, tests of neuromuscular function, and patient-reported outcomes?

Based on preliminary data, I hypothesized:

Hypothesis 4a

Determining hop height from flight time measured by wearable accelerometers mounted at the waist or shank will be highly reliable and valid in healthy subjects and subjects after ACL reconstruction. Intraclass Correlation Coefficient values for intra-rater reliability, inter-rater reliability, and concurrent validity will exceed 0.80. Waist and shank locations will demonstrate similar reliability and validity. No differences in reliability or validity will exist between injured and healthy subjects.

Rationale: Results from analysis of preliminary data suggest high intra-rater reliability, inter-rater reliability, and concurrent validity using accelerometers at the waist and shank to estimate jump height. In addition, previous research showed that using accelerometers to measure jump height via flight time is highly repeatable in healthy subjects (Casartelli et al., 2010, Choukou et al., 2014). Previous research also showed that accelerometers can measure jump height effectively at the waist or shank in healthy subjects (Picerno et al., 2011, Casartelli et al., 2010, Castagna et al., 2013, Elvin et al., 2007, Quagliarella et al., 2010). Although the amplitude of acceleration signal has the potential to be different during take-off and/or landing, the basic shape of the signal with relationship to take-off and landing should remain similar in healthy and injured limbs and should not affect the reliability of this method.

Hypothesis 4b

Systematic error will be less than 2 cm for both accelerometer-based methods of estimating hop height when compared to the criterion standard of a force platform. Bland-Altman 95% limits of agreement, one estimate of random error, will be approximately 6 cm.

Rationale: Systematic errors for estimating jump height from accelerometers demonstrate in healthy subjects considerable variability, with values ranging from approximately 2 cm to 6 cm (Picerno et al., 2011, Choukou et al., 2014, Castagna et al., 2013, Casartelli et al., 2010). However, values at the upper end of this range were based on a scoring algorithm that is prone to systematic error (Casartelli et al., 2010). Systematic error from this study should fall at the low end of this range due to a scoring algorithm that more precisely estimates the moments of take-off and landing. Random error may be expressed by several methods, but the most common expression in comparable studies as 95% limits of agreement on Bland-Altman plots. Previous research for estimating jump height from accelerometers in healthy subjects demonstrates 95% limits of agreement that generally range between 6 cm and 11 cm

(Casartelli et al., 2010, Castagna et al., 2013, Picerno et al., 2011). I expect that values from this study would fall toward the lower end of this range.

Hypothesis 4c

Associations between single leg vertical hop height and quadriceps performance will exceed 0.50. Associations between asymmetry in single leg vertical hop height and patient-reported outcomes will exceed 0.40.

Rationale: Associations between peak quadriceps strength and hop performance have been reported at 0.51 for vertical hopping (Petschnig et al., 1998) and 0.41 to 0.62 for the hop for distance (Wilk et al., 1994). Other research reported that symmetry in quadriceps strength was a significant predictor in single leg hop performance (Schmitt et al., 2012). Associations between patient-reported outcomes and hop performance have ranged between 0.40 and 0.50 (Ageberg et al., 2008, Wilk et al., 1994, Ra et al., 2014). Relationships between patient-reported outcomes and hop performance are typically lower than for neuromuscular performance and hop performance. Patient-reported outcomes frequently demonstrate a ceiling effect when compared to performance-based measures like tests of maximal neuromuscular function and physical performance (Mendias et al., 2013, Ra et al., 2014). Thus, I expect the relationship between two performance-based measures (hop height and peak quadriceps strength) to be stronger than the relationship between hop height and patient-reported outcome.

Alterations in Biomechanics after Anterior Cruciate Ligament Reconstruction

Mechanical adaptations after ACL reconstruction during gait and other weight bearing activities are largest in the sagittal plane, where a “stiffening” strategy of the knee is frequently observed. Hallmarks of this strategy include reductions in peak internal knee extension moment, range of sagittal plane knee excursion, and peak knee flexion angle during the loading response phase of gait (Berchuck et al., 1990, Noyes et al., 1992, Wexler et al., 1998, DeVita et al., 1998, Webster et al., 2005, Hurd and Snyder-Mackler, 2007, Hall et al., 2012, Sturnieks et al., 2008a). Alterations in gait mechanics are common during level walking where the demand on the lower extremity is low. In activities where the demand on the lower extremity is greater (e.g., running, jumping, stair ascent and descent), mechanical alterations are greater (Rudolph et al., 2001, Kuenze et al., 2013, Thambyah et al., 2004, Hooper et al., 2002, Gao et al., 2012, Hall et al., 2012, Ernst et al., 2000).

These altered loading patterns are believed to play a role in the high rates of post-traumatic osteoarthritis (PTOA) and second ACL tears in people who undergo ACL reconstruction (Oiestad et al., 2010, Holm et al., 2010, Keays et al., 2010, Pinczewski et al., 2007, Hewett et al., 2006, Paterno et al., 2012, Liikavainio et al., 2007). Despite the implications of asymmetry in sagittal plane mechanics caused by the stiffening pattern at the knee, it remains prohibitive for clinicians and researchers in non-laboratory settings to evaluate for these impairments. Non-instrumented clinical gait assessment fails to detect the magnitude of asymmetry present. Even the most stringent patient-reported and performance-based criteria for return to sport are incapable of detecting clinically-meaningful differences in knee extension moment between limbs (Di Stasi et al., 2013). An inexpensive, clinically-feasible method of detecting significant and clinically-meaningful differences in sagittal plane mechanics would empower clinicians and researchers in non-laboratory settings.

Chapter 5, which is entitled “Using Portable Accelerometers to Evaluate Lower Extremity Mechanics after ACL Reconstruction”, assessed the ability of wearable accelerometers to identify asymmetry in sagittal plane knee mechanics in subjects who have undergone ACL reconstruction surgery. Chapter 3 established the walk and step down test as a valid and functionally relevant outcome measure. Chapter 5 built upon these findings and determined the ability of wearable accelerometers to evaluate for functional movement asymmetry between limbs.

Chapter 5 used a cross-sectional design intended to appreciate alterations in sagittal plane knee mechanics during and after rehabilitation from ACL reconstruction. Data from wearable accelerometers mounted on subjects' waists and shanks will be compared with data from instrumented gait analysis via motion capture. This will determine the feasibility of using wireless accelerometer sensors to evaluate for adaptations in sagittal plane knee mechanics. This study will be the first to apply a clinically-feasible method to obtain a biomechanical signature of altered sagittal plane mechanics during gait and a highly relevant daily activity. This research is a critical step toward empowering clinicians to evaluate adaptations in sagittal plane mechanics during a routine clinical visit without expensive equipment. This study will enable scientists performing large-scale and-or multicenter longitudinal studies to quickly collect biomechanical data; detailed motion capture studies are typically impractical in these types of projects. Large scale studies are important to understanding the degree to which asymmetrical movement patterns contribute to subsequent knee injury and chronic joint disease.

The specific questions this study sought to answer are: Do differences in acceleration exist between the involved limb, contralateral limb, and healthy control subjects during level walking and walk and step down tasks? What are the relationships between pelvic acceleration and expected alterations in biomechanics during the loading response phase of gait?

Based on preliminary data, I hypothesized:

Hypothesis 5a

Significant differences in acceleration at the waist and shank will exist between the ACL-reconstructed limb, the uninvolved limb, and control subjects during level walking and stepping down after ACL reconstruction. These differences will be more pronounced with stepping down compared to level walking.

Rationale: Based on expected abnormalities in sagittal plane mechanics during gait and functional activity after ACL reconstruction, alterations in vertical acceleration at the pelvis are probable. The results from Chapter 3 confirm that subjects step down harder onto the uninvolved limb when compared to the involved limb, leading to significant increases in ground reaction force amplitude and loading rate. In addition, decreased peak internal knee extension moment, sagittal plane knee excursion, and peak knee flexion angle during the loading response phase of gait are characteristic of a stiffening pattern in the sagittal plane. This leads to larger vertical ground reaction force and a higher rate of loading with less attenuation at the knee (Paterno et al., 2007, Myer et al., 2012). Higher forces are passed to more proximal body segments (Ernst et al., 2000, Di Stasi et al., 2013). Higher vertical accelerations at the pelvis are likely. Deviations from normal mechanics are likely to be magnified under conditions of greater demand like stepping down from a raised platform (Houck and Yack, 2003). Thus, I expect acceleration measured at the waist to discriminate between normal and altered mechanics in these tasks.

Hypothesis 5b

Acceleration at the shank will be strongly associated with peak vertical ground reaction force of the leading leg ($r > 0.50$) and peak knee extension moment of the trailing leg ($r < -0.50$). Acceleration at the waist will be moderately associated with peak vertical ground reaction force ($r > 0.40$) and internal knee extension moment of the leading leg (r < -0.40). Acceleration measured at the waist will be best predicted by regression analysis that incorporates strategies from both the trailing and leading limbs. Rationale: Two different adaptations contribute to asymmetries in acceleration for subjects after ACL reconstruction. The first adaptation involves the role of the trailing limb as it lowers the body's mass during late stance when stepping down. When the involved leg is trailing I expect this leg to be less effective at lowering the body's mass during stance. As a direct result of this, larger vertical ground reaction force and higher acceleration measured at the shank of the uninvolved leg will follow. This mechanism will also provide bias for higher acceleration measured at the uninvolved side of the waist when stepping down.

The second adaptation involves the role of the leading leg during loading response. When stepping onto the involved leg, I expect lower knee extension moments and less sagittal plane knee motion (Houck and Yack, 2003). When compared to normal biomechanics, this "stiffening strategy" should result in higher acceleration at the waist when walking on level ground. When stepping down, adaptations of the leading leg compete with those from the trailing leg. These competing influences of the leading and trailing limbs are expected to result in lower Pearson Product Moment correlation coefficients between knee extension moments and acceleration measured at the waist than between knee extension moments and acceleration measured at the shank. Acceleration measured at the waist is expected to be better predicted by regression analyses incorporating the strategies of both the leading and trailing limbs.

Significance

The goal of the series of studies in this thesis was to investigate simple methods of obtaining meaningful biomechanical information in people after knee surgery. This work was designed to spur further research directed at enabling clinicians in a broad range of settings to better evaluate movement, collect clinical outcomes data, and participate in clinical research. The applications of wearable sensors under investigation in this thesis are not intended to replace biomechanical studies using 3D motion capture systems. More detailed biomechanical analysis should continue to be used in most research and whenever feasible. Rather, the aim was to enable clinicians to better evaluate movement and perform clinical research in settings and designs where typical biomechanical analyses are impractical. The use of wearable sensors in rehabilitation is in its infancy. This research has the potential to help advance mobile health and comprehensive outcome measurement in orthopaedic rehabilitation. These enabling studies were intended to be a “stepping stone” in a line of research directed at helping physical therapists better perform their hallmark skill — evaluating and treating abnormal human movement.

CHAPTER 2

INSTRUMENTED OUTCOME MEASURES PRESENT NEW OPPORTUNITIES IN ORTHOPAEDIC REHABILITATION

Introduction

Despite the fact that the majority of patients' time is spent away from formal rehabilitation for musculoskeletal conditions, the effect of this behavior on patient outcome is largely unknown. But, habitual behavior may prove equally important to patient outcome as formal rehabilitation itself. Research suggests that habitual, unstructured physical activity throughout the day may be equally important for the prevention of obesity and chronic disease as formal exercise (Kim et al., 2013, White et al., 2015, Strath et al., 2008, Clarke and Janssen, 2014). Personal monitoring technology like mobile applications ("apps") and inexpensive fitness trackers have exploded in popularity. These are designed to measure and motivate habitual behaviors that extend beyond formal bouts of exercise. Within healthcare, similar technology may be used to provide a rich and comprehensive data set that permits insight into patterns of habitual behavior and empowers patients by providing real-time feedback to the user. Mobile health (mHealth) technology is a key element able to make health information available to patients and clinicians regardless of time or location, a concept known as ubiquitous health (Rodgers, 2014, Steinhubl et al., 2015). Instrumented outcome measures (IOMs) quantity habitual behavior that primarily occurs outside of formal rehabilitation. By virtue of placement and role within the healthcare system, physical therapists are optimally positioned to exploit this technology to advance rehabilitation and demonstrate leadership in a changing healthcare landscape.

As the technology underlying consumer-oriented devices has improved and the cost decreased, similar trends are evident for wearable sensors designed for health and rehabilitation. These sensors have become unobtrusive to wear and the instrumentation has improved in precision and reliability. Memory storage within these devices has become sufficiently large to store weeks of continuously-sampled data, and electronic platforms for data management have become highly adaptable and user-friendly. The gap between wearable sensors designed for research and those designed for everyday consumer use has narrowed, and will likely continue to narrow in coming years. Altogether, implementation of IOMs has become feasible in most clinical and research settings, and will likely become even easier in the near future.

As part of the Medical Rehabilitation Research Infrastructure Network, the National Institutes of Health funds the Boston Rehabilitation Outcome Center (ROC) to serve as a center of excellence and provide consultation support for rehabilitation professionals and scientists wishing to improve and better understand outcome measurement. For rehabilitation practice and research, Boston ROC characterizes outcomes in three primary categories: Patient/Clinician Reported Outcomes (PROs), Performance Based Measures (PBMs), and Instrumented Outcome Measures.(http://www.bu.edu/bostonroc, 2015). For the purpose of this paper, we define IOMs as methods that use wearable sensors to provide objective assessment of mechanical or physiological events. These are frequently used over an extended period of time and may be beneficial in assessing patient outcome, aiding in diagnosis, and providing evidence for therapeutic intervention (http://www.bu.edu/bostonroc, 2015). IOMs are particularly useful in gathering data during free living and thus are highly ecologically valid.

IOMs provide unique and complementary information to PROs and PBMs and therefore play a role in fully understanding a wide variety of patient outcomes. While PROs rely on patient self-report or clinician judgment, IOMs rely exclusively on objective data gathered from sensors. And while PBMs measure performance on discrete, standardized tasks, data from IOMs are derived from free-living, ecologically valid situations over a comparatively long period of time. Thus, IOMs objectively record habitual behavior as opposed to PBMs that measure optimal physical capacity.

To provide an example of how each type of outcome may complement the others and contribute to an overall construct, consider how a study by van Weering et al. (van Weering et al., 2011) measured physical activity by three methods in a sample of patients with chronic low back pain. The authors used the Baecke Physical Activity Questionnaire to capture self-reported physical activity. A PRO like this is helpful to gain insight into the effect of perceived disability on physical activity. But, PROs are open to recall bias and confounding influence from psychosocial variables and thus may not accurately reflect patients' true physical activity levels (van Weering et al., 2011, Verbunt et al., 2005, Huijnen et al., 2010, Sabia et al., 2014, Schuna et al., 2013). The authors used the timed up and go test as a PBM to assess subjects' physical capability of movement in a standardized fashion. But, a PBM like this does not offer insight into how physical activity is affected by disability nor does it provide insight into the activity in which subjects participate. Finally, the authors used a wireless, accelerometer-based physical activity monitor to measure physical activity continuously over the period of 5 days. This IOM provided a record of the frequency, intensity, and duration of movement and therefore served as an objective measure of mobility. Each class of outcome provided unique and complementary insight into characteristics of patients' physical activity. Together, all three types of outcomes provided a comprehensive description of physical activity and mobility.

The purpose of this commentary is to provide a current perspective on the use of instrumented outcome measures in orthopaedic and sports rehabilitation. This commentary provides a brief familiarization with major categories of IOMs, reviews applications of wearable sensors in orthopaedics and sports medicine, and reviews use of IOMs in musculoskeletal rehabilitation. In light of rapidly expanding technology in healthcare, this commentary offers suggestions for how IOMs may be integrated into routine clinical practice and research in this era of rapidly expanding mHealth technology.

Types of Sensors

A wide variety of wearable sensors is available and may be classified according to their primary purpose and typical use (Table 2.1). These sensors are available as single monitors. But, many wearable monitors are now multi-dimensional and contain a variety of different sensors onboard one wearable monitor. This allows monitors traditionally associated with only one purpose to be used for many purposes. For example, many “physical activity monitors” now include not only accelerometers aligned with the three cardinal planes of the device, but also include triaxial gyro sensors, inclinometers, and magnetometers capable of sensing changes in orientation, and luminance meters that quantify the lux level of ambient light. In addition, many monitors are capable of integrating information from separate sensors (e.g., heart rate and GPS) into a single data file. Systems for IOM integration exist that contain similar on-board elements to activity monitors (e.g., accelerometers, light sensors, magnetometers, gyro sensors), but also are capable of incorporating portable systems for physiologic measurement such as electrocardiography and electromyography. These devices, such as those manufactured by Shimmer (Dublin, Ireland) and Cambridge Neurotechnology (Cambridge, UK) are devised to serve as modular platforms for collecting IOM data from multiple types of sensors.

Sensor Type	Primary Purpose	Measurement Units	Typical Use
Accelerometer	Detect intensity and frequency of movement via acceleration	Gravitational equivalent (g)	Monitoring physical activity and sleep
Gyro sensor	Sense orientation via change in angular momentum	Angular velocity (^{0}/sec)	Postural orientation. Increase 3-D movement recognition when used with accelerometers
Magnetometer	Detect change in orientation relative to the earth's magnetic field	Strength of magnetic field (Tesla)	Increase 3-D movement recognition when used with accelerometers and gyrosensors
Inclinometer	Sense inclination relative to the surface of the earth	Angle (^{0})	Postural orientation
Global Positioning Sensor (GPS)	Collect 3-D positioning and time information via satellite signal	Geographic Coordinate System	Routes of travel, distance covered, speed of movement; community integration
Luminance (lux) Meter	Detect ambient light	Candela per square meter (cd/m^2)	Determine indoor vs. outdoor activity; may be useful for community integration
Heart Rate	Detect rate and relative timing of heart beat via the heart's electrical activity	Beats/min	Intensity of physical activity Heart rate variability measures autonomic nervous system regulation
Force/Pressure Sensor	Detect foot pressure, force, and timing pattern via sensors worn within the shoe	Force: Newtons (N) Pressure: Pascale (Pa)	Foot pressure, force, and timing during gait
Skin Conductance Response/ Skin Galvanic Response	Measures electrical conductance on skin due to changing electrolyte content in sweat	Electrical resistance [Ohm (Ω)]	Determine emotional and physical stress due to sympathetic nervous activity

Table 2.1: Types of sensors and their primary use in mobile health

The ability to integrate data from multiple sensors has the potential to create a rich set of data able to comprehensively characterize client outcome. Several software options exist for data management with IOMs. Most consumer-oriented devices synchronize with personal mobile devices via mobile application, and typically also have software programs capable of downloading data for longer-term storage. Some monitors designed for research purposes require proprietary software to download data. These software packages typically offer user-friendly methods for data processing. While these software applications may offer several options for data management, they may require an annual licensing fee to provide updates and support. Similar to consumer-oriented devices, many monitors designed for research also can communicate with personal mobile devices via mobile application. Finally, many researchers who prefer the option of customizing data processing will write custom software that is able to use raw data files downloaded from the monitor.

Application of Wearable Sensors

Physical Activity Monitoring

The use of accelerometers by public health professionals and epidemiologists to objectively monitor physical activity has increased exponentially in the last 10 years (Troiano et al., 2014). For example, the Centers for Disease Control use accelerometer-based physical activity monitors in the National Health and Nutrition Examination Survey, a large program of longitudinal studies (Troiano et al., 2008). These accelerometers are designed for extended wear and are small, wireless, and unobtrusive in order to maximize wear compliance. To allow for several weeks of monitoring without recharging, long battery life and large data storage capacity are important considerations for these devices. On-board data storage capacity is common up to 4 GB, which is enough to store over a month of continuously sampled data at sampling frequencies of up to 100 Hz. If a physical activity monitor will be used to provide feedback and motivation to the wearer in a similar fashion as a consumer-oriented fitness monitor, the device should be able to transmit data to a mobile device and/or contain a display that offers basic data for real-time feedback (e.g., number of steps taken). Because many activity monitors contain additional sensors beyond accelerometers, care should be taken to select a monitor that matches all its intended purposes. Finally, if planning to detect high intensity activity, the activity monitor should possess an acceleration range of measurement wide enough to capture most running and jumping activities. Most devices today measure at least ± 8 g, with some commercially available that measure up to ± 16 g. A range of ± 8g should be adequate to measure most running activities at the waist. But, if mounted on the shank while running or jumping, an accelerometer of ± 20g may be necessary (Mathie et al., 2004, Mizrahi et al., 2000).

Physical activity monitors may be worn at a variety of locations on the body. Classically, monitors are worn around at the waist and secured with a belt or on snug-fitting clothing (Troiano et al., 2008). The advantage of mounting on the trunk is that the accelerometer measures whole body accelerations due primarily to impact from the feet striking the ground. Thus, trunk-mounted accelerometry is well-suited to capture accelerations related to the impact of ambulation and weight bearing activity. Most criteria to estimate activity intensity were developed with activity monitors mounted on the waist, therefore requiring wear at this site if such information is to be used reliably.

Wearing a monitor on the wrist is believed to increase subjects' wear compliance (Rosenberger et al., 2013), and also captures gesturing with the upper extremity. Research has generally shown high measures of association between energy expenditure and acceleration measured at the waist or the wrist (Zhang et al., 2012, Parkka et al., 2007), and physical activity data for NHANES are recorded from the wrist. But, systematic differences exist between wearing an accelerometer sensor on the wrist vs. the waist (Tudor-Locke et al., 2014, Hildebrand et al., 2014), so consistency in measurement is needed to permit comparison across time or between subjects. Accelerometers mounted at the waist may also have greater association with energy expenditure than accelerometers worn at the wrist (Rosenberger et al., 2013). Research is under way using advanced methods of data analysis to use physical activity monitors to identify specific activities like walking, household chores, and other daily activities (Trost et al., 2012, Trost et al., 2014, Mannini and Sabatini, 2010, Long et al., 2009). If physical activity monitors are used for this purpose, then it is essential that the wear site matches that of the algorithm used for identification.

Physical activity monitors may provide data in terms of a discrete number of steps taken (similar to a pedometer), as “counts”, or as raw acceleration data. Counts are summaries of physical activity composed of arbitrary units and are often proprietary (Chen and Bassett, 2005). Count data is derived by a series of data processing steps whereby raw acceleration data is divided into selectable epochs of time (generally ranging from 1 to 60 seconds), then processed to create a smooth “curve” that represents average physical activity intensity for a specific population over a course of time. The area under this curve is then expressed in units of “counts” (Chen and Bassett, 2005). Thus for each epoch of time, an activity monitor would provide a value that estimated the total volume of physical activity for that time period. In many cases counts differ substantially between brands and models of activity monitors, thus making comparison difficult (Ward et al., 2005). Count data from activity monitors worn primarily at the waist have been compared with measures of energy expenditure to derive “cut points”, or stratified criteria to define the intensity of physical activity (Rosenberger et al., 2013, Matthew, 2005). Thus, using count data from physical activity monitors may serve as an estimate for energy expenditure in specific populations.

Count data and cut-points have recently fallen out of favor in physical activity research due to several factors. The multiple steps of data processing needed to obtain average acceleration over an epoch of time leads to a loss of resolution for specific, non-repetitive bouts of acceleration experienced, especially for longer epochs (Chen and Bassett, 2005, Chen et al., 2012). In addition, most estimates for cut points based on count data were derived from wear at the waist, making analysis of count data obtained from non-traditional sites of wear problematic. Furthermore, estimates given by count data are not available for many populations, including individuals with impairments from musculoskeletal conditions. Many researchers now prefer to evaluate physical activity based on the raw acceleration signal from activity monitors, although the clinical feasibility of this is limited at this time due to the time needed to process raw data. Ideally, software applications will be developed that permit clinicians to rapidly evaluate acceleration from activity monitors based on raw acceleration rather than cut points.

A number of software user-interface systems allow for access to both raw and processed data, thereby allowing the researcher to select their approach to data processing. For example, summarizing physical activity intensity in epochs of time or discrete number of steps could be advantageous if data storage capacity is limited or if researchers have little familiarity with raw acceleration signal. In addition, count data may be desirable if planning to estimate the amount of time spent in various intensity levels of activity based on previously developed cut points, permitting that the subject population and wear site match those of the original research. Because simplified data is easier for consumers to understand, fitness trackers and activity monitor software typically provide feedback to the user expressed either in terms of counts or discrete number of steps per day. This information can be helpful in motivating behavior and enabling clients to obtain recommended levels of activity.

Among subjects with musculoskeletal pain conditions, accelerometer-based activity monitoring has been studied primarily in subjects with chronic pain. Many investigators in this area have reasoned that because general activity and exercise are commonly-prescribed treatments for chronic pain, it is worthwhile to investigate how chronic pain and physical activity interact. Studies have investigated objectively-measured physical activity in subjects with chronic low back pain (Verbunt et al., 2001, Bousema et al., 2007, van Weering et al., 2009, Alschuler et al., 2011a, Alschuler et al., 2011b, Bussmann et al., 1998), hip and knee osteoarthritis (de Groot et al., 2008b, Murphy et al., 2008, Holsgaard-Larsen and Roos, 2012, Farr et al., 2008, Farr et al., 2010, White et al., 2013, White et al., 2012, White et al., 2014), rheumatoid arthritis (Prioreschi et al., 2013, Piva et al., 2010, Lee et al., 2012), fibromyalgia (Kop et al., 2005), and upper extremity complex regional pain syndrome (Schasfoort et al., 2004). Patient-reported outcomes of perceived function, pain, kinesiophobia, and pain catastrophizing were generally not highly associated with objectively-measured physical activity, suggesting that objectively-measured physical activity is a separate construct from patient reported outcomes (Verbunt et al., 2001, White et al., 2013, Bousema et al., 2007, van Weering et al., 2009, Piva et al., 2010, Alschuler et al., 2011a, Alschuler et al., 2011b).

Few studies have been published that use objectively measured physical activity as an outcome following acute musculoskeletal injury. In one study, investigators used wrist-mounted activity monitors over a three-week period to compare general activity in those with chronic versus acute low back pain (Liszka-Hackzell and Martin, 2004). In two other studies, investigators recorded weight bearing activity using thigh-mounted accelerometers over a 7-day period and compared two different treatment approaches after acute ankle sprain (Bleakley et al., 2010, Tully et al., 2012). These studies are among the few that used accelerometer-based activity monitoring as an outcome measure in acute orthopaedic rehabilitation. Only one of these papers provided raw values for steps taken per day after acute orthopaedic injury, highlighting the need for greater understanding of normal ranges after acute injury (Tully et al., 2012). In an excellent example of how physical activity monitoring may be used to provide evidence for therapeutic efficacy, Farr et al (Farr et al., 2010) evaluated the effect of a generalized progressive resistance training program in patients with early knee osteoarthritis and found that patients undergoing resistive training were significantly more active at 9 months after intervention versus patients with self-management alone.

Accelerometer-based activity monitoring has been studied in several orthopaedic post-surgical populations. For example, activity monitors have used to investigate post-surgical morbidity and subsequent recovery following major orthopaedic surgery prior to hospital discharge (Taraldsen et al., 2013). Several more studies have used accelerometers to record daily physical activity as a measure of intermediate and long-term recovery following surgery. This has been performed in subjects after rehabilitation from arthroscopic partial meniscectomy (Ilich et al., 2013), total knee or hip arthroplasty (de Groot et al., 2008a, Kuhn et al., 2013, Brandes et al., 2011), and lower extremity tumor resection (Bekkering et al., 2012, Rosenbaum et al., 2008). Similar trends emerge across these studies. First, patient-reported outcomes of perceived function and physical activity were poorly to moderately associated with physical activity and accounted for relatively little change in physical activity when evaluated with linear regression. Second, performance-based measures also were poorly associated with physical activity (de Groot et al., 2008a, Ilich et al., 2013). Poor-to-moderate association and low R^2 value between patient-reported outcomes and objectively-measured physical activity suggest that instruments of perceived function measure a different construct than actual daily physical activity in these populations. In addition, improvements in patient-reported outcomes and performance-based measures typically observed in these subject populations aren't necessarily accompanied by increases in objective physical activity.

Several recommendations exist when assessing physical activity in subjects with musculoskeletal pain. An accelerometer should be used for at least 5 days (to include one weekend day) to provide an estimate of habitual activity, data should be analyzed only from days that include over 10 hours of wear-time during waking hours, and accelerometer data should be augmented with a diary in order to account for causes of data interruption and non-wear time (Verbunt et al., 2012). In order to foster comparison between studies, data should be reported in terms of g's or m/s^2 rather than counts (Chen et al., 2012).

Physical activity monitoring has great potential in rehabilitation. For example, physical therapists may track weightbearing steps after knee surgery in order to determine how the volume and intensity of weightbearing outside of formal rehabilitation affects neuromuscular function and overall recovery. Patients with chronic pain and their therapists may collectively monitor general physical activity in an effort to increase patient compliance with activity recommendations and help understand how pain and activity interact. Following an injury involving the upper extremity, a wrist-worn activity monitor may be helpful in understanding how patients regain use of their arm and hand during daily life. Serial physical activity monitoring may also be helpful in determining treatment efficacy of procedures intended to improve activity tolerance and participation.

Wearable Sensors in Biomechanical Monitoring

Accelerometers have long been used in biomechanical laboratory-based studies. For example, accelerometers mounted on the lower trunk may reliably describe general parameters of gait such as cadence, step length, and walking speed, and have also been used to record specific gait events (Auvinet et al., 2002, Zijlstra and Hof, 2003, Kose et al., 2012, Kobsar et al., 2013). In addition, lower-trunk-mounted accelerometers may reliably discriminate between normal and altered walking patterns (Senden et al., 2011). When mounted at the shank, accelerometers show strong association with peak ground reaction force (Elvin et al., 2007), and when mounted at the shank and trunk simultaneously, they have characterized shock attenuation during running (Mizrahi et al., 2000). The research-grade devices used in studies such as these have been accelerometers designed primarily for use within a laboratory. These devices gather triaxial raw acceleration data at high collection frequencies (often up to 1000 Hz) over a range of acceleration wide enough to capture all physiologic acceleration (30 g). The data from these devices are precise and highly reliable due to strict calibration standards by the manufacturer. However, most devices designed for biomechanical research have not been wireless and were often quite expensive, thus restricting their use to a biomechanical research laboratory.

Technological advances in wearable accelerometer sensors not only have implications for physical activity monitoring, but also for potential employment as detectors of biomechanical events in non-laboratory settings (Horak et al., 2015). Investigators have recently begun to experiment with physical activity monitors for use in biomechanical studies. Raw data from these devices have shown strong association between lower trunk acceleration and vertical ground reaction force during walking, jogging, and jumping activities (Rowlands and Stiles, 2012, Rowlands et al., 2013). This relationship has many potential applications in in free-living situations. For example, the magnitude of acceleration may help quantify mechanical loading necessary for optimal bone health (Janz, 2003, Rowlands and Stiles, 2012, Stiles et al., 2013). It may also have application in understanding and being able to predict repetitive stress injuries (Neugebauer et al., 2014). In addition, our lab is investigating wearable accelerometers as a method to objectively assess asymmetry in gait and related functional tasks in non-laboratory settings. Technological advancements in wearable accelerometers now enable detailed analysis of movement for clinicians and researchers in non-laboratory settings.

The addition of other sensors onboard physical activity monitors permits further exploration of movement biomechanics. Specifically, gyro sensors and magnetometers may combine with accelerometers to increase 3-D movement recognition. Inertial Measurement Units (IMUs) that contain accelerometers, gyro sensors, and magnetometers are frequently described as possessing “9 degrees of freedom”, indicating that each type of sensor can detect perturbation in all three cardinal planes. Sensors such as these have been used in motion analysis research to describe limb and joint position in non-laboratory environments (Leardini et al., 2014, Luinge and Veltink, 2005, Seel et al., 2014, Horak et al., 2015).

Pilot work in our lab investigated the relationship between peak acceleration, peak vertical ground reaction force, and peak internal knee extension moment. We recorded acceleration from the lateral pelvis while subjects performed 5 trials of level walking and stepping down onto a force plate during ongoing gait from a 25.4 cm platform at 4 different cadences of gait. Acceleration data was sampled at 1000 Hz, force plate data sampled at 1000 Hz, and motion capture data sampled at 120 Hz simultaneously. Our results not only confirm those of previous studies suggesting a high association between vertical pelvic acceleration and vertical ground reaction force (r = 0.851) (Janz, 2003, Rowlands and Stiles, 2012, Stiles et al., 2013), but also suggest that vertical acceleration at the pelvis was strongly associated with knee extension moment (r = 0.824; Figure 2.1). These findings suggest that accelerometers may provide a clinically-feasible signature for underlying biomechanics. This has potential implication for accelerometers to be used in clinical settings to quantify functional limb symmetry after orthopaedic injury.

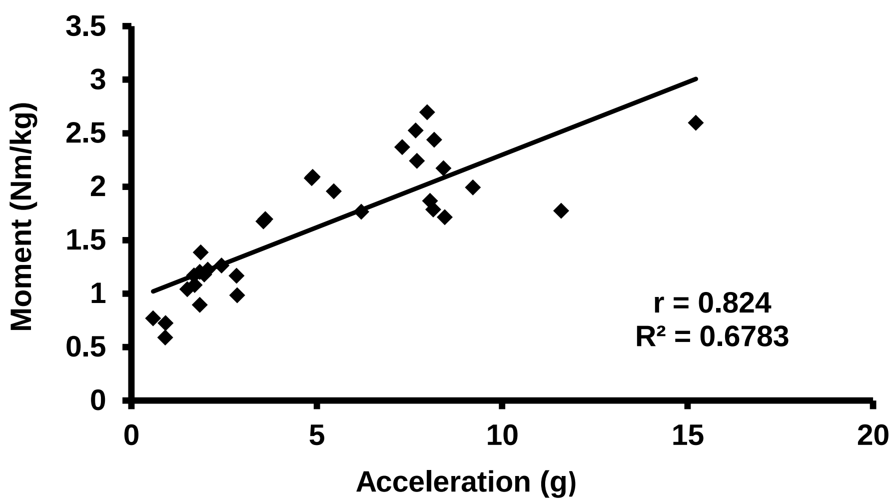


Figure 2.1: Peak vertical acceleration at the lateral pelvis vs. peak internal knee extension moment

Sleep Monitoring

Accelerometer-based sensors may also be used to provide an estimate of sleep in lieu of polysomnography. Polysomnography (PSG) is considered the gold standard of sleep monitoring but is complicated by equipment that may be deemed obtrusive by subjects and actually interfere with sleep. Estimating sleep with accelerometers, a practice known as sleep actigraphy, assesses sleep by monitoring movement. When a subject is moving, they are scored as awake, and when they are stationary for an extended period of time, they are scored as asleep. Algorithms that look at 30 or 60 second epochs of time have been validated for scoring sleep and are often included within software for activity and sleep monitors (Cole et al., 1992, Sadeh et al., 1994). Sleep actigraphy is more sensitive to wake time than sleep diaries (Martin and Hakim, 2011, Ancoli-Israel et al., 2003), actigraphy does not interfere with sleep to the same extent as PSG, and monitors can easily be worn for multiple nights (Martin and Hakim, 2011).

Because of these advantages, actigraphy has emerged as a reliable alternative to PSG and has been studied extensively in sleep medicine research. Studies comparing the validity of sleep actigraphy to PSG uniformly indicate high overall agreement rates between actigraphy and PSG, especially in normal subjects without disordered sleep (Van de Water et al., 2011b). In addition, the sensitivity of actigraphy in detecting sleep is very high in virtually all populations. That is to say, when a subject shows evidence of sleep via PSG, actigraphy also indicates sleep. But, the greatest concern is that the specificity of actigraphy is generally less than 50% (Paquet et al., 2007). That is to say, when there is evidence of wakefulness via PSG, actigraphy falsely scores much of this as sleep. This phenomenon can be explained because many people lie motionless even while awake. The poor rate of specificity for sleep actigraphy is systematic in nature (Pollak et al., 2001). Despite a systematic error in predicting wakefulness when compared to PSG, sleep actigraphy is highly consistent and thus is sensitive to detect changes in sleep behavior over time (Vallieres and Morin, 2003). Sleep medicine guidelines and review papers recommend sleep actigraphy for within-subject or longitudinal studies (Morgenthaler et al., 2007, Ancoli-Israel et al., 2003, 1995, Sadeh et al., 1995, Sadeh and Acebo, 2002, Sadeh, 2011, Van de Water et al., 2011b, Martin and Hakim, 2011). Although wear sites are reported at numerous locations on the body, the non-dominant wrist is by far the most common and is generally recommended (Ancoli-Israel et al., 2003).

Few studies have used actigraphy to monitor sleep in subjects with musculoskeletal pain, and most deal with subjects in chronic pain (Korszun et al., 2002, van de Water et al., 2011a, Harman et al., 2002, Lunde et al., 2010, Tang et al., 2012, Kop et al., 2005). In a trend similar to self-reported versus objectively measured physical activity, self-reported sleep measured via questionnaire and/or diary were not highly associated with objectively measured sleep with actigraphy and/or PSG (van de Water et al., 2011a). Furthermore, subjects with chronic pain and depression generally suffered more disordered sleep than subjects with chronic pain alone (Korszun et al., 2002). To our knowledge, no studies exist that investigate the relationship between acute pain, disordered sleep, and physical performance. We are also unaware of any studies investigating the interaction between physical therapy intervention and objectively-measured sleep. However, sleep actigraphy may be helpful for physical therapists to determine how their treatment approach interacts with sleep.

Postural Control

In clinical settings without access to a force plate capable of providing data on center of pressure excursion for balance and postural sway, accelerometers can provide objective assessment on control of the center of mass during functional tasks and balancing activities in subjects at risk for falls (Culhane et al., 2005, Senden et al., 2012, Whitney et al., 2011, Deshmukh et al., 2012, Janssen et al., 2008). Pelvic acceleration is significantly associated with force plate center of pressure (Whitney et al., 2011) and clinical balance tests like the Berg Balance Score and Timed Up and Go (O'Sullivan et al., 2009). Pelvic acceleration demonstrates high test-retest reliability during standardized tasks (Whitney et al., 2011) and is able to discriminate between subjects with and without a history of falls (Doheny et al., 2012). Furthermore, wireless accelerometer sensors may be used to monitor for remote fall detection (Kangas et al., 2008). Objective assessment of center of mass control not only has potential uses fall risk assessment in the elderly, but also for measuring control of the center of mass in orthopaedic and sports rehabilitation.

Community Integration

Global positioning sensors are nearly ubiquitous in most modern cell phones and other mobile devices like watches and tablet computers. Mobile applications use these sensors to provide user feedback on routes, distance, and speed of travel. Community integration is central to understanding the level of disability that patients may encounter as a result of their impairments. Thus, global position sensors have potential to objectively evaluate community integration and permit therapists to gain a deeper understanding of patients' disability. Two recent studies (Jayaraman et al., 2013, Hordacre et al., 2014) reported objectively measured physical activity (in discrete steps) and community integration in lower extremity amputees by integrating data from an accelerometer-based activity monitor attached to the a prosthetic leg and a separately worn GPS monitor into one data file. Similarly, Worringham et al. (Worringham et al., 2011) used a portable single-lead ECG logger, GPS sensor, and accelerometer-based activity monitor to track cardiac rehabilitation patients during remote exercise. Data were wirelessly transmitted to the user's cell phone for real-time user feedback, and were also transmitted remotely to a secure server for real-time monitoring by the cardiac rehabilitation professional.

Wearable Biosensors in Wellness & Human Performance

When training high level athletes, wearable systems of sensors capable of providing real-time feedback on training load and physiologic response to training are increasing in popularity. Accelerometry, GPS, and heart rate monitoring are commonly used to quantify and provide real-time feedback on individual training load (Cummins et al., 2013, Dellaserra et al., 2014). Meanwhile, sensors capable of monitoring skin galvanic response, body temperature, and blood chemistry are able to provide a comprehensive picture of the real-time physiologic status of the user (Chen et al., 2012). Resting heart rate, heart rate variability, sleep qualiity, and EEG are all used to assess for overreaching and assess athletes' readiness to adapt to additional training stimuli. The primary goals of using systems like these are to maximize performance, avoid overreaching, reduce the chance for injuries, and provide early recognition of heat injuries and adverse cardiac events. Many of these systems use biosensors that employ nanotechnology and can be as unobtrusive as a flexible stamp or removable tattoo.

Strength and conditioning professionals use integrated systems of IOMs to monitor physiological response to training and stress (HRV, HR, etc.) with the goal of individualizing the dosing parameters of training and rest, and recognizing when the body is most ready to respond to the additional physiologic stress of exercise. Systems like these have potential for use in rehabilitation as clinicians seek to provide the optimal dose at the optimal time. In addition, systems like these may help therapists to understand the physiologic effect of disease, emotional distress, sleep disturbance, and the implications these have on readiness for therapeutic exercise. Although patients may seem less likely to experience overtraining/overreaching when compared to elite athletes, patients are perhaps more susceptible to stress from pain, emotional distress, sleep disturbance, and lifestyle changes due to disability.

Accelerometers mounted on helmets, mouth guards, or directly on an athlete's head are used to quantify the frequency and magnitude of accelerations underlying concussion during sporting activity (Beckwith et al., 2013, Mihalik et al., 2012, Guskiewicz et al., 2007). Currently, detecting injury threshold has proven difficult to due to high individual variability of modifying factors for concussion and generally poor association between magnitude of head acceleration and clinical symptoms (Guskiewicz and Mihalik, 2011, Herring et al., 2011).

Challenges and Potential Issues

One of the major challenges with integrating rapidly evolving innovations in IOMs and mobile health is balancing their benefit versus HIPAA privacy concerns. The Department of Health and Human Services' omnibus final rule enacted in 2013 expanded HIPAA privacy standards. Now, innovative mobile technology like cloud storage providers and mobile monitors are considered "business associates" of covered entities in accordance with the Health Information Technology for Economic and Clinical Health (HITECH) Act (2013). This requirement poses difficulty given the fact that few developers of mobile applications devise their products with HIPAA compliance in mind (Luxton et al., 2012). Furthermore, many apps currently share user information with third parties for marketing purposes, which poses a massive threat to privacy in a healthcare context (Luxton et al., 2012, Steinhubl et al., 2015). Rather than relying on consumer-oriented devices, there is a need to develop medical sensor networks and applications for mHealth that incorporate additional security measures like password protection and encryption (Kumar and Lee, 2012).

In addition to HIPAA concerns, some clients may be uncomfortable with having their physical activity, location, sleep, and physiological status monitored and view this as a threat to personal privacy (Mohammadzadeh and Safdari, 2014, Al Ameen et al., 2012, Kumar and Lee, 2012). Therefore, it is important that clinicians and researchers be considerate of these concerns and accurately explain the potential benefit of monitoring and the extent to which data will be analyzed or used.

Any outcome measure has the potential to increase clinician and/or client burden, and using IOMs are no different. Although most wearable sensors are relatively unobtrusive, wearing them still poses some burden to the client. Therefore, devices should be selected that minimize this burden. Many devices are now water resistant, which allows for a wrist-worn monitor to be worn continuously during a period of data collection, even while bathing or swimming. This capability has led some researchers to attach these monitors with a semi-permanent method like a plastic wrist band, thus improving compliance for daily wear.

In addition to client burden, routinely using IOMs in clinical practice may introduce burden to the clinician. Although wearable sensors are able to provide vast amounts of information, the volume of data from these sensors can be large and difficult to handle (Troiano et al., 2014). For this reason, user-friendly software that interfaces between the sensors and a computer or portable electronic device is critical. Many user-friendly software programs are proprietary and require a significant annual licensing fee that may dissuade users with a limited budget. In addition, relying on programs that use proprietary scoring algorithms is dissuaded by researchers in favor of using open-source software that decreases cost to the user and also permits comparison of data across studies with different devices (Chen et al., 2012, Ward et al., 2005). For those with limited skills in computer programming, open source software is somewhat limited at this time. There is a need for open-source software that is user-friendly, intuitive, yet allows for rich sets of data to be rapidly processed.

Although wearable sensors can integrate data from multiple sources and create a comprehensive data set, this is not without technical challenges. When data are logged from multiple sensors at once (e.g., accelerometer, global positioning sensor, gyro sensor, and lux meter), data storage, battery life, and processing power are taxed. Thus, the time over which data can be collected is diminished considerably. Due to a finite amount of power within the data processors onboard wearable sensors, collecting extra sources of data is also more likely to lead to missing data. This is especially true when data are transmitted wirelessly, as is the case when a sensor is paired with a portable electronic device for real-time feedback. Data storage, battery life, and processing power are similarly reduced when data are collected at a higher sampling frequency. Thus, when designing protocols that use wearable sensors to collect and store data, clinicians and scientists must remain somewhat parsimonious in their approach.

Many consumer-oriented fitness trackers and cell phone accelerometers that use wireless networks and/or cloud storage currently lack the ability to provide raw acceleration data and therefore can’t be compared against gold standards to determine their validity and reliability. Until this occurs, fitness trackers should be considered of questionable quality for research and data reduction, especially in health care research. But, these devices have several outstanding attributes. They are generally easy to wear, attractive in appearance, offer feedback to the user, and have software programs or apps that are very easy to use. Clinicians, researchers, and industry should collaborate to include such user-friendly attributes into research-quality equipment that meets privacy standards for protected health information.

Conclusion

Wearable sensors have the potential to revolutionize the way healthcare is practiced. The capability and user-friendliness of wearable sensors will progress. These sensors will expand how we characterize patient outcome. They will assist us in understanding habitual free-living behavior that has profound impact on the treatment and prevention of disease and injury. They can also help us evaluate movement biomechanics when 3D motion capture systems are not practical. Wearable sensors will be critical in expanding patients' and clinicians' access to health information regardless of time or location. This empowers patients to take ownership of their own care and may provide additional motivation on rehabilitation compliance. Furthermore, wearable sensors may provide new insight into correct dosing of physical activity and exercise during rehabilitation. Expanded use of IOMs for these purposes will help us understand how habitual behavior affects recovery from injury or illness and how physiologic and emotional stress affects readiness for therapeutic intervention. This awareness will enable us to implement formal rehabilitation when clients are best able to adapt to therapeutic stress, and will allow us to optimally dose therapeutic intervention outside of formal rehabilitation.

CHAPTER 3

WALKING AND STEPPING DOWN: A SIMPLE AND RELEVANT FUNCTIONAL OUTCOME MEASURE

Introduction

Knee trauma and surgery, including arthroscopic partial meniscectomy and anterior cruciate ligament reconstruction, frequently lead to significant muscle atrophy (Mendias et al., 2013), weakness (Williams et al., 2005, Sturnieks et al., 2008a, Gapeyeva et al., 2000), activation failure (Urbach and Awiszus, 2002, Hart et al., 2010b, Glatthorn et al., 2010), and altered mechanics (Hall et al., 2012, Sturnieks et al., 2008a, Sturnieks et al., 2008b). The most pronounced adaptations in knee mechanics occur in the sagittal plane during the loading response phase of gait and include reductions in internal knee extension moment, peak knee flexion angle, and sagittal plane knee excursion from initial contact to loading response. Several mechanisms have been theorized to contribute to these adaptations, including quadriceps muscle weakness (Lewek et al., 2002) and altered patterns of neuromuscular control (Williams et al., 2004).

When compared with level walking, characteristic alterations in knee mechanics show larger effect sizes during the performance of tasks requiring greater demand on the quadriceps. This relationship is consistent across tasks such as jogging (Rudolph et al., 2001, Kuenze et al., 2013), stair ascent and descent (Thambyah et al., 2004, Hooper et al., 2002, Gao et al., 2012, Hall et al., 2012), vertical jump take-off and landing (Ernst et al., 2000), and while stepping down while walking (Houck and Yack, 2003).

Stepping down while walking (i.e., walk and step down test) is a common functional task most people perform daily throughout all stages of life. For example, people perform this when stepping off a curb to cross a street or when entering a parking lot after exiting a store. This routine task places significant demand on the thigh muscles, which control the descent of the body's mass. Houck and Yack (2003) report that stepping down like this requires similar quadriceps torque as running at 3.2 m/s (7.16 mph). This may be one of few performance-based measures appropriate for a wide range of patients with varying tolerances for physical activity throughout the lifespan. Based on its outstanding functional relevance, high quadriceps demand, and simplicity to perform in a clinical setting, this may be an excellent candidate for use as performance-based clinical outcomes measure for a wide variety of patients.

In musculoskeletal and sports injury research the current standard for obtaining biomechanical information is with instrumented motion analysis systems. These systems require expensive equipment, dedicated space, and lengthy times for data collection that are prohibitive in most clinical settings and in large multi-center research studies. Thus, there is a capability gap for those in clinical settings to evaluate for biomechanical changes that are clinically meaningful yet not detectable by routine clinical examination. It is possible that meaningful data about lower extremity biomechanics may be obtained quickly from only a force platform and raised walkway that may be constructed inexpensively or even improvised by using the back of a treadmill.

Performance on the walk and step down task has shown to be dependent upon parameters like step height, foot strike pattern, and gait velocity. A forefoot strike pattern of the leading leg is more common at lower velocities of gait and with higher step height (van Dieen et al., 2008, Spanjaard et al., 2008). When compared to heel striking, a forefoot strike pattern shifted more negative work toward the leading leg, in particular toward the ankle plantar flexors. This method attenuated peak vertical ground reaction force due to earlier ground contact of the leading leg, lower downward velocity of the center of mass prior to impact, and longer stance time of the leading leg. Forefoot striking also led to lower internal knee extension moment and negative work at the knee of the leading leg (van Dieen et al., 2008).

Performance on the walk and step down task has shown to change in response to muscle force output and aging. In response to quadriceps or triceps surae muscle fatigue, subjects demonstrated a shift in work toward non-fatigued muscle groups (Barbieri et al., 2014). In addition, quadriceps fatigue led to increased variability of trailing leg placement, increased step width, and decreased step duration of the leading leg (Barbieri et al., 2014, Barbieri et al., 2013). When compared with younger subjects, the elderly favored a forefoot method of foot strike for the leading leg, showed higher downward velocity of their centers of mass prior to foot strike, and landed with a “stiffer” leg using a smaller range of sagittal plane knee motion (van Dieen and Pijnappels, 2009, Hortobagyi and DeVita, 1999). Although these studies did not report measures of quadriceps strength, such differences in performance suggest lower normalized quadriceps strength in the elderly when compared with healthy younger subjects, a commonly-reported finding elsewhere (Skelton et al., 2002, Moreland et al., 2004).

The walk and step down task has not been evaluated as an outcome measure in subjects after knee surgery who are at risk for quadriceps dysfunction and altered mechanics. It is unknown how people’s performance of this task changes after knee surgery. The relationships between ground reaction force data from the walk and step down task and measures of quadriceps function and commonly-reported patient-reported outcomes have not been evaluated.

The aims of this study were to: 1) identify differences in performance of the walk and step down task between limbs and across time for subjects after knee surgery, and 2) identify relationships between ground reaction force data from stepping down (peak vertical ground reaction force amplitude and timing within stance, overall stance time, and method of foot strike), measures of quadriceps function (strength, voluntary activation, and atrophy), and patient-reported outcomes (Pain Visual Analog Scale and KOOS) before and after knee surgery. I hypothesized that differences in vertical ground reaction force amplitude, timing, and methods of foot strike would occur between limbs at all measurement points, but would be most pronounced early after surgery. In addition, I hypothesized that a moderate to strong relationship ($r < -0.50$) would be present between ground reaction force data and measures of quadriceps function of the trailing limb, while a low to moderate relationship ($r < -0.25$) would exist between ground reaction force data and patient-reported outcomes relating to the trailing limb. Overall, I expected that information gathered from walking and stepping down would be complementary to patient-reported outcomes and clinical exam and help fill a capability gap for clinicians and researchers in clinical settings.

Methods

Design: This study employed a longitudinal design intended to appreciate differences in in how the walk and step down test is performed in subjects undergoing one of two common knee surgeries.

Subjects: This study included a total of 28 subjects who had one of two common knee surgeries: Arthroscopic partial meniscectomy (APM, n = 18) or anterior cruciate ligament reconstruction (ACLR, n = 10). APM subjects were included after informed consent if they were between 18 and 65 years old and were scheduled to undergo knee arthroscopy for a meniscus tear at the University of Iowa Hospitals and Clinics. Subjects were excluded for the following reasons:1) Concomitant knee ligament injury; 2) fracture of the femur, tibia, fibula, or patella within the prior year; 3) history of a lower body nerve injury, lumbar radiculopathy, or neurological disorder; 4) history of a quadriceps muscle tear; 5) BMI greater than 40; 6) the history or presence of another medical condition that is likely to affect the person's ability to safely perform the study or would impact the validity of the results; 7) inability to have an MRI; and 8) pregnancy. Subjects were also excluded from analysis if the surgeon requested the subject not bear weight throughout the available range of motion, and were therefore unable to participate in the walk and step down test. APM subjects in this study were participants in a clinical trial investigating the effect of different rehabilitation protocols on neuromuscular performance and functional outcome.

ACLR subjects were included if they were between the ages of 18 and 35 years old and regularly participated in level 1 or 2 activity prior to injury. Subjects were excluded for the following reasons: 1) Multiligamentous injury; 2) fracture of the femur, tibia, fibula, or patella within the prior year; 3) concomitant surgery requiring a modified rehabilitation protocol; 4) history of a lower body nerve injury, lumbar radiculopathy, or neurological disorder; 5) history of a quadriceps muscle tear; 6) BMI greater than 40 kg/m^2; 7) the history or presence of another medical condition that is likely to affect the person's ability to safely perform the study or would impact the validity of the results; 8) inability to have an MRI; and 9) pregnancy.

Data collection: Data for all subjects were collected at three time points: Approximately 1 week pre-operatively, 2-weeks postoperatively, and 5 weeks postoperatively. Subjects with ACL reconstruction were also tested at 6 months postoperatively.

Walk and step down: For the walk and step down test, subjects walked off a raised platform 25.4 cm high and 2.44 m long and stepped onto a portable force plate (Accugait, AMTI, Watertown, MA). The length of the platform allowed for three steps prior to stepping down. Because the aim of this study was to identify how patients choose to walk and step down (an ecologically-valid approach), the velocity of gait, cadence, type of footwear, and method of foot strike onto the force plate were not controlled. Rather, subjects were instructed to step off the platform and continue walking in the same manner they would step off a curb to cross the street. After three to five trials of familiarization per side, 5 trials were first collected with the subject stepping down onto the force plate with the involved leg, followed by 5 trials stepping down onto the uninvolved leg. Because of the high demand required of the trailing leg to lower the center of mass, this order of testing was selected in order to afford subjects familiarization with the task in the less demanding of the two situations first. Force plate data were collected at 200 Hz and analyzed with NetForce software (AMTI, Watertown, MA). Dependent variables of interest from the force plate included 1) Peak vertical ground reaction force (VGRF) during deceleration; 2) Relative timing of peak deceleration VGRF, expressed as a percent of stance phase; 3) Impulse of VGRF from initial contact to peak deceleration; 4) Impulse of VGRF from initial contact to midstance, 5) Stance time on the force plate; and 6) Method of foot strike with the leading leg (Figure 3.1). For continuous variables, the mean of 5 trials was computed for each side at each time point.

Neuromuscular testing: In order to permit comparison of differences on the walk and step down test to differences in quadriceps neuromuscular function, subjects underwent neuromuscular strength testing bilaterally, with the uninvolved leg tested first. Subjects sat on a HUMAC NORM Testing and Rehabilitation System (CSMI, Stoughton, MA) with the hips and knees flexed to 85 and 90 degrees, respectively. A seat belt, chest straps, and thigh strap secured subjects to the chair and subjects folded their arms across their chest. This angle of knee flexion minimized strain on the ACL graft (Escamilla et al., 2012) and allowed for consistent measurement between subjects with ACL reconstruction and arthroscopy. A Velcro strap secured the lower limb of subjects to the torque arm of the dynamometer with the distal edge of the shin pad placed approximately 5 cm proximal to the medial malleolus. A 2 x 2 inch adhesive stimulating electrode was placed over the femoral nerve at the femoral triangle and connected to an FDA-approved stimulator (model D57AH, Digitimer Ltd., Hertfordshire, England).

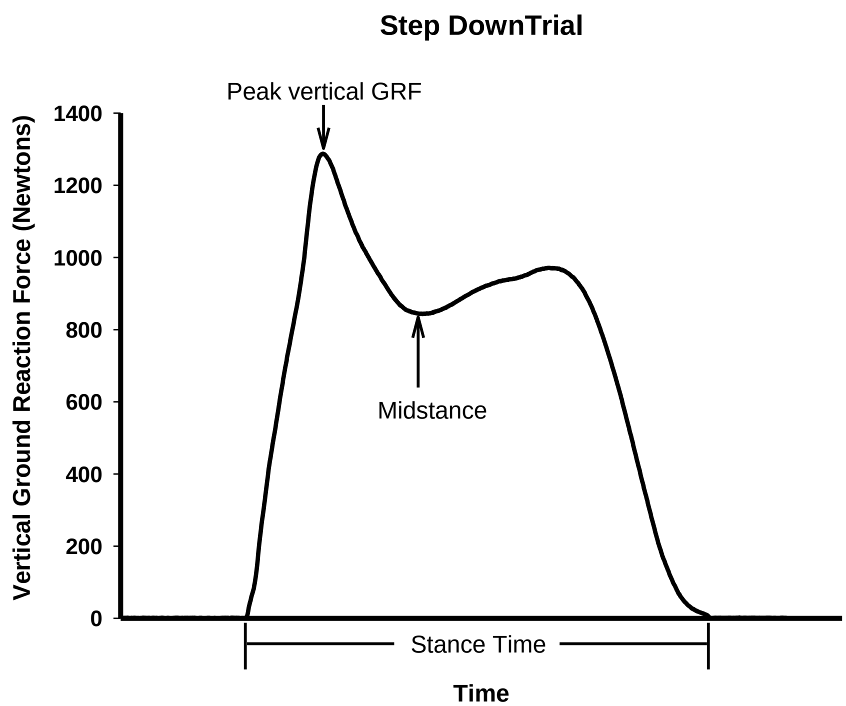


Figure 3.1: Example of vertical ground reaction force during a step down trial

Subjects performed familiarization trials at 50%, 75%, and 90% of perceived effort for knee extension and knee flexion. After these, subjects performed at least two maximal voluntary isometric contractions (MVICs) of 5 seconds duration with two minutes of rest in between trials for knee extension and knee flexion. Loud verbal encouragement and real-time feedback of torque development were provided to encourage maximal effort. For consistency, subjects were required to perform a minimum of two trials within ± 5% of each other. After subjects demonstrated consistent voluntary MVIC, they performed a minimum of two additional trials of knee extension in order to determine voluntary activation of the quadriceps. For each trial, subjects received a torque-triggered superimposed doublet at supramaximal stimulation (100 Hz, 1000 µs, 400 V) after they reached the force threshold defined by baseline MVIC testing (Krishnan et al., 2009, Krishnan and Williams, 2010). Voluntary activation was calculated via the interpolated twitch technique, although in this study a supramaximal 100 Hz doublet was used rather than a twitch. For consistency, subjects were required to perform a minimum of two trials within +/- 5% of each other. Data were sampled at 1000 Hz, collected with an AD Instruments PowerLab 16/30, and processed via LabChart 7 software (AD Instruments, Bella Vista, Australia).

Imaging: Axial T1-weighted magnetic resonance imaging (MRI) images were obtained from the level of the tibial tubercle to the iliac crest of both limbs at a slice thickness of 5 mm and gap of 10 mm using a 3.0T Siemens TIM Trio scanner (Siemens, Munich, Germany). Cross-sectional area of the quadriceps muscle group was manually traced and quadriceps volume computed using MIPAV software (CIT, NIH). Quadriceps volume was considered in terms of raw volume and as a percent of the uninvolved limb at the same time point.

Patient reported outcomes: Subjects completed patient-reported outcomes to gain insight into self-perceived functional status, pain, and physical activity. After all walk and step down trials were complete, subjects rated the pain experienced during the walk and step down test on a visual analog scale. Subjects also completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) (Roos et al., 1998b) and UCLA Activity Rating scales (Amstutz et al., 1984, Zahiri et al., 1998). The KOOS is highly reliable and responsive to change (Roos et al., 1998b, Roos et al., 1998a). The UCLA Activity score ranges from 1 to 10, with a lower score indicating less activity. Subjects checked one of 10 boxes that best represented the intensity of their physical activity from 1 ("no physical activity, dependent on others") to 10 ("regular participation in impact sports").

Data analysis: Demographic information from the APM and ACLR groups were compared with Mann-Whitney U-tests with the level of significance set to 0.05 in order to determine difference between groups. A two-way repeated measures ANOVA with a level of significance of 0.05 was used to investigate differences limbs and time points for variables of WSD performance and quadriceps function. A 2x3 model was applied to each force plate variable and variable of quadriceps function for all subjects together and the APM group individually. Because of the addition of the fourth time point for ACLR subjects, a 2x4 repeated measures ANOVA evaluated for differences in this group. The assumption of sphericity was assessed with Mauchly's test. In cases where sphericity was violated, corrected levels of significance were used according to the following convention: Greenhouse-Geiser correction was used when the computed Epsilon was 0.75 or greater and the more conservative Huynh-Feldt correction was used when the computed Epsilon was less than 0.75 (Peat, 2014). When overall significance was reached, post-hoc testing was performed using Bonferroni-corrected pairwise comparisons. For limb-symmetry indices calculated from force platform or strength data, a one-way repeated measures ANOVA model with a level of significance at 0.05 was used to investigate differences over time. Effect sizes for variables that attained statistical significance were obtained from Partial Eta squared values and converted to values for Cohen's *f* via the following equation: $=Cohen's\, f = \sqrt{\eta^2}/(1-\eta^2)$. When interpreting Cohen's *f*, a value of 0.1 is considered a small effect size, 0.25 is considered a medium effect size, and 0.4 considered a large effect size (Peat, 2014, Cohen, 1992).

To determine the association between performances on the walk and step down test and measures of quadriceps function and patient-reported outcome, Pearson's product moment correlation matrices were computed. Separate matrices were computed between force plate variables from the leading limb and quadriceps data for the leading and trailing limbs (Figure 3.2). This permitted insight into the role of quadriceps function of both limbs while stepping down during ongoing gait.

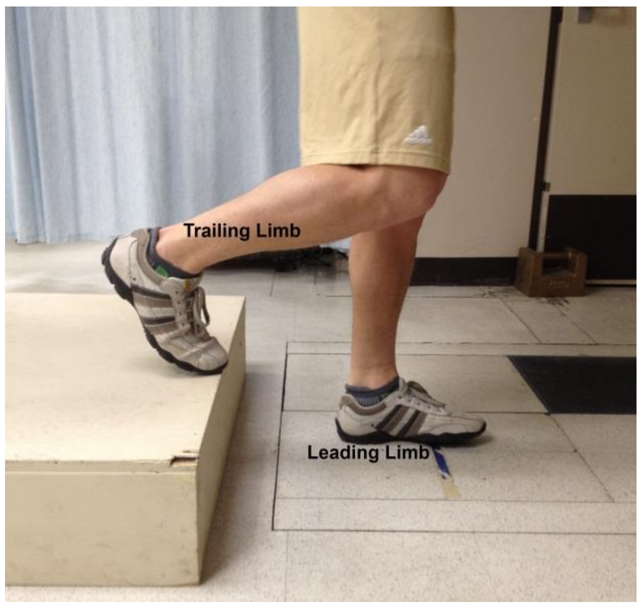


Figure 3.2: Example of a step down trial

In order to investigate how subjects with high and low quadriceps asymmetry differed in completing the WSD task, all subjects and data collection time points were pooled and then divided into quartiles based on quadriceps index. The assumption of homogeneity of variance for all variables of interest was evaluated with Levene's test. The Kruskal-Wallis test was performed to evaluate for differences between quartiles. In cases of statistical significance from the Kruskal-Wallis test, pairwise post-hoc testing was performed using Mann-Whitney U Tests with a level of significance of 0.05. Nonparametric tests were used because sample sizes were not equal for all quartiles and several variables of interest did not meet the assumption of homogeneity of variance (atrophy, foot strike method, asymmetry of stance time, and timing of F_{zMax}). SPSS version 23 software performed all statistical analysis (SPSS Inc, Chicago, IL).

Results

Compared with the ACL group the APM group was significantly older. The APM group also scored lower on pre-operative KOOS, yet this difference was not statistically significant due to small sample size. Other characteristics were similar (Table 3.1).

For all trials of the walk and step down task, peak vertical ground reaction force (GRF) was observed at the deceleration peak, which occurred at approximately 15-20% of stance during the loading response during gait (Figure 3.1). This differs from level walking, in which the vertical ground reaction force peaks from the deceleration and acceleration phases are similar and peak vertical GRF may occur either during deceration (weight acceptance) or acceleration (push-off).

With APM and ACLR subjects analyzed together, several between-limb differences were observed in ground reaction force data (Table 3.2). When compared to the uninvolved limb, the involved limb demonstrated shorter stance time, lower amplitude of peak vertical GRF, later onset of peak vertical GRF within stance, and lower values for the impulses From initial contact to peak vertical GRF and from initial contact to midstance. No significant main effects for Time or significant interactions of Limb x Time were observed for any variable. Subjects favored the forefoot method of foot strike bilaterally at all data collection points. Between-limb differences were also observed in normalized quadriceps strength, quadriceps volume, voluntary activation, and specific torque. The involved quadriceps was significantly atrophied and weaker than the uninvolved. In addition, the involved quadriceps demonstrated greater strength and specific torque at 4 weeks when compared to 2 weeks, and greater voluntary activation at 4 weeks when compared to the pre-operative time point.

	APM subjects	**ACLR subjects**
Sex	11 M, 7 F	5 M, 5 F
Inv Side	8 R, 10 L	6 R, 4 L
Age	42.3* ± 14.7	25.3 ± 6.8
Height (m)	1.7 ± 0.10	1.8 ± 0.08
Mass (kg)	80.4 ± 11.5	82.4 ± 13.9
BMI	26.5 ± 2.9	27.0 ± 4.5
KOOS4	48.0 ± 13.3	57.9 ± 12.8
IKDC	n/a	55.2 ± 18.7
UCLA Activity	6.5 ± 2.2	5.8 ± 2.2

Table 3.1 Subject demographics

Values are mean ± standard deviation

*$P < 0.05$

Effect sizes for between-limb differences in ground reaction force variables that attained statistical significance were high, and compared favorably to the effect sizes seen for variables of quadriceps performance (Table 3.3). The largest effect size from among the force plate variables was normalized peak vertical GRF. This was similar to the effect size for normalized quadriceps strength.

Relationships between variables for ground reaction force and quadriceps function of the leading limb were all below 0.50 with the exception of limb symmetry indices of peak vertical GRF and quadriceps strength, which was 0.548 (Table 3.4).

Variable	Limb	Pre-Op	2 weeks	4 weeks	Limb	Time	Interaction
Stance Time (sec)	Involved	0.70 ± 0.09	0.70 ± 0.08	0.68 ± 0.08	0.004	0.051	0.081
	Uninvolved	0.72 ± 0.13	0.76 ± 0.15	0.72 ± 0.11			
Peak Vertical GRF (% body weight)	Involved	169 ± 42%	158 ± 47%	168 ± 42%	<0.001	0.106	0.062
	Uninvolved	193 ± 43%	193 ± 44%	200 ± 44%			
Peak Vertical GRF Limb Symmetry		0.88 ± 0.11	0.82 ± 0.15[‡]	0.85 ± 0.13		0.053	
Peak Vertical GRF timing (% stance)	Involved	14.9 ± 5.5%	17.3 ± 4.9%	15.8 ± 4.2%	<0.001	0.168	0.076
	Uninvolved	13.7 ± 4.8%	13.7 ± 4.6%	13.5 ± 4.6%			
Peak Vertical GRF Impulse (N/sec)	Involved	80 ± 27	86 ± 27	81 ± 22	0.005	0.199	0.987
	Uninvolved	88 ± 37	95 ± 29	90 ± 31			
Midstance Impulse(N/sec)	Involved	287 ± 61	282 ± 67	292 ± 57	<0.001	0.413	0.153
	Uninvolved	325 ± 68	317 ± 65	311 ± 64			
Strike Method (% ff strike)	Involved	70 ± 38%	80 ± 33%	79 ±35%	0.42	0.179	0.153
	Uninvolved	76 ± 38%	84 ± 34%	79 ±39%			

Table 3.2: Changes in force plate variables over time in both limbs for arthroscopic partial meniscectomy and ACL-reconstructed subjects grouped together

Values are mean ± standard deviation

P < 0.05

GRF = ground reaction force ff = forefoot

Variable	Effect Size
Stance Time	0.55
Normalized peak vertical GRF	1.39
Peak vertical GRF Timing	0.88
Peak vertical GRF Impulse	0.54
Midstance Impulse	0.72
Quadriceps Volume	0.87
Normalized Quadriceps Strength	1.32
Quadriceps Voluntary Activation	0.41
Quadriceps Specific Torque	1.56

Table 3.3: Effect sizes for ground reaction force and quadriceps data (Cohen's *f*)

Ground reaction force parameters best associated with leading limb quadriceps function were peak vertical GRF, peak vertical GRF limb symmetry index, and stance time. Conversely, measures of quadriceps function best associated with these ground reaction force parameters include normalized quadriceps strength, quadriceps strength symmetry, and atrophy. Quadriceps weakness and atrophy of the leading limb were associated with longer stance time, lower peak vertical GRF, greater asymmetry of peak vertical GRF, and later timing of peak vertical GRF within stance. Higher KOOS scores were associated with shorter stance times and higher peak vertical GRF onto the involved limb.

Relationships between variables for ground reaction force and quadriceps function of the trailing limb also rarely exceeded 0.50 (Table 3.5). Ground reaction force parameters best associated with trailing limb quadriceps function were peak vertical GRF, peak vertical GRF limb symmetry index, midstance impulse, and stance time.

	KOOS4	Leading Quad Volume	Atrophy	Leading Quad MVIC	Quad Index	Leading Quad VA	Leading Spec Torque
Stance Time	-0.445‡	-0.160*	0.280‡	-0.232‡	-0.328‡	-0.132	0.005
Peak Vertical GRF	0.221*	0.155*	-0.439‡	0.431‡	0.436‡	0.027	0.301‡
Peak Vertical GRF LSI	0.255*	0.177	-0.458‡	0.360‡	0.548‡	0.190	0.283‡
Peak Vertical GRF timing	-0.111	-0.188‡	0.104	-0.179*	-0.172	0.162*	-0.024
Peak Vertical GRF Impulse	0.030	0.153*	-0.042	0.102	0.001	0.195‡	0.305‡
Midstance Impulse	-0.037	0.423‡	0.057	0.156*	-0.034	0.081	0.282‡

Table 3.4: Relationships between step down parameters and quadriceps data from the leading limb for all subjects. Relationships between step down parameters and KOOS scores, atrophy, and Quad Index are for the involved limb

LSI = Limb Symmetry Index (Involved / Uninvolved)

GRF = ground reaction force VA = Voluntary Activation

*P < 0.05 $^{\ddagger}P$ < 0.01

	KOOS4	Trailing Quad volume	Atrophy	Trailing Quad MVIC	Quad index	Trailing Quad VA	Trailing Spec Torque
Stance Time	-0.502‡	-0.245‡	0.183	-0.392‡	-0.523‡	-0.233‡	-0.179*
Peak vertical GRF	0.072	-0.020	-0.083	0.197‡	0.119	-0.117	0.059
Peak vertical GRF LSI	0.255*	0.178	0.011	0.377‡	0.548‡	0.250*	0.335‡
Peak vertical GRF timing	0.138	-0.093	0.074	0.022	0.091	0.216‡	0.202‡
Peak vertical GRF Impulse	0.031	0.084	0.017	0.014	-0.105	0.116	0.238‡
Midstance Impulse	-0.093	0.346‡	0.098	0.022	-0.204*	-0.075	0.140

Table 3.5: Relationships between step down parameters and quadriceps data from the trailing limb for all subjects. Relationships between step down parameters and KOOS scores, atrophy, and Quad Index are for the uninvolved limb

LSI = Limb Symmetry Index (Involved / Uninvolved)

GRF = ground reaction force VA = Voluntary Activation

*$P < 0.05$ $^{\ddagger}P < 0.01$

Similar to relationships with the leading limb, measures of quadriceps function of the trailing limb highest associated with these step down parameters include normalized quadriceps strength, quadriceps strength symmetry, and atrophy. Quadriceps weakness and strength asymmetry of the trailing limb were associated with longer stance time, lower peak vertical GRF, and greater asymmetry of peak vertical GRF. But, quadriceps weakness had no effect on the timing peak vertical GRF within stance. Higher KOOS scores were associated with shorter stance times onto the uninvolved limb, but were not significantly associated with the amplitude or timing of peak vertical GRF.

Trials within the upper quartile for quadriceps index also demonstrated high symmetry between limbs in the amplitude of peak vertical GRF (Table 3.6). Compared to other quartiles, the first quartile also demonstrated greatest symmetry in overall stance time. The upper quartile also demonstrated greatest symmetry for the timing of peak vertical GRF, although this was not statistically significant from other quartiles. Symmetry for the amplitude of peak vertical GRF, the timing of peak vertical GRF, and overall stance time decreased with each lower quartile. Thus, trials in the fourth quartile demonstrated lowest symmetry in the amplitude of peak vertical GRF, the timing of peak vertical GRF, and stance time. A large disparity existed between quartiles for the coefficients of determination between quadriceps index and peak vertical GRF limb symmetry index. For the first and fourth quartiles, R^2 values were significant and moderate, indicating that quadriceps muscle symmetry significantly predicted symmetry of peak vertical GRF for trials in these quartiles. But, this same relationship was not observed for the middle two quartiles, where the coefficient of determination between quad index and peak vertical GRF limb symmetry index was weak and not significant.

Despite differences between quartiles, all four quartiles also demonstrated similarities. The majority of trials within every quartile demonstrated earlier timing of peak vertical GRF onto the uninvolved limb versus the involved limb. Furthermore, the proportion of trials within each quartile that demonstrated quicker peak vertical GRF onto the uninvolved limb was similar (70%, 67%, 83%, and 74% of trials in the first, second, third, and fourth quartiles, respectively). Trials in all quartiles demonstrated preference for the forefoot method for both limbs.

With APM subjects analyzed as a subgroup, several between-limb differences were observed in ground reaction force data (Table 3.7). When compared to the uninvolved limb, the involved limb demonstrated lower amplitude of peak vertical GRF, later onset of peak vertical GRF within stance, and lower values for the impulse from initial contact to midstance. Significant differences were not observed for stance time, peak vertical GRF limb symmetry index, the impulse from initial contact to peak vertical GRF, or method of foot strike. No significant main effects for Time or significant interactions of Limb x Time existed for any variable. Subjects favored the forefoot method of foot strike bilaterally at all data collection points.

With ACLR subjects analyzed as a subgroup, significant main effects for stance time and peak vertical GRF were observed for both Time and Limb factors. Significant Limb x Time interactions were also observed for stance time and the amplitude of peak vertical GRF (Table 3.8). Compared to the uninvolved limb, the involved limb demonstrated shorter overall stance time and lower overall amplitude of peak vertical GRF. At two weeks, subjects demonstrated longer overall stance time and higher overall peak vertical GRF when compared to four weeks and six months. At six months, subjects demonstrated shorter overall stance time compared to 2 weeks and 4 weeks. Differences in stance time between limbs were greatest at 2 weeks due to significantly longer stance time onto the uninvolved limb when compared to the same limb at 6 months. Stance time at 6 months was the most symmetrical between limbs and also the

Quartile (mean QI)	Peak VGRF LSI	R^2 QI vs. Peak VGRF LSI	Δ Peak VGRF Timing (Inv-UI; %stance)	Δ Stance Time (Inv-UI; sec)	Forefoot Strike	KOOS4	Atrophy
First (0.99)	0.93 ± 0.07*‡	0.18	1.23 ± 3.48%	-0.013 ± 0.06*	73%	65 ± 17*	2 ± 5%*‡
Second (0.84)	0.88 ± 0.04*	0.051	1.37 ± 3.45%	-0.015 ± 0.04*	77%	57 ± 20	4 ± 4%*‡
Third (0.75)	0.85 ± 0.03	0.0086	2.39 ± 3.21%	-0.030 ± 0.07	78%	61 ± 16*	14 ± 6%*
Fourth (0.56)	0.77 ± 0.14	0.29	4.01 ± 5.87%	-0.090 ± 0.13	83%	48 ± 14	21 ± 9%

Table 3.6: Description of performance on the walk and step down task based on quartiles grouped by quadriceps index (QI)

Values are as mean ± standard deviation

*$P < 0.05$ ‡$P < 0.01$

Variable	Limb	Pre-Op	2 weeks	4 weeks	Limb	Time	Interaction
Stance Time (sec)	Involved	0.71 ± 0.10	0.70 ± 0.07	0.69 ± 0.08	0.126	0.488	0.912
	Uninvolved	0.73 ± 0.14	0.73 ± 0.11	0.72 ± 0.12			
Peak Vertical GRF (% body weight)	Involved	174 ± 46%	175 ± 47%	175 ± 46%	<0.001	0.982	0.704
	Uninvolved	200 ± 51%	200 ± 52%	198 ± 53%			
Peak Vertical GRF Limb Symmetry		0.88 ± 0.11	0.88 ± 0.12	0.89 ± 11		0.839	
Peak Vertical GRF timing (% stance)	Involved	15.0 ± 5.2%	16.3 ± 4.7%	16.3 ± 4.6%	0.008	0.148	0.704
	Uninvolved	13.2 ± 5.4%	15.2 ± 4.1%	14.5 ± 4.9%			
Peak Vertical GRF Impulse (N/sec)	Involved	80 ± 28	85 ± 24	84 ± 19	0.12	0.07	0.141
	Uninvolved	80 ± 36	97 ± 30	90 ± 32			
Midstance Impulse(N/sec)	Involved	275 ± 49	285 ± 63	286 ± 52	0.007	0.992	0.713
	Uninvolved	321 ± 65	311 ± 56	308 ± 60			
Strike Method (% ff strike)	Involved	75 ± 34%	82 ± 29%	80 ± 37%	0.539	0.408	0.713
	Uninvolved	79 ± 37%	89 ± 29%	80 ± 39%			

Table 3.7: Changes in force plate variables over time in both limbs for subjects undergoing arthroscopic partial meniscectomy

Values are mean ± standard deviation

$P < 0.05$

GRF = ground reaction force ff = forefoot

Variable	Limb	Pre-Op	2 weeks	4 weeks	6 months	Limb	Time	Interaction
Stance Time (sec)	Involved	0.68 ± 0.09	0.71 ± 0.10* ‡**	0.66 ± 0.08 **	0.64 ± 0.08	0.018	<0.001	0.031
	Uninvolved	0.71 ± 0.14	0.83 ± 0.19†	0.71 ± 0.10	0.64 ± 0.09			
Peak Vertical GRF (% body weight)	Involved	162 ± 33%	128 ± 32%*† ‡**	154 ± 29%*	178 ± 36%	<0.001	<0.001	<0.001
	Uninvolved	180 ± 16%	182 ± 22%	202 ± 20%	197 ± 29%			
Peak Vertical GRF Limb Symmetry		0.89 ± 0.12	0.70 ± 0.13**•	0.76 ± 0.12**	0.90 ± 0.11		<0.001	
Peak Vertical GRF timing (% stance)	Involved	14.7 ± 6.1%	19.2 ± 5.1%*	14.8 ± 3.6%	14.4 ± 6.3%	0.002	0.452	0.01
	Uninvolved	14.4 ± 3.6%	11.0 ± 4.3%	11.7 ± 3.6%	13.4 ± 6.4%			
Peak Vertical GRF Impulse (N/sec)	Involved	80 ± 26	88 ± 34	76 ± 26	92 ± 31	0.014	0.278	0.390
	Uninvolved	103 ± 35	91 ± 29	91 ± 30	99 ± 36			
Midstance Impulse(N/sec)	Involved	307 ± 76	278 ± 76	302 ± 66	308 ± 69	0.003	0.118	0.4529
	Uninvolved	333 ± 76	326 ± 81	315 ± 73	333 ± 72			
Strike Method (% ff strike)	Involved	60 ± 45%	76 ± 40%	78 ± 33%	70 ± 45%	0.568	0.394	0.361
	Uninvolved	72 ± 41%	74 ± 41%	76 ± 42%	74 ± 41%			

Table 3.8: Changes in force plate variables over time in both limbs for subjects undergoing ACL reconstruction

Values are mean ± standard deviation

GRF = ground reaction force ff = forefoot

‡Main effect for Time from values at 4 weeks, *P* < 0.05

** Main effect for Time from values at 6 months, *P* < 0.05

•Main effect for Time from values pre-operatively, *P* < 0.05

†Within-limb differences from values at 6 months, *P* < 0.0083

*Significant vs. Uninvolved at same time point, *P* < 0.0083 shortest in duration. Differences between limbs for peak vertical GRF were large and significant at two weeks and four weeks, and were driven by significantly lower values onto the involved at 2 weeks and a combination of significantly high values onto the uninvolved and relatively low values onto the involved at 4 weeks.

When compared to the uninvolved limb, the involved limb demonstrated smaller impulses. No significant main effect for Time or interaction of Time x Limb were observed for either impulse. Overall, peak vertical GRF occurred at a later time point in stance for the involved versus the uninvolved limb. But, significant differences were observed only at 2 and 4 weeks, and were caused by meaningful yet non-significant adaptations bilaterally. Peak vertical GRF of the involved limb occurred later in stance at 2 weeks, and peak vertical GRF of the uninvolved limb occurred earlier in stance at 2 weeks and 4 weeks. Peak vertical GRF limb symmetry index at 2 weeks was significantly lower than preoperatively and at 6 months, and the 4 week peak vertical GRF LSI was significantly lower than at 6 months. Subjects favored the forefoot method of foot strike bilaterally at all data collection points.

Discussion

The primary aim of this study was to determine differences between limbs and across time in the performance of stepping down while walking. As hypothesized, peak vertical GRF was larger and occurred earlier within stance when stepping onto the uninvolved limb. Peak vertical GRF demonstrated the largest effect size. Overall stance time also demonstrated consistent differences and a large effect size. Contrary to my hypothesis, longer stance time occurred onto the uninvolved limb rather than the involved limb. These results were similar to those from Houck and Yack (2003) who reported longer stance times from ACL-deficient subjects than for healthy controls. Due to smaller effect sizes, the two impulses evaluated in this study offered no advantage over peak vertical GRF in identifying asymmetry of this task. In light of these results, future studies should monitor for differences in the amplitude and timing of peak vertical GRF and overall stance time.

In contrast to my hypothesis, differences over time were not significant in the group as a whole or in the APM cohort alone. Rather, within-limb differences were only significant for the ACL cohort when analyzed independently. Results from this analysis indicate that adaptions occur bilaterally, primarily during periods of greatest overall dysfunction. The primary adaptations include lower amplitude and later timing of peak vertical GRF onto the involved limb, higher amplitude and earlier timing of peak vertical GRF onto the uninvolved limb, and longer stance time onto the uninvolved limb.

The second aim of this study was to define the relationships between parameters of ground reaction force and variables of quadriceps performance. The results indicate that normalized quadriceps strength, quadriceps index, and quadriceps atrophy of the leading limb demonstrated the strongest associations with peak vertical GRF and peak vertical GRF limb symmetry index. Subjects with greater asymmetry in quadriceps strength and atrophy also demonstrated greatest asymmetry in stance time and peak vertical GRF amplitude and timing. Although statistically significant, these relationships were lower than hypothesized.

I hypothesized that quadriceps strength and volume of the trailing limb would have a significantly negative relationship with the amplitude of peak vertical GRF for the leading limb. The results indicate that the direction of this relationship was opposite to what I hypothesized. As the quadriceps of the trailing limb became weaker and less symmetrical with the leading limb, a corresponding *decrease* in peak vertical GRF occurred when stepping onto the leading limb. The direction of this relationship appears counter-intuitive. Overall, the relationships between parameters of ground reaction force and quadriceps performance suggest that quadriceps function of the leading limb may be more important than the trailing limb.

Analysis of step down parameters by quartiles based on symmetry of quadriceps strength indicated that subjects with greater asymmetry in quadriceps strength and atrophy also demonstrated greater asymmetry in stance time and peak vertical GRF amplitude and timing. Of note, asymmetry in quadriceps strength only explained a significant portion of asymmetry in peak vertical GRF in subjects within either the highest or lowest quartile. There is a large amount of variability in peak vertical GRF still left to explain. This is especially true in subjects within the middle two quartiles.

The results of self-reported function show a similar trend to that observed with quadriceps strength. I hypothesized that self-reported function of the trailing limb would be weakly and negatively related to the amplitude of peak vertical GRF for the leading limb. This was not the case. Lower self-reported function of the trailing limb was not significantly associated with peak vertical GRF. Rather, lower self-reported function of the *leading* limb was significantly but weakly associated with lower amplitude and symmetry of peak vertical GRF. The relationships between parameters of ground reaction force and self-reported function suggest that perceived disability of the leading limb may be more important than for the trailing limb.

This study leads to several questions of interest pertaining to the mechanisms underlying the results. Why do self-perceived function, quadriceps muscle strength, and quadriceps atrophy appear more important for the leading limb than in the trailing limb while performing this task? One potential explanation may lie in the rate of force development required of the leading and trailing limbs. The rate of force development required by the leading limb during the loading response phase of gait is much faster than for the trailing limb during late midstance and terminal stance. My data indicate that peak vertical ground reaction force occurred in the leading limb approximately 90 to 150 msec after initial contact. Because gait velocity and cadence were not prescribed, many subjects elected to slow their approach before stepping down, in particular when stepping down onto the uninvolved limb. It is possible that slowing down is an adaptation to impairment in the ability to rapidly generate force. Recent research suggests that the ability to rapidly develop force is impaired to a greater extent after knee surgery than MVIC via slow, ramped contraction (Maffiuletti et al., 2010, Knezevic et al., 2014, Angelozzi et al., 2012).

A second explanation may lie in the magnitude of knee extension moment required by the leading and trailing limbs. Because of the large knee flexion angle associated with the trailing limb while stepping down, the demand on the quadriceps was hypothesized to be higher onto the trailing limb than the leading limb. But, this assumption may not be true. While stepping down, ground reaction force from the trailing limb figures to be much lower than for the leading limb during loading response. A review of the literature on knee joint moments during stair descent supports this assertion. Few studies have examined joint moments while stepping down during ongoing gait. Houck and Yack (2003) reported mean internal knee extension moments of 1.71 Nm/kg for healthy subjects for a step height of 20 cm. Two other studies reported mean internal knee extension moments ranging from 1.3 to 2.0 Nm/kg for healthy subjects stepping down from a 10 cm platform (Barbieri et al., 2014, van Dieen et al., 2008). None of these studies reported knee extension moments for the trailing limb, making direct comparison impossible. But, estimates may be derived from studies involving the trailing limb during step-over stair descent. Studies reported peak internal knee extension moments during the second half of stance that range from 1.0 Nm/kg to

1.4 Nm/kg for descent of stairs ranging in height from 15 cm to 25 cm (Spanjaard et al., 2008, Cluff and Robertson, 2011, Beaulieu et al., 2008, Novak and Brouwer, 2011, McFadyen and Winter, 1988). A more detailed analysis of stepping down during ongoing gait with instrumented gait analysis and force platforms under the leading and tailing limbs may help further answer this question.

Another question raised by this study is which factors beyond quadriceps strength may explain changes in step down performance. Several factors are likely involved. First, many subjects after knee surgery demonstrate kinesiophobia, a fear of movement and/or re-injury that has shown to be a significant obstacle to a full return to pre-injury level of activity (Chmielewski et al., 2008, Flanigan et al., 2013). Such a lack of confidence in the involved limb may help to explain adaptions in performance when stepping onto the involved limb. In this study, I did not administer a patient-reported outcome designed to measure fear of movement and re-injury like the Tampa Scale of Kinesiophobia-11 (Woby et al., 2005). Future research should assess this factor.

Second, adaptations in gait mechanics may also explain changes in step down performance. The hallmarks of these adaptations appear in the sagittal plane during the loading response phase of gait, where a “stiffening” strategy of the knee is frequently observed. In this strategy, the involved limb demonstrates reductions in peak knee flexion angle and peak internal knee extension moment (Berchuck et al., 1990, Noyes et al., 1992, Wexler et al., 1998, DeVita et al., 1998, Webster et al., 2005, Hurd and Snyder-Mackler, 2007, Hall et al., 2012). Houck and Yack (2003) reported similar adaptations in ACL-deficient subjects while stepping down during ongoing gait.

In addition to quadriceps weakness, altered neuromuscular control strategy is believed to play a major role in mechanical gait adaptations. Increased co-contraction of the hamstrings helps transfer the method of support from the knee to the hip (Roberts et al., 1999, Alkjaer et al., 2003, Rudolph et al., 2001). The importance of this neuromuscular control strategy may be illustrated by the findings of Roewer et al (Roewer et al., 2011), where gait abnormalities persisted up to 2 years in ACL-reconstructed subjects with high quadriceps strength. Although not evaluated in this study, such alterations in neuromuscular control may help explain additional variability while stepping down. Future research investigating mechanisms behind asymmetry while stepping down should consider including instrumented gait analysis with EMG on key lower extremity muscle groups. Finally, the platform height used in this study (25.4 cm) is higher than typical stairs and curbs, and also higher than those used in previous studies where subjects were evaluated while walking and stepping down. This fact may also have influenced the results. In the future, it may be desirable to standardize the height of the step near 20 cm. This is similar to the height that people encounter daily with stairs and curbs.

One of the aims of this study was to understand how people choose to perform the walk and step down task after ACL and meniscus injury and subsequent surgery. Subjects were allowed to self-select their walking speed and method of foot strike. Allowing subjects this freedom likely produced data that was more representative of the approach they use in daily life than if those factors were controlled. But, allowing subjects to determine velocity, cadence, and method of foot strike may have attenuated differences between legs and over time. Such attenuation may have altered the relationship between parameters of ground reaction force and quadriceps performance. Despite this likely attenuation, I observed significant associations indicating that the walk and step down task may be a simple functional test with high clinical value. It is recommended that future studies control the method of foot strike and the velocity or cadence of gait in order to elicit differences between limbs or time points.

This study investigated the walk and step down task in two cohorts of surgical patients, APM and ACLR. These cohorts were selected because they typically fall on different parts along a continuum of surgical severity and neuromuscular involvement. Although these results are specific to these populations, they likely have implication for other groups of patients who may present with similar dysfunction.

The decision to include these two cohorts led to an interesting finding. Subjects in both groups presented similar at the conclusion of monitoring. Asymmetry in stance time and the amplitude and timing of peak vertical GRF were similar in APM subjects at 5 weeks and ALCR subjects at 6 months postoperatively. This makes sense, but has not been reported to my knowledge. Despite the low severity of trauma to the joint with APM, these subjects still demonstrated significant asymmetry while stepping down during gait. This finding further supports the value of this task. Asymmetry was most pronounced in ACL subjects early after surgery when neuromuscular impairments were highest. This supports the construct validity of the walk and step down task. Changes over time in this group support the responsiveness of this task. I expect that the walk and step down task will be of similar benefit to other clinical populations that show large impairments in mechanics and strength. This includes patients after total knee arthroplasty (Mandeville et al., 2007, Yoshida et al., 2012) and the elderly at high risk for falls (Skelton et al., 2002, Laroche et al., 2012). The simplicity and relevance across the lifespan are key advantages of this task over other measures that are more complicated or less relevant to daily life.

A major aim of this research was to investigate a simple and clinically feasible test with the potential for providing meaningful information about patient outcome. Part of feasibility relates to the equipment necessary to perform the test. This study used a force platform similar to those in many academic and clinical sites that perform research. This study enables clinicians and researchers at these locations to utilize force platforms without motion capture to obtain meaningful information about functional limb symmetry. Force platforms can provide other useful data in rehabilitation but can be expensive and therefore are not routine equipment in many clinics. But, it may be a reasonable investment. The cost of a portable, research-grade force platform rivals that of some exercise equipment and is significantly less expensive than a device like an isokinetic dynamometer. Single or dual-axis force platforms designed for education and basic applications are also available at lower costs and may be sufficient for this task. Wearable sensors including accelerometers also hold promise for use in functional testing in clinical settings.

Conclusion

This study investigated the walk and step down task, a simple and highly relevant functional outcome applicable to people throughout the lifespan. This study found significant asymmetry in cohorts of subjects with meniscus and ACL injury and subsequent surgery. Asymmetries were greater in those with higher neuromuscular impairment. The effect sizes of these adaptations were large, in particular for peak vertical ground reaction force. I expect the relationship between ground reaction force and neuromuscular function to be enhanced by controlling for gait parameters when performing this task. This initial analysis suggests that stepping down while walking may be a simple functional test applicable in clinical settings and in large multi-center trials where motion capture is not available.

CHAPTER 4

MEASUREMENT OF SINGLE LEG VERTICAL HOP HEIGHT FROM FLIGHT TIME USING WEARABLE ACCELEROMETERS

Introduction

Vertical hop testing is a popular functional performance-based measure for healthy athletes and subjects nearing return to full activity after injury and subsequent rehabilitation. Within orthopaedic and sports rehabilitation, several groups of researchers have advocated for the use of hop testing to help quantify patient outcome and determine readiness to return to full activity (Gustavsson et al., 2006, Noyes et al., 1991, Fitzgerald et al., 2000, Grindem et al., 2011, Logerstedt et al., 2012). The single leg vertical hop possesses high test-retest reliability (Gustavsson et al., 2006) and is also among the most sensitive tests for detecting between-limb differences in performance after ACL reconstruction (Petschnig et al., 1998, Thomee et al., 2012). Differences in single leg hopping for height are common after rehabilitation from anterior cruciate ligament (ACL) reconstruction is typically complete (Petschnig et al., 1998, de Fontenay et al., 2014, Ernst et al., 2000, Thomee et al., 2012). Because they are novel tasks, hop testing for distance and hop testing requiring repeated hops lacks functional relevance for many people. This is especially true for tests that require consecutive hops with or without a cutting component, such as the triple hop for distance, the triple cross-over hop, the 6 meter timed hop (Noyes et al., 1991), and the side hop (Ageberg et al., 2008). In contrast, most athletes are accustomed to taking off from one leg for maximal height. Thus, the vertical hop may have greater practical value when compared to horizontal hopping, especially when consecutive hops are required when hopping for distance or time.

Many methods exist for scoring hop height, but force platforms are considered the gold standard (Leard et al., 2007, Castagna et al., 2013, Casartelli et al., 2010). Although highly reliable and accurate, these systems have limited use in clinical or field-based measurement because of high cost and lack of portability. Several field-based methods exist for calculating jump or hop height. The jump and reach method requires that subjects reach as high overhead as possible during flight and manually move as many swiveling plastic slats as possible on a device like a Vertec (Sports Imports, Hilliard OH). Other techniques measure the time of flight and then calculate hop height based on a simple equation. For example, contact mats identify flight time via force or pressure sensors, and optical timing systems use the interruption of beams from photoelectric cells placed just off the ground to estimate flight times. Contact mats and photoelectric cells compare favorably against force platforms (Leard et al., 2007, Castagna et al., 2013, Glatthorn et al., 2011). Although contact mats and photoelectric cells also may be used to time events, these devices have few other functions.

Wireless accelerometer sensors secured to the shank or trunk have also been used to determine flight time and calculate jump height from a double-leg take-off and landing. These small, wireless accelerometers have recently undergone rapid improvement in technology and cost, and are frequently used to objectively assess physical activity and movement. Because of the low relative cost of these sensors and their capability to provide other meaningful measurements of movement and performance, this method is particularly attractive.

Previous research investigating the use of accelerometers to measure jump height shows promise, but has limitations. Several studies report moderate to high agreement between jump height measured by accelerometers and force platforms (Picerno et al., 2011, Castagna et al., 2013, Choukou et al., 2014). But, systematic bias has also been reported between accelerometer-based methods and criterion standards, with accelerometers generally over-estimating hop height (Casartelli et al., 2010, Choukou et al., 2014, Castagna et al., 2013). Other studies report very high correlation between jump heights calculated from accelerometers mounted on the shank and force platforms (Elvin et al., 2007, Palma, 2008). But because several of these do not report absolute agreement, systematic error between methods is unclear (Elvin et al., 2007, Quagliarella et al., 2010). The methods of scoring reported in these studies were variable and in some cases not well defined, especially for determining the moment of take-off (Picerno et al., 2011, Castagna et al., 2013). In addition, subjects were frequently restricted from swinging their arms during the vertical jump (Castagna et al., 2013, Picerno et al., 2011, Casartelli et al., 2010). This practice may simplify the acceleration signal and data processing, but it also decreases the functional relevance of such jumps and may limit subjects' maximal effort by introducing a novel movement strategy. In addition, these studies focused on determining the height from two-legged jumps without including single leg hops. Because the amplitude of height between these two tasks differs, relative measurement error may differ too. These studies relied heavily on samples of healthy subjects. None have included subjects with knee pathology.

The primary purpose of this study is to determine the accuracy and consistency of wearable accelerometer sensors to estimate single leg vertical hop height in healthy people and ACL-reconstructed individuals. The secondary purpose of this study is to determine the relationships between hop height and neuromuscular performance of the quadriceps and hamstrings muscles.

Methods

Subjects: Twenty-four subjects participated in this study approved by University of Iowa's Institutional Review Board. Twelve subjects (six male, six female) had previously undergone isolated anterior cruciate ligament (ACL) reconstruction 6 to 19 months earlier. Twelve subjects without a history of knee injury who were age and gender-matched with the ACL-reconstructed subjects served as control subjects. The control group permitted comparison of the accuracy of scoring hop height from accelerometers between healthy subjects and subjects with previous injury. Male and non-pregnant female subjects were eligible for inclusion if they were between the ages of 18 and 35 years of age with a BMI no greater than 35 kg/m^2. ACL-reconstructed subjects were excluded if they had multiligamentous injury, concomitant meniscal repair, demonstrated a grossly symmetrical gait (e.g., no noticeable limp), or were unable to flex their knee past 120 degrees. The time period of six months to two years postoperatively was selected because this is a time point frequently cited for return to full activity (Kvist, 2004) and rehabilitation is typically complete by this point. Previous research has demonstrated that subjects within this time frame after ACL reconstruction frequently demonstrate alterations in lower extremity muscle strength, mechanics, and functional performance (Roewer et al., 2011, Thomee et al., 2012, Ernst et al., 2000). The 18 year old boundary increased the likelihood of skeletal maturity, and the 35 year old boundary increased the homogeneity in baseline physical characteristics.

<u>*Single leg vertical hop*</u>**:** Subjects hopped from and landed onto a rectangular force platform with dimensions of 40 cm x 60 cm (Model 9865B, Kistler Instrument Corp., Winterthur, Switzerland). Force platform data was collected at a sampling frequency of 360 Hz and filtered at 50 Hz using a fourth order Butterworth low-pass filter. To allow for an approach typical of functional activity, subjects were permitted to use their natural hopping technique without restricting countermovement, trunk flexion, or arm swing (Figure 4.1). After 3 to 5 familiarization trials per side, subjects performed 5 hopping trials on each limb with the uninvolved leg collected first for ACL-reconstructed subjects.

For a valid trial the foot was required to be entirely on the force platform for take-off and landing. Data from the force platform served as the criterion measure for calculating flight time. Ground reaction force values registering zero (± 5 N for noise) and indicating no contact with the force platform were defined as time in flight. Hop height was calculated from the time of flight via the following equation (Linthorne, 2001): $Height_{hop} = (9.81\, m/s^2 \times t_{flight}{}^2)/8.$

Figure 4.1: Example of a single leg hop trial

Accelerometry: Subjects wore two Shimmer 3 accelerometers (± 16 g, Shimmer Sensing, Dublin, Ireland; Figure 4.2) on the same side as the hopping limb with a collection frequency of 256 Hz. One accelerometer was affixed to the waist at the mid-axillary line just inferior to the iliac crest, and the second accelerometer was secured to the proximal lateral shank just distal to the fibular neck. Placement at these locations permitted investigation into the reliability and validity of scoring at these anatomic sites. Firm-fitting elastic belts secured the accelerometers around the pelvis and shank with the accelerometers aligned as close as possible to the global coordinate system axes while the subjects were standing. This method produces adequate belt tension to reduce noise from poor fixation (Mizrahi et al., 2000). This method of fixation did not interfere with motion capture marker placement, is feasible in a clinical setting, and has been used by others in biomechanical research (Rowlands and Stiles, 2012, Rowlands et al., 2013).

Figure 4.2: Shimmer 3 accelerometer

Analysis of the frequency spectrum of take-off and landing events was performed via Fast Fourier Transformation. Based on this analysis, accelerometer data was filtered at 50 Hz using a fourth order Butterworth low-pass filter. This cutoff frequency was necessary to preserve the acceleration signal from impact, which was routinely detected at over 40 Hz. This approach reduced the potential for contamination from high-frequency noise (Zijlstra and Hof, 2003, Fong and Chan, 2010). The vertical component of acceleration is the largest component among the three orthogonal planes and is of primary interest in determining flight time.

Time of flight was scored manually from accelerometer data using the following method developed during pilot testing: Take-off from the waist and shank accelerometers were defined by the first major positive deflection following the large upward acceleration signal (Figure 4.3). Landing from the waist and shank accelerometers was defined by the deflection point of the first positive acceleration to exceed 10 m/s^2 in amplitude. Hop height was calculated from the time of flight via the same equation used to score force platform data. A random number generator was used to select thirty trials (15 waist, 15 shank) from both groups (60 trials total) to assess for the reliability of the measurement technique. The rater completed a blinded second assessment of these trials to assess for intra-rater reliability. A second rater completed a blinded assessment of these trials to determine inter-rater reliability.

Neuromuscular testing: For neuromuscular strength testing, subjects sat on a HUMAC NORM Testing and Rehabilitation System (CSMI, Stoughton, MA) with the hips and knees flexed to 85 degrees and 60 degrees, respectively. A seat belt, chest straps, and thigh strap secured subjects to the chair (Figure 4.4). This approach optimizes the length-tension relationship of the quadriceps and permits insight into side-to-side ratios of both quadriceps and hamstring muscle groups (Krishnan and Williams, 2014). Force data was collected with a custom apparatus consisting of a modified shin guard and load cell affixed to the rigid arm of the HUMAC NORM. The shin guard was attached to the shank with a Velcro strap approximately 5 cm proximal to the medial malleolus. The load cell (model LPU-500, Transducer Techniques, Temecula, CA) was placed in series between the shin guard and rigid arm. The same testing configuration was used bilaterally. Subjects performed a minimum of 3 maximal voluntary isometric contractions

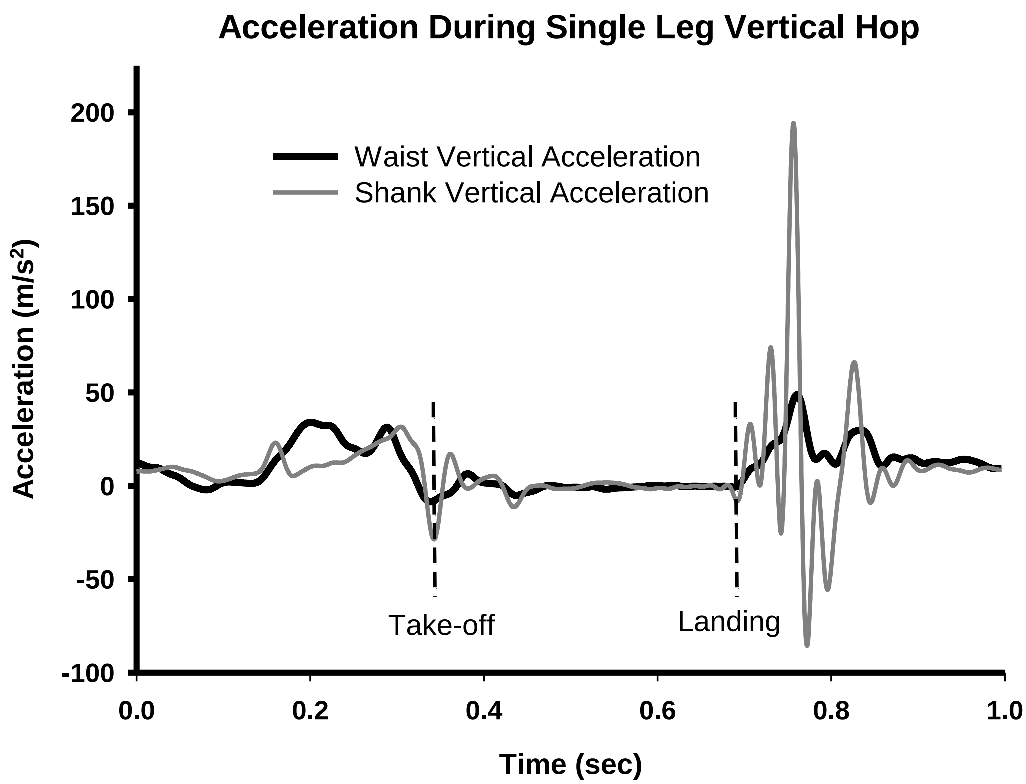


Figure 4.3: Sample data over one second from a hop trial with acceleration measured at the waist and shank

(MVICs) of 5 seconds duration with two minutes of rest in between trials for knee extension and knee flexion. Familiarization trials at 50%, 75%, and 90% of perceived effort, loud verbal encouragement, and real-time feedback of force development were provided to encourage maximal effort. Subjects were required to perform a minimum of two trials within ± 5% of each other to ensure reliability of the data. In addition to MVIC contractions, five trials of rapid knee extension and knee flexion were collected with the instructions to kick out or pull back *as fast* and as hard as possible. Data from these trials was collected to permit secondary analysis of how performance in single leg hopping relates not only to quadriceps and hamstrings strength, but also the maximal rate of force development. Data was sampled at 1000 Hz, collected with an AD Instruments PowerLab 16/30, and processed via LabChart 8 software (AD Instruments, Bella Vista, Australia).

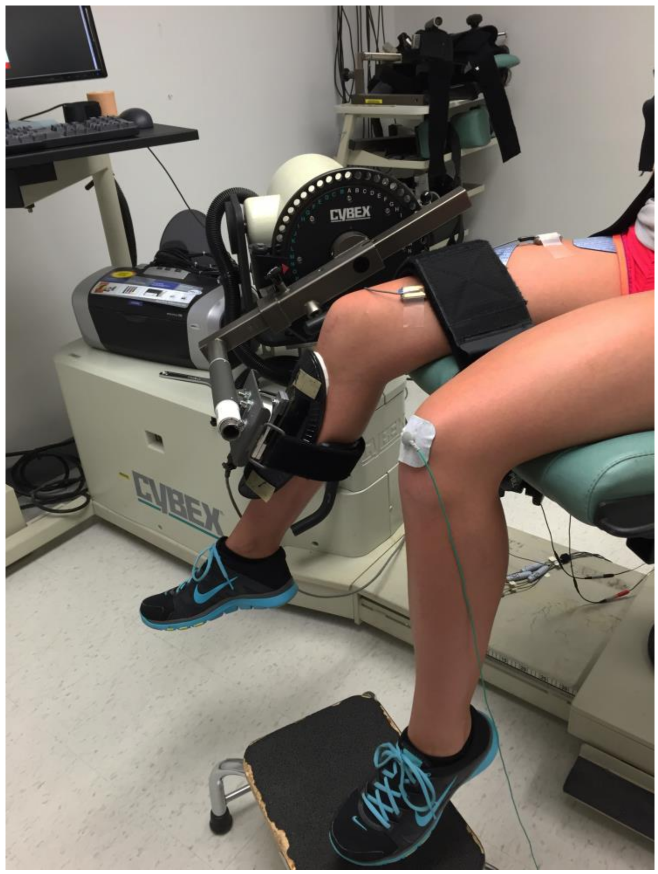


Figure 4.4: Testing configuration for quadriceps neuromuscular performance

<u>Patient-reported outcomes</u>: Patient-reported measures assessed pain and self-perceived function. A visual analog scale assessed pain on a 100 mm scale during a functional activity (stepping down while walking). The International Knee Documentation Committee (IKDC) subjective knee evaluation form assessed knee symptoms and self-

perceived function during sport and strenuous activity. The IKDC subjective form is highly reliable and shows high construct validity in subjects with ACL reconstruction (Irrgang et al., 2001, van Meer et al., 2013). The Global Rating of Knee Function (also known as the SANE rating) provided a single numerical score between 0 (complete disability) and 100 (no disability) in response to the question “On a scale of 0 to 100, how would you rate the function of your knee (with 100 being complete function)?”(Williams et al., 2000). The Tampa Scale of Kinesiophobia-11 (Woby et al., 2005) measured fear of activity and re-injury, which has been shown to be a significant obstacle to full return to pre-injury activity level (Chmielewski et al., 2008, Flanigan et al., 2013). Similarly, the ACL Return to Sport after Injury (ACL-RSI) measured psychosocial response to returning to sport after ACL reconstruction. This has shown good reliability and ability to help predict return to sport after ACL reconstruction (Webster et al., 2008, Langford et al., 2009, Muller et al., 2014). The Marx activity scale measured participation in athletic activities and has shown high reliability (Marx et al., 2001). These measures of self-reported pain, perceived function, and psychosocial response to activity allow for a comprehensive characterization of the subject pool and permit secondary analysis into the relationship between perceived function, fear avoidance, and hop performance measured in this study.

Data Analysis

Group baseline characteristics and patient-reported outcomes: Descriptive statistics were calculated on continuous variables and the results of dichotomous outcomes summarized. Independent t-tests with the level of significance set to 0.05 were used to evaluate for differences in characteristics between ACL-reconstructed and control subjects.

Reliability and validity: A two-way random, absolute agreement, single measures intraclass correlation coefficient [ICC (2,1)] was used to determine intra-rater and inter-rater reliability for manually scoring hop height from accelerometry. A two-way mixed, absolute agreement, single measures intraclass correlation coefficient [ICC(3,1)] was used to determine the concurrent validity of using the acceleration scoring method to calculate hop height versus the criterion standard of hop height calculated from force platform data.

Error: Bland-Altman plots were developed to determine systematic error and 95% limits of agreement between techniques. In order to understand discrepancy between accelerometer methods and the force platform, Measurement Error was computed from the standard deviation of the difference scores ($ME = SD\ of\ differences/\sqrt{2}$). These error terms were then used to compute 95% error ranges (Error x 1.96) and Minimal Detectable Changes ($MDC = Error \times 1.96\ \times\ \sqrt{2}$) for each scoring method. In addition, coefficients of variation were computed for hop heights across the range observed in this study using the following formula: $CV = 2\ x\ ME/(Mean1 + Mean2)$.

Differences between limbs and groups: Separate 2 x 3 (Limb x Method) Analyses of Variance (ANOVA) were performed to assess for differences in hop heights within the ACL-reconstructed and Control groups based on the limb and the method of scoring hop height (waist accelerometer, shank accelerometer, or force platform). Paired t-tests with the level of significance set to 0.05 were selected to evaluate for between-limb differences in each group for variables describing neuromuscular function. Limb symmetry indices (LSIs) were calculated for the ACL-reconstructed group by dividing the result from the involved leg by the result from the uninvolved leg. For control subjects, LSIs were calculated by dividing the result for the nondominant limb by the result for the dominant limb. A 2 x 3 (Group x Method) repeated measures analysis of variance with the level of significance set to 0.05 was performed to assess for differences in LSI between groups and the three methods of scoring hop height. Independent t-tests with the level of significance set to 0.05 were used to assess for differences in limb symmetry between groups for variables obtained from neuromuscular testing. In the presence of a main effect, Bonferroni-corrected pairwise comparisons were employed as post hoc tests for all ANOVAs. Between-limb effect sizes for the ACL-reconstructed group were computed via Cohen's *d* for hop height and significant variables of neuromuscular function. When interpreting Cohen's *d*, a value of 0.2 is considered a small effect size, 0.5 is considered a medium effect size, and 0.8 considered a large effect size (Cohen, 1992).

Relationships between hop height and neuromuscular performance: Pearson's Product Moment Correlations were used to determine associations between hop heights calculated via each method (waist accelerometer, shank accelerometer, and force plate), quadriceps and hamstrings muscle strength and rate of force development, and patient-reported outcomes. Pearson's Product Moment Correlations were also computed to explore associations between LSIs in hop height calculated via all three methods, patient-reported outcomes, and LSIs for variables of neuromuscular. When interpreting Pearson's Product Moment Correlations, values between 0.25 to 0.50 were considered fair, values between 0.50 and 0.75 were considered moderate-to-good, and values over 0.75 were considered good-to-excellent (Portney, 1993). SPSS version 23 software was used to perform all statistical analysis (SPSS Inc, Chicago, IL).

Results

Group baseline characteristics and patient-reported outcomes: ACL-reconstructed subjects and control subjects were similar in age, height, and self-reported activity level. ACL-reconstructed subjects demonstrated significantly higher body mass and BMI when compared to control subjects. ACL-reconstructed subjects scored significantly lower than control subjects on self-reported knee function, readiness to return to sport, pain, and global knee rating (Table 4.1).

Reliability and validity: Intra-rater and inter-rater reliability for estimating hop height from waist and shank-mounted accelerometers were excellent, and were similar for control subjects and for ACL-reconstructed limbs (Tables 4.2 and 4.3). This indicates a high degree of reliability using the scoring method at either anatomical site. For intra-rater reliability, tighter confidence intervals were observed for measuring at the waist than at the shank. Tighter confidence intervals were observed for inter-rater reliability than for intra-rater reliability.

	Control	**ACL-Reconstructed**
Age (years)	24.58 ± 2.68	26.25 ± 5.41
Sex	6 M, 6 F	6 M, 6 F
Time since surgery (months)	n/a	10.33 ± 3.4 (range 6-19)
Mass (kg)	66.03 ± 13.55	80.98 ± 11.67 *
Height (m)	1.69 ± 0.073	1.74 ± 0.071
BMI (kg/m^2)	22.82 ± 3.43	26.58 ± 3.29 *
ACL-RSI (%)	97.29 ± 4.39	56.88 ± 23.1 *
IKDC (%)	98.65 ± 1.83	82.47 ± 11.42 *
Tampa Scale for Kinesiophobia-11	16.33 ± 3.28	19.58 ± 4.56
Pain Visual Analog Scale (mm)	0	2.50 ± 3.94 *
Marx Activity	11.67 ± 4.10	10.92 ± 3.63
Global Knee Rating	98.92 ± 2.87	87.08 ± 9.40 *

Table 4.1: Subject characteristics and self-reported outcomes

*P < 0.05 from Control group

		ICC (2,1)	95% CI	Sig
All subjects	Waist	0.968	0.935, 0.985	*P* < 0.001
	Shank	0.944	0.888, 0.973	*P* < 0.001
Control subjects	Waist	0.993	0.979, 0.998	*P* < 0.001
	Shank	0.958	0.883, 0.986	*P* < 0.001
Reconstructed limb	Waist	0.937	0.828, 0.978	*P* < 0.001
	Shank	0.916	0.750, 0.972	*P* < 0.001

Table 4.2: Intra-rater reliability of calculating hop height from accelerometers

		ICC (2,1)	95% CI	Sig
All subjects	Waist	0.988	0.974, 0.994	*P* < 0.001
	Shank	0.977	0.953, 0.989	*P* < 0.001
Control subjects	Waist	0.976	0.930, 0.992	*P* < 0.001
	Shank	0.991	0.973, 0.997	*P* < 0.001
Reconstructed limb	Waist	0.987	0.962, 0.996	*P* < 0.001
	Shank	0.958	0.934, 0.993	*P* < 0.001

Table 4.3: Inter-rater reliability of calculating hop height from accelerometers

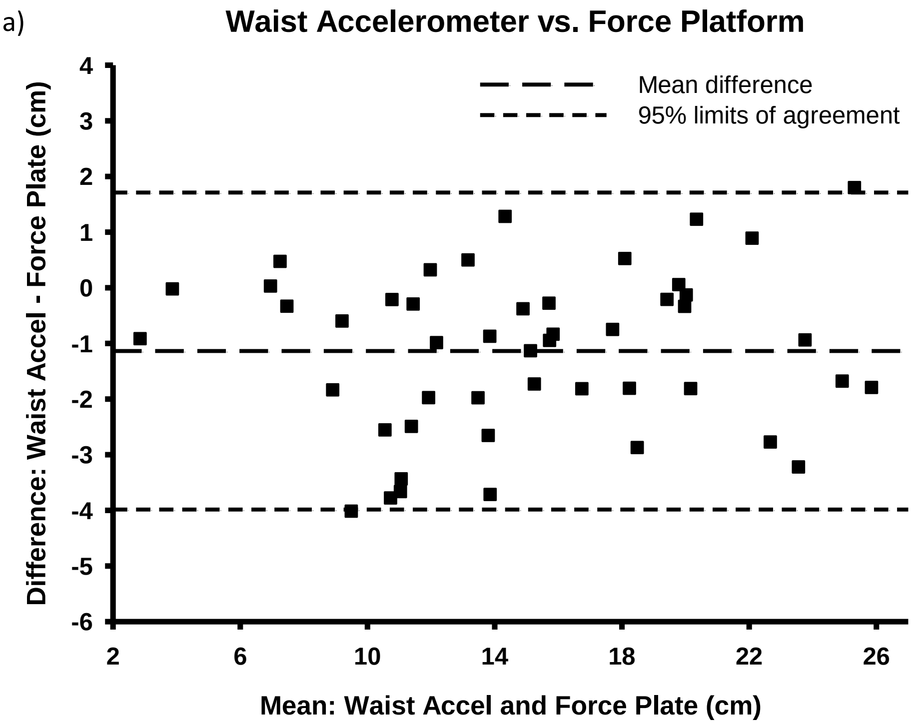
		ICC (3,1)	95% CI	Sig
All subjects	Waist	0.948	0.792, 0.980	*P* < 0.001
	Shank	0.978	0.961, 0.988	*P* < 0.001
Control subjects	Waist	0.933	0.654, 0.979	*P* < 0.001
	Shank	0.965	0.923, 0.985	*P* < 0.001
Reconstructed limb	Waist	0.974	0.825, 0.994	*P* < 0.001
	Shank	0.981	0.937, 0.994	*P* < 0.001

Table 4.4: Concurrent validity of calculating hop height from accelerometers secured at the waist and shank versus a force platform

Concurrent validity for estimating hop height from accelerometers compared to the criterion standard of a force platform was computed using the mean of five trials for each limb (n = 48; Table 4.4). This analysis showed that concurrent validity was high for waist-mounted and shank-mounted accelerometers, indicating a high degree of validity using accelerometers at either site versus the force platform. Confidence intervals were tighter at the shank than for the waist, indicating less variability with this technique when compared to the criterion standard of the force platform. Concurrent validity was similar for ACL-reconstructed subjects and healthy control subjects.

Error: Bland-Altman plots were developed in order to assess for systematic error between measuring hop height with accelerometers and the force platform (Figure 4.5). The Bland-Altman plot for the waist accelerometer method showed a mean difference just below -1 cm, indicating that the waist accelerometer method tended to systematically underestimate hop height by approximately 1 cm when compared to the force platform. Conversely, the plot for the shank accelerometer method showed a mean difference very close to zero, indicating virtually no systematic error with this method. The shank accelerometer method contained 95% limits of agreement that were slightly less than for the waist accelerometer (4.64 cm versus 5.70 cm, respectively), indicating less random error with the shank accelerometer method than for the waist method. Differences in the Waist vs Force Platform and the Shank vs. Force Platform plots demonstrated equal variance across the range of mean hop values (P = 0.72 and P = 0.48, respectively, with Levene's test), indicating that errors in scoring were constant across the range of hop heights.

Measurement Error, 95% Error Range, and Minimal Detectable Change were all less for estimating hop height with accelerometers at the shank versus the waist (Table 4.5). Error Range indicates that the average of all possible measurements is within a a)



b)

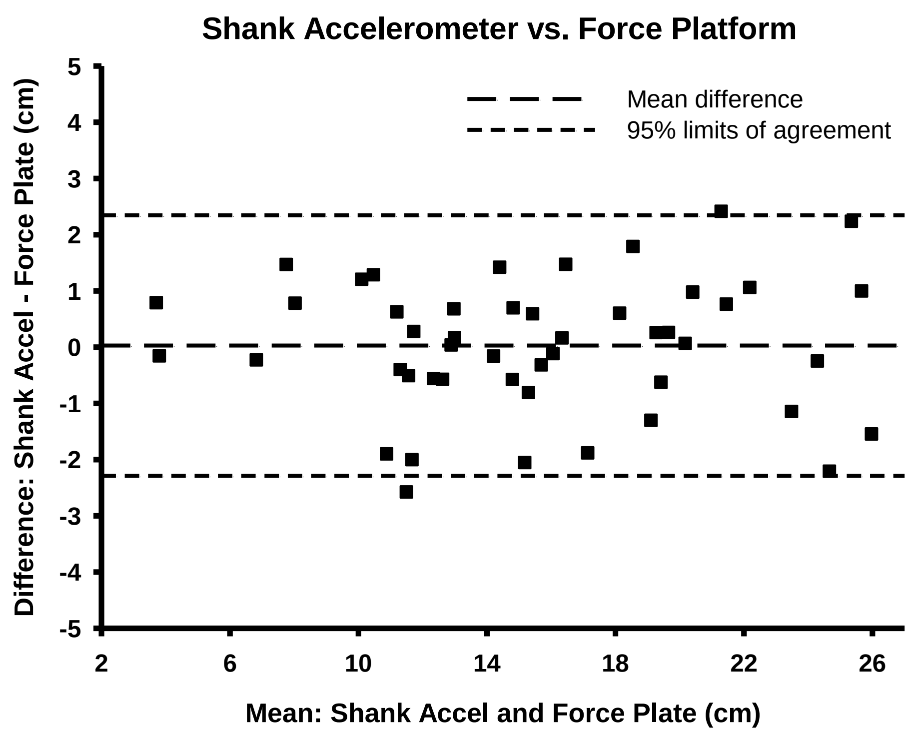


Figure 4.5: Bland-Altman plots for differences in hop height measured with the force platform and a) Waist accelerometer or b) Shank accelerometer

certain range above and below the actual measurement taken. Thus, it is a confidence interval about a single measurement (which in this study was the mean of 5 trials). Minimal Detectable Change provides an estimate of the ability of the instrument to detect meaningful differences between two measurements (i.e., between two means of five trials each). Coefficients of variation for each method ranged between about 20% at the lower end of observed hop heights and 4% at the upper end (Table 4.6).

	Waist Accelerometer	Shank Accelerometer
Measurement Error (cm)	1.03	0.84
95% Error Range (cm)	2.01	1.64
Minimal Detectable Change (cm)	2.85	2.32

Table 4.5: Error associated with using accelerometers to estimate single leg vertical hop height

Hop Height	Waist Accelerometer	Shank Accelerometer
5 cm	20.6%	16.8%
10 cm	10.3%	8.4%
15 cm	6.9%	5.6%
20 cm	5.2%	4.2%
25 cm	4.1%	3.4%

Table 4.6: Coefficients of Variation for sample hop heights

Differences between limbs and groups: Evaluation for differences in hop height by Limb and Method revealed a significant main effect for the method of scoring hop height in the ACLR group and control group ($P = 0.007$ and 0.001, respectively; Figure 4.6). In both groups, waist accelerometers underestimated hop height when compared to the shank accelerometer and force platform. No difference was detected between scoring with the shank accelerometer and force platform ($P = 0.284$). The involved limb of the ACL-reconstructed group demonstrated a significantly lower hop height than the uninvolved limb ($P < 0.001$), but no difference was detected between the dominant and non-dominant limbs of the control group ($P = 0.588$). Significant interaction between Limb and Method was present in the ACL-reconstructed group ($P = 0.035$). In the uninvolved limb, the shank method over-estimated hop height. This same trend was not apparent for the involved limb.

Evaluation for between-limb differences in quadriceps and hamstrings function in the ACL-reconstructed group revealed significantly weaker quadriceps MVIC and slower MRFD in the involved limb compared to the uninvolved limb ($P = 0.003$ and $P = 0.002$, respectively). The involved limb also demonstrated significantly weaker hamstrings MVIC than the uninvolved limb ($P = 0.028$). The control group did not demonstrate between-limb differences for any measure of quadriceps or hamstring function.

When differences in LSI for hopping based on the subject group and method of scoring were evaluated, a significant main effect was present for Group ($P < 0.001$; Figure 4.7). The ACL-reconstructed group demonstrated significantly less symmetry than the control group. Bonferroni-corrected post-hoc testing revealed this difference between groups was consistent for scoring hop height from the waist accelerometer, shank accelerometer, or force platform ($P < 0.0023$). No significant differences in hop symmetry existed between scoring hop height from the waist accelerometer, shank

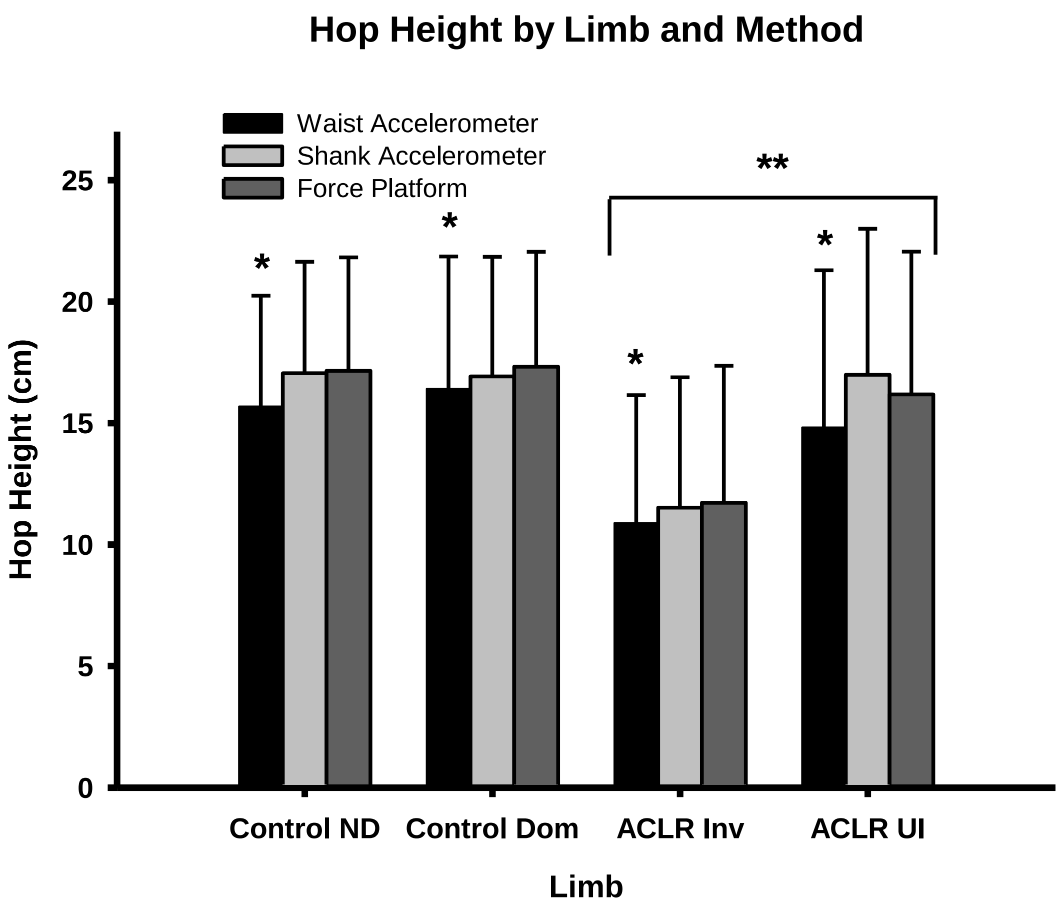


Figure 4.6: Mean hop height by limb and method for Control and ACL-reconstructed subjects.

* *P* < 0.05 for Waist Accelerometer when compared to Shank Accelerometer and Force Platform

***P* < 0.05 for the ACL-reconstructed limb when compared to the uninvolved limb

ND = Non-dominant limb   Dom = Dominant limb

Inv = Involved limb   UI = Uninvolved limb

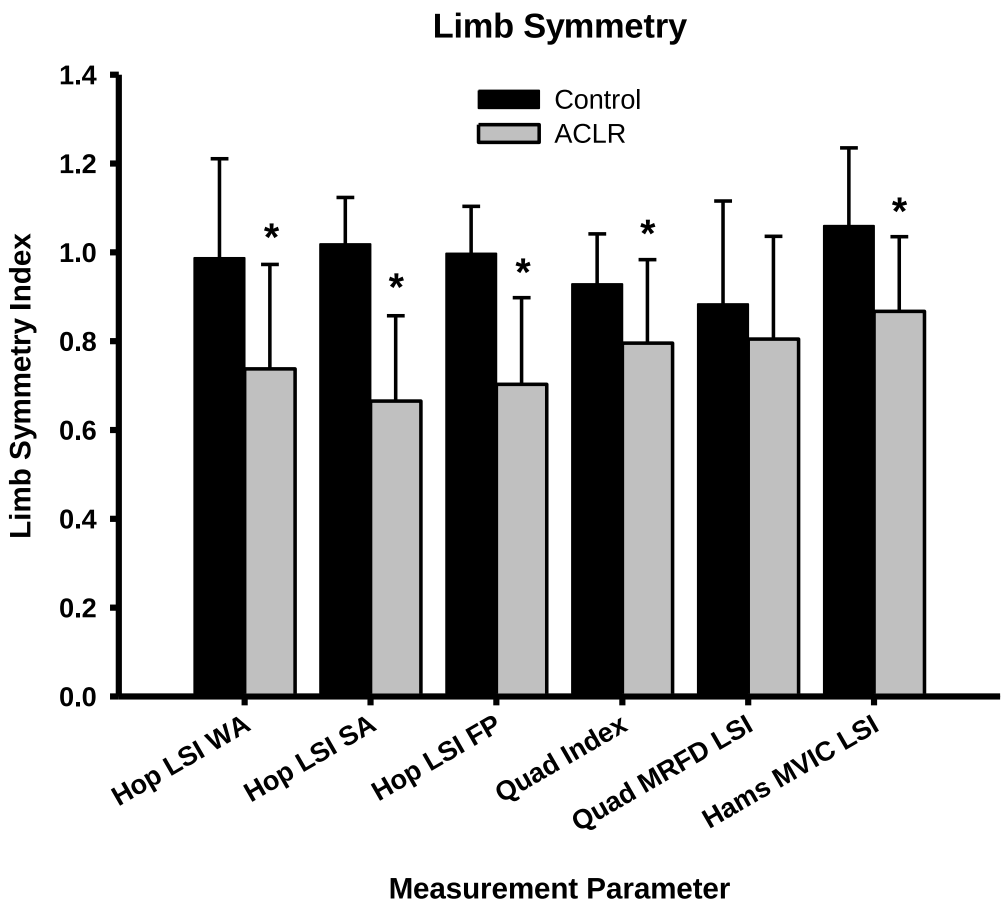


Figure 4.7: Limb Symmetry Indices (LSIs) for hopping and tests of neuromuscular function for the quadriceps and hamstrings

* $P < 0.05$ for the ACL-reconstructed group when compared to Control group

WA = Waist Accelerometer SA = Shank Accelerometer

FP = Force Platform

MRFD = Maximal Rate of Force Development accelerometer, or force platform ($P = 0.665$). Significant differences in LSI between subject groups were also observed for quadriceps index ($P = 0.047$) and hamstring MVIC ($P = 0.013$).

Effect sizes for between-limb differences in hop height in the ACL-reconstructed group were large for all three methods, and exceeded the effect sizes seen for variables of quadriceps and hamstring performance (Table 4.7). Although the effect size was large for the waist accelerometer method ($d = 1.18$), the shank and force platform methods demonstrated even larger effect sizes ($d = 1.69$ and $d = 1.67$, respectively). For these methods, the lower bounds of 95% confidence intervals exceeded an effect size of 1.0, indicating high confidence in the large effect size.

	Effect size (Cohen's *d*)	**95% CI**
Hop WA	1.18	0.54, 1.81
Hop SA	1.69	1.05, 2.32
Hop FP	1.67	1.03, 2.30
Quad MVIC	1.12	0.48, 1.75
Quad MRFD	1.13	0.49, 1.76
Hams MVIC	0.73	0.10, 1.37

Table 4.7: Between-limb effect sizes for hopping and thigh muscle function in ACL-reconstructed subjects (Cohen's *d*)

WA = Waist Accelerometer SA = Shank Accelerometer

FP = Force Platform MRFD = Maximal Rate of Force Development

Relationships between hop height and neuromuscular performance: The strengths of association between hop height and measures of thigh muscle function in all subjects were moderate to-good for all methods of scoring hop height (Table 4.8). Stronger associations between hop height, MVIC, and maximal rate of force development were observed for the quadriceps (range r = 0.617 to 0.675) than for the hamstrings (range r = 0.433 to 0.530) for all hop scoring methods. Hop height was more strongly associated with quadriceps maximal rate of force development than quadriceps MVIC.

Relationships between hop height and thigh muscle function were higher in the ACL-reconstructed group than with all subjects analyzed together (Table 4.9). For ACL-reconstructed subjects, the association between hop height and quadriceps MVIC and maximum rate of force development exceeded 0.80, indicating good to excellent strength of relationship. Associations between hop height and hamstrings MVIC and maximal rate of force development approached or exceeded 0.70.

	Waist Accelerometer	**Shank Accelerometer**	**Force Platform**
Quadriceps MVIC	0.618 †	0.645 †	0.617 †
Quadriceps Normalized MVIC	0.489 †	0.497 †	0.527 †
Quadriceps MRFD	0.675 †	0.635 †	0.656 †
Hamstrings MVIC	0.530 †	0.523 †	0.504 †
Hamstrings MRFD	0.491 †	0.440 †	0.433 †

Table 4.8: Pearson's Product Moment Correlations between 3 methods of calculating hop height and thigh muscle function for all subjects (n = 48 limbs)

$^{\dagger}P < 0.01$

	Waist Accelerometer	Shank Accelerometer	Force Platform
Quad MVIC	0.812 †	0.803 †	0.833 †
Quad Normalized MVIC	0.692 †	0.716 †	0.742 †
Quad MRFD	0.826 †	0.787 †	0.803 †
Hamstrings MVIC	0.671 †	0.628 †	0.634 †
Hamstrings MRFD	0.730 †	0.637 †	0.658 †

Table 4.9: Pearson's Product Moment Correlations between 3 methods of calculating hop height and thigh muscle function for ACL-reconstructed subjects (n =24 limbs)

†$P < 0.01$

Limb symmetry indices for hop height were significantly and strongly associated with several self-reported outcomes when all subjects were analyzed together. This was particularly true for values scored via the shank accelerometer or force platform (Table 4.10). The relationship between IKDC scores and hop height asymmetry was particularly strong, with values exceeding 0.80 for hops scored at the waist and shank. Good-to-moderate associations were observed for the relationship between hop height asymmetry and scores on the ACL-RSI, Pain Visual Analog Scale, and Global Knee Rating scale. The strengths of association between hop height asymmetry and asymmetry in neuromuscular performance variables were significant only for quadriceps index and asymmetry in quadriceps maximal rate of force development. The strengths of relationship in these cases were generally moderate.

When ACL-reconstructed subjects were analyzed as a subgroup, limb symmetry indices for hop height continued to demonstrate significant and strong association with IKDC scores, and also demonstrated significant and strong association with Pain ratings (Table 4.11). The relationships between hop height asymmetry and other patient-reported outcomes were lower than those observed with all subjects, and were all statistically insignificant. In addition, the associations between hop height asymmetry and asymmetry in neuromuscular performance were all statistically insignificant.

	LSI WA	**LSI SA**	**LSI FP**
ACL-RSI	0.389	0.685 †	0.676 †
IKDC	0.617 †	0.836 †	0.856 †
TSK	-0.436 *	-0.404	-0.391
Pain VAS	-0.608 †	-0.710 †	-0.727 †
Marx Activity	-0.068	0.137	0.265
Global Knee Rating	0.474 *	0.671 †	0.686 †
Quadriceps Index	0.428 *	0.434 *	0.540 †
LSI Quadriceps MRFD	0.563 †	0.343	0.424 *
LSI Hamstrings MVIC	0.204	0.325	0.302
LSI Hamstrings MRFD	0.241	0.012	-0.014

Table 4.10: Pearson's Product Moment Correlations between Limb Symmetry Indices (LSIs) of vertical hop height measured by three methods, patient-reported outcomes for *all subjects*, and limb symmetry in variables of neuromuscular performance (n = 24)

WA = Vertical hop height measured with an accelerometer at the waist

SA = Vertical hop height measured with an accelerometer at the shank

FP = Vertical hop height measured with a force platform

*P<0.05 †P<0.01

	LSI WA	LSI SA	LSI FP
ACL-RSI	0.056	0.254	0.299
IKDC	0.585 *	0.750 †	0.809 †
TSK	-0.197	-0.024	0.038
Pain VAS	-0.697 *	-0.754 †	-0.757 †
Marx Activity	0.075	0.270	0.445
Global Knee Rating	0.268	0.385	0.461
Quadriceps Index	0.180	0.272	0.452
LSI Quad MRFD	0.458	0.379	0.531
LSI Hamstrings MVIC	-0.166	-0.04	-0.115
LSI Hamstrings MRFD	0.153	0.118	-0.097

Table 4.11: Pearson's Product Moment Correlations between Limb Symmetry Indices of vertical hop height measured by three methods, limb symmetry in variables of neuromuscular performance, and patient-reported outcomes for *ACL-reconstructed subjects* (n = 12)

LSI = Limb Symmetry Index

WA = Vertical hop height measured with an accelerometer at the waist

SA = Vertical hop height measured with an accelerometer at the shank

FP = Vertical hop height measured with a force platform

*P<0.05 †P<0.01

Discussion

Reliability and validity: The primary aims of this study were to determine the reliability and concurrent validity of using accelerometers to estimate single leg vertical hop height. Coefficients for intra-rater reliability, inter-rater reliability, and concurrent validity all exceeded 0.90, which surpassed the hypothesized value of 0.80. Furthermore, reliability and validity coefficients were high for estimating hop height at the waist and shank, and were similar for healthy and ACL-reconstructed subjects. The reliability coefficients reported in this study are similar to those reported by Casartelli et al (2010), and higher than other studies, which generally varied between 0.80 and 0.90 (Picerno et al., 2011, Castagna et al., 2013, Choukou et al., 2014). Previous studies report validity coefficients ranging from 0.71 to 0.98 (Picerno et al., 2011, Choukou et al., 2014, Castagna et al., 2013, Casartelli et al., 2010). The validity coefficients reported in this study fell toward the higher end of this range, exceeded only by a value of 0.98 reported by Casartelli et al. (2010).

Reliability coefficients in this study were slightly higher and had tighter confidence intervals for inter-rater reliability than for intra-rater reliability, which is unusual. Although the reasons for this are unclear, I speculate that the primary rater, who spent a far greater volume of time scoring hop height, was more likely to deviate from the strict criteria of the scoring algorithm and rely on occasional subjective judgement during his second rating. Conversely, the secondary rater was more likely to adhere strictly to the scoring algorithm, leading to more reproducible results. This finding actually bolsters support for the utility and generalizability of the scoring algorithm used in this study rather than subjective judgement based on experience.

This is the first study to my knowledge that investigated the reliability, validity, and measurement error for using accelerometers to estimate single leg vertical hop rather than double leg vertical jump. This is noteworthy because for a single leg hop, the height is approximately half that of a double leg jump. Errors in scoring were constant across differing jump heights. Thus, errors affecting the reliability and validity are theoretically larger in a single leg hop than a double leg jump. This is also the first study to my knowledge that investigated the reliability, validity, and measurement error of using accelerometers to estimate single leg vertical hop in a cohort of injured subjects with neuromuscular impairment who may use different hopping strategies than healthy subjects (Ernst et al., 2000). Our results indicate excellent reliability and concurrent validity in ACL-reconstructed and healthy subjects. Furthermore, systematic and random measurement errors reported in our study were low. The favorable comparison of our results against those reported in the literature for healthy subjects performing double leg jumping is encouraging. This bolsters the generalizability of the technique to injured subjects, and should foster clinician confidence in using this method.

Error: The second aim of this study was to determine the error, effect size, and explore clinical implications of using accelerometers to estimate single leg vertical hop height. In support of my hypothesis, systematic and random errors were low. The systematic error reported in our study was negligible for the shank accelerometer while the waist accelerometer tended to underestimate hop height by approximately 1 cm. We chose to report error in terms of hop height rather than flight time in this study. Because the formula for estimating hop height from flight time contains the *square* of flight time, errors in estimating hop height are magnified when compared to scoring the time of flight. For example, with a typical hop height of 15 cm, 1 cm of error is the result from only 12 msec of error in flight time estimation.

Previous studies using accelerometers to estimate double leg jump height report a range of systematic error ranging. Picerno et al. (2011) reported 20 msec of systematic error with an inertial measurement unit comprised of accelerometers and gyro sensors placed at midline of the trunk at the L5 level (corresponding to about 1.75 cm of error given a 15 cm hop height). Castagna (2013), Choukou et al. (2014) and Casartelli et al. (2010) reported systematic error ranging between 3 cm and 6 cm while measuring acceleration at the pelvis with the Myotest, a commercially-available device (Myotest SA, Switzerland).

The 95% limits of agreement reported in the Bland Altman plots for shank and waist accelerometers in our study (4.64 cm and 5.70 cm, respectively) compare favorably against those reported in previous research. Picerno et al. (2011) reported

95% limits of agreement of 10.9 cm with custom sensors. In three studies using the Myotest, the 95% limits of agreement for countermovement jumps ranged between 6 cm and 28 cm (Choukou et al., 2014, Casartelli et al., 2010, Castagna et al., 2013). The measurement errors reported in this study (1.03 cm and 0.84 cm for the waist and shank accelerometers, respectively) were also lower than previously reported by Casartelli et al. (2010) for random error.

The low values for systematic error, random error, and 95% limits of agreement obtained in our study may be explained by the scoring algorithm used in this study versus the Myotest. Our scoring algorithm identified take-off as the deflection points after the positive acceleration and landing as the deflection point that preceded a significant (10 m/s^2) acceleration at landing. In contrast, the Myotest first integrated the acceleration signal, and then identified take-off as the peak positive velocity and landing as the peak negative velocity. This method likely underestimates the time at take-off and overestimates the time at landing, thus lengthening flight time at both ends of the hop (Casartelli et al., 2010).

In contrast to my hypothesis, error was not similar at both anatomic locations. Rather, the waist systematically underestimated hop height by about 1 cm. This could be related to the natural filtering effect of measuring acceleration at the more proximal location of the waist, where the exact timing of events happening at the ground may be more difficult to detect. Anecdotally, the acceleration signal at the shank featured more distinct deflections than that at the waist. Measuring closer to where the event is taking place (the ground) removes the natural filtering effect of the lower extremity.

The values provided in our study for error range and minimal detectable change are designed to provide the end-user with a guide to understand the meaning and practical relevance of differences in hop height measured with this technique. Error range provides a 95% confidence interval for a single measurement (the mean of 5 trials in this study). The error ranges in this study for waist and shank accelerometers were 2.01 cm and 1.64 cm, respectively. Thus, for a hop height of 15 cm measured via a waist accelerometer, the average of all possible measurements, which would be expected to be close to the true measurement, would be within the range 15 ± 2.01, or 12.99 cm to 17.01 cm. In contrast, minimal detectable change describes the ability of an instrument to detect meaningful differences between two measurements (Weir, 2005). This is more useful in evaluating for between-limb asymmetry or within-limb changes before and after intervention. The minimal detectable changes reported in this study for waist and shank accelerometers were 2.85 and 2.32, respectively. Thus, at the waist or the shank, a difference between two measurements less than 2 cm should be interpreted as not meaningful, while a difference above 3 cm should be viewed as meaningful, regardless of hop height.

Caution should be taken not to generalize values for Measurement Error, error range, and minimal detectable change reported in this study to other accelerometer-based measurement systems for hop height. The values for these terms in this study are specific to our scoring algorithm and our approach with data management. Commercially-available, accelerometer-based systems for scoring jump height use different scoring algorithms and likely have different error characteristics.

The measurement range in this study was wide, with vertical hops ranging from 3 cm to 26 cm as judged by the force platform (criterion standard). Error remained relatively constant across the range of hop height rather than varying significantly by the height of the hop. This wide range of height with fairly constant error improves the generalizability of the scoring method across a range of hop height. Thus, the utility of accelerometers for estimating hop height may be hampered less by the variability of the error, and more by the coefficient of variation at low hop heights. Given a typical hop height of 15 cm, a difference of 3 cm represents a 20% deficit, and has corresponding coefficients of variation of 6.9% at the waist and 5.6% at the shank. This 20% deficit may be difficult to detect with visual observation alone. But, given a hop height of 6 cm, a 3 cm difference represents a 50% deficit, and has corresponding coefficients of variation of 17% at the waist and 14% at the shank. A 50% difference would likely be quite obvious with visual observation alone. Given these measurement characteristics, it seems prudent that a hop height of 10 cm (corresponding to a 30% difference between limbs and coefficients of variation less than 10%) is required for accelerometers to exceed visual observation alone in detecting asymmetry between limbs.

Differences between limbs and groups: The effect sizes observed for between-limb differences in vertical hop height from all three methods were large and exceeded those for peak quadriceps strength and rate of force development. Similarly, the limb symmetry indices observed from hopping were lower than those from neuromuscular testing. These findings are consistent with reports from previous studies that suggest that single leg vertical hopping is among the most sensitive tests to identify asymmetry between limbs in patients after ACL-reconstruction (Thomee et al., 2012, Gustavsson et al., 2006). We report LSIs in this study ranging between 67% and 74%, depending on the method of scoring. This asymmetry is similar to previous studies for vertical hopping that report LSIs ranging between 74% and 76% at a similar time point after ACL-reconstruction (de Fontenay et al., 2014, Gustavsson et al., 2006, Petschnig et al., 1998). Furthermore, these values were lower (i.e., greater asymmetry) than LSIs reported for horizontal hopping (Gustavsson et al., 2006, Thomee et al., 2012, Petschnig et al., 1998).

Our results for estimating hop height using portable accelerometers are promising, especially when compared to previous studies. Accelerometers worn at the shank and waist displayed similar limb symmetry indices. However, the shank location outperformed the waist location in several ways. Although the point estimates for concurrent validity were similar at both locations, the shank location demonstrated tighter confidence intervals than the waist location. Second, the shank location demonstrated virtually no systematic error when compared to the criterion standard, while the waist under predicted hop height by about 1 cm. The shank location also demonstrated less random error than at the waist, leading to lower values for error range, minimal detectable change, and coefficients of variation. Third, the effect size for between-limb differences in the ACL-reconstructed cohort was larger and more similar to the force platform at the shank location than at the waist location. Both raters also anecdotally reported greater ease in estimating take-off and landing from the shank location, where deflections were obvious, even from hop heights at the lower end of the range. Thus, for greatest measurement accuracy and precision, we recommend the accelerometer be placed at the shank rather than the waist when possible.

Relationships between hop height and neuromuscular performance: The third aim of this study of this study was to define the relationships between hop heights, variables of thigh muscle performance, and patient-reported outcome. The results indicate that hop height scored by all methods were most strongly associated with quadriceps MVIC and maximal rate of force development. These relationships exceeded hypothesized correlation coefficients of 0.50. Strengths of association were moderate with all subjects analyzed together and good-to-excellent with ACL-reconstructed subjects alone (Portney, 1993). The associations between vertical hop height and peak quadriceps MVIC in ACL-reconstructed subjects exceeded 0.80 for all methods of scoring. These values exceed those reported previously in the literature, where associations between peak strength and hopping have been reported at 0.51 for vertical hopping (Petschnig et al., 1998) and 0.41 to 0.62 for the hop for distance (Wilk et al., 1994). The strong relationship between quadriceps strength and hop height reported in our study also lends support to previous research that reported symmetry in quadriceps strength was a significant predictor in single leg hop performance (Schmitt et al., 2012).

Relationships between hop height symmetry and patient-reported outcomes (PROs) revealed moderate to strong association with IKDC and Pain scores with all subjects analyzed together and strong association with ACL-reconstructed subjects analyzed as a subgroup. These relationships far exceeded the hypothesized correlation coefficient of 0.40. Our results also exceeded associations between hop tests and patient-reported outcome described in previous research in subjects at a similar time point after ACL-reconstruction. Previous studies reported correlation coefficients that ranged between 0.40 and 0.50 between hop testing and several patient-reported outcomes, including the IKDC (Ageberg et al., 2008, Wilk et al., 1994, Ra et al., 2014). One factor that may account for the differences between our correlation coefficients and previously reported values is which variables were actually compared. Two of these previous studies compared a patient-reported outcome score to actual hop performance. In contrast, we compared PRO scores to limb symmetry indices for hop performance rather than raw score. But, Ra et al (2014) did compare IKDC scores to LSI values for hopping for distance, and still reported correlation coefficient of less than 0.50. In this case, the difference may be related to the hop testing performed (i.e., vertical hop vs. hop for distance). Relationships between hop height symmetry and PROs also exceeded the relationships between hop height symmetry and between-limb symmetry in variables of neuromuscular performance. This finding affirms the utility of self-perceived function and pain in influencing maximal physical performance.

Scoring hop height from flight time has inherent limitations regardless of measurement technique. The flight time technique assumes that the limb is in the same position at take-off and landing (Kibele, 1998). Thus, it assumes that the maximal height of the hop occurs at exactly 50% of the flight time, which is simplistic. Despite these assumptions, scoring hop height from flight time is a commonly-accepted technique used by force platforms, contact mats, and photoelectric cells. The accuracy of this technique exceeds that from the jump and reach method (Leard et al., 2007). Using flight time from force platform data is commonly accepted as a gold standard (Leard et al., 2007, Castagna et al., 2013, Palma, 2008). The magnitude of error associated with the flight time method of calculating hop height (several millimeters) is small when compared to the magnitude of between-limb differences deemed to be clinically meaningful (several centimeters).

The accelerometer sensors mounted at the waist and shank were secured firmly with elastic bands and molded brackets designed specifically to contain the accelerometers. Despite these efforts, it is possible that the accelerometers were subjected to soft tissue movement artefact which may have led to error in measurement. If movement error did occur due to soft tissue artefact, this error may have been retained due to the cut-off frequency of 50 Hz for low-pass filtering. However, the 50 Hz cut-off frequency was necessary in order to preserve signal present during landing, especially at the shank location.

This study is part of a line of research that investigates novel uses of wireless, wearable accelerometer sensors to characterize patient outcome in an expedient fashion. This line of research leverages recent improvements in the technology of these accelerometers to permit unfiltered acceleration data over a full physiologic range to be collected at a high frequency. These data can either be stored on a memory card within the sensor or streamed real-time to a computer or portable electronic device. Ideally, accelerometers will be linked to applications on portable electronic devices. This is significant because it would permit real-time processing and instantaneous feedback to the patient/client and clinician. Because accelerometer sensors are already being used to objectively measure physical activity and estimate sleep, these devices have potential to provide a diverse set of data capable of characterizing patient outcome in new and exciting ways.

Conclusion

This study investigated the ability of wearable accelerometer sensors to estimate single leg vertical hop height, a common performance-based measure that has high functional relevance to athletic participation. The results of this study affirm excellent reliability, concurrent validity, and low error of this method in healthy subjects and a cohort of subjects after ACL-reconstruction. This is particularly true when acceleration is measured at the shank location. Estimates for error range and minimal detectable change should guide practical application of this technique in clinical practice. This analysis suggests that accelerometers may be used as an inexpensive and expedient method of quantifying hop height.

CHAPTER 5

USING PORTABLE ACCELEROMETERS TO EVALUATE LOWER EXTREMITY MECHANICS AFTER ACL RECONSTRUCTION

Introduction

Anterior cruciate ligament (ACL) injury is common, especially among young, active people. The yearly incidence of ACL injuries has been estimated as high as 36.9 per 10,000 people (Gianotti et al., 2009), and up to 300,000 ACL reconstruction surgeries are performed annually in the United States (Cohen and Sekiya, 2007). Despite surgery and subsequent rehabilitation, overall outcome is not ideal. Individuals with ACL-reconstructed knees show persistent abnormalities in lower extremity biomechanics, are at increased risk for subsequent ACL injuries, achieve low rates of return to pre-injury levels of activity, and have a marked increase in the incidence of post-traumatic osteoarthritis (OA). In order to track and improve patient outcomes, patient-reported outcomes or performance-based measures are typically used. These provide important information about perceived function and specific task performance, respectively. But, these types of outcomes are incapable of providing important biomechanical assessment.

Impairments after ACL reconstruction are common and may persist after rehabilitation is complete and patients have returned to full activity, including participation in high-level sports (Hart et al., 2010a, Roewer et al., 2011). These impairments include quadriceps dysfunction and altered neuromuscular control (Williams et al., 2005, Urbach and Awiszus, 2002, de Jong et al., 2007, Krishnan and Williams, 2011). Mechanical adaptations during gait and other functional activity are also common, and appear most apparent in the sagittal plane, where a “stiffening” strategy of the knee is frequently observed. Hallmarks of altered sagittal plane mechanics include reductions in peak internal knee extension moment, range of sagittal plane knee excursion, and peak knee flexion angle during the loading response phase of gait (Berchuck et al., 1990, Noyes et al., 1992, Wexler et al., 1998, DeVita et al., 1998, Webster et al., 2005, Hurd and Snyder-Mackler, 2007, Hall et al., 2012). This stiffening pattern also results in correspondingly high values in vertical ground reaction force while landing from a jumping activity (Hewett et al., 2005, Paterno et al., 2010). Lewek et al. (2002) report a strong association between altered gait mechanics and quadriceps weakness. More recent evidence shows that mechanical alterations persist well beyond discharge from post-surgical ACL rehabilitation even if quadriceps strength averages greater than 90% of the opposite side (Roewer et al., 2011). Altered gait patterns have also been observed in people who are able to successfully pass a strict battery of performance tests designed to assess readiness for return to full sports participation. However, patient and performance-based return to sports criteria alone are insufficient to identify clinically-meaningful side-to-side differences in sagittal plane mechanics (Di Stasi et al., 2013).

Using motion capture, altered gait mechanics can be observed in level walking where the demand on the lower extremity is low. In activities with higher demand (e.g., running, jumping, stair ascent and descent), mechanical alterations are greater (Rudolph et al., 2001, Kuenze et al., 2013, Thambyah et al., 2004, Hooper et al., 2002, Gao et al., 2012, Hall et al., 2012, Ernst et al., 2000). Stepping down during ongoing gait is a common functional task in daily living. For example, people perform this when stepping off a curb to cross a street. This routine task places significant demand on the thigh muscles, which control the descent of the body’s mass. This high demand and functional relevance has led researchers to use this task as a method of assessing differences in high and low functioning ACL-deficient and control (Houck and Yack, 2003).

The altered loading pattern caused by abnormal biomechanics is one factor believed to play a role in the high rates of post-traumatic osteoarthritis (PTOA) and subsequent knee ligament injury seen in individuals after ACL reconstruction and arthroscopic partial meniscectomy (Liikavainio et al., 2007, Sturnieks et al., 2008a). The prevalence of PTOA after ACL reconstruction has been reported at 48% six years after surgery and 71% 10-15 years after surgery (Oiestad et al., 2010, Holm et al., 2010, Keays et al., 2010). The prevalence of subsequent ACL injury after ACL reconstruction has been reported at up to 26% at 10 years after surgery (Pinczewski et al., 2007). Despite the high prevalence of these sequelae and the contributing role believed to be played by abnormal sagittal plane mechanics, it remains difficult for clinicians to appreciate subtle biomechanical changes during clinical examination without expensive motion capture and analysis systems. These motion capture systems are rare in clinical settings. Moreover, the time required to collect and analyze biomechanical data with motion capture based biomechanical studies is prohibitive in most clinical enterprises as well as in large multi-center research studies. A low cost, efficient approach to collecting meaningful biomechanical data in a clinically-feasible manner would be a significant development with the potential to advance clinical practice and multi-center research in meaningful ways.

Accelerometers mounted at the pelvis have been used to reliably record gait events. Strong associations exist between vertical acceleration measured at the pelvis and vertical ground reaction force (Rowlands and Stiles, 2012). Thus, accelerometers may offer a practical alternative for instrumented assessment of gait mechanics that would make screening for altered kinematics and kinetics possible without access to a gait analysis laboratory.

The purpose of this investigation was to take the first step in the process of developing a low cost, clinically feasible, portable collection system able to detect functional movement symmetry. The specific purpose was to identify the ability of wearable accelerometer sensors to detect movement asymmetry characterized by underlying alterations in sagittal plane knee mechanics. I hypothesized that wearable accelerometers would provide a clinically-feasible method for identifying asymmetry associated with abnormal gait biomechanics in subjects who have undergone ACL reconstruction.

Methods

Subjects: Thirty subjects participated in this study approved by University of Iowa's Institutional Review Board. Fifteen subjects (8 male, 7 female) had previously undergone isolated anterior cruciate ligament (ACL) reconstruction 5 to 19 months earlier. Fifteen subjects without a history of knee injury who were age and gender-matched with the ACL reconstructed subjects served as controls. A control group was enrolled because bilateral adaptations may occur in ACL-reconstructed subjects. The time frame of 5-19 months was selected because formal rehabilitation typically ceases near the lower limit of this time spectrum, and this time point is frequently cited for return to full activity (Kvist, 2004). But, asymmetry in lower extremity muscle strength, mechanics, and functional performance are still common during this time frame (Thomee et al., 2012, Ernst et al., 2000). This asymmetry improves in many subjects between six and 24 months postoperatively (Roewer et al., 2011). Male and non-pregnant female subjects were eligible for inclusion if they were between the ages of 18 and 40 years of age with a BMI no greater than 35 kg/m^2. The 18 year old boundary increased the likelihood of skeletal maturity, and the 40 year old boundary increased the homogeneity in baseline physical characteristics of the subject pool. ACL-reconstructed subjects were excluded if they had multiligamentous injury, concomitant meniscal repair, demonstrated a grossly symmetrical gait (e.g., no noticeable limp), or were unable to flex their knee past 120 degrees.

Movement Biomechanics: Movement biomechanics were analyzed during over ground walking and stepping down during ongoing gait. These two functional tasks represent a typical spectrum of difficulty during activities of daily living. Kinematic data were collected at a sampling frequency of 60 Hz using the Optotrak Motion Analysis System (Model 3020, Northern Digital, Inc., Waterloo, Ontario). This system was calibrated according to manufacturer's guidelines prior to each session of data collection. Ground reaction force data was collected using two force platforms with a sampling frequency of 360 Hz (Model 9865B, Kistler Instrument Corp., Winterthur, Switzerland). A system of 21 infrared light emitting diodes (IREDs) served as tracking markers and were affixed to the pelvis, thighs, shanks, and feet and secured with double-backed adhesive tape and cover rolls (Fabrifoam, Exton, PA). Investigators digitized the following anatomic landmarks bilaterally using a probe with 6 IREDs and Visual 3-D software (C-Motion, Germantown, MD): Anterior and posterior superior iliac spines, medial and lateral femoral epicondyles, medial and lateral malleoli, proximal calcaneus, distal calcaneus, fifth metatarsal head, and distal second toe. Subject-specific models were created for each subject using the above marker system. This approach is similar to previous work (Houck and Yack, 2003).

For level walking, subjects walked along a 10 m walkway that included two force platforms flush with the floor. For stepping down, subjects walked along a 4.6 m platform raised 20 cm from ground level and stepped down onto a force platform. A 20 cm wooden box was bolted to the first force platform and served as an extension of the 20-cm raised platform (Figure 5.1). Thus, ground reaction force data was collected for the trailing and leading limbs in this task. The distance from the edge of the box on first force platform and the center of the second force platform was 41 cm. Cadence was standardized with a metronome set to 113 steps/min for level walking and stepping down in order to achieve consistency between trials and tasks and reduce variability in ground reaction force associated with different cadences (Rowlands et al., 2013). This cadence is typical during community ambulation by adults (Finley and Cody, 1970). To reduce variability and enhance the ability to detect differences in knee extension moments, subjects were required to use a heel-strike pattern during all trials of both walking tasks (Houck and Yack, 2003). For both tasks, only trials in which the entire foot made contact within the force platforms were accepted. Subjects performed five valid trials of each task bilaterally after adequate familiarization, with the less-difficult task of level walking performed first.

Visual 3-D software was used to process kinematic and kinetic data obtained during walking and step down trials. Marker trajectory data was interpolated with a maximum gap of 10 frames, and then low-pass filtered at 6 Hz with a fourth order Butterworth filter. The frequency cut-off of 6 Hz is consistent with previous work for level walking and stepping down (Hurd and Snyder-Mackler, 2007, Roewer et al., 2011, Houck and Yack, 2003, Hartigan et al., 2009). In addition, Houck and Yack (2003) previously used this cutoff frequency for the same task performed in this study with the same data collection system. Ground reaction force data was low-pass filtered at 50 Hz with a fourth order, zero–lag Butterworth filter.

Accelerometry: Subjects wore two identical Shimmer 3 accelerometers with a collection frequency of 256 Hz (± 16 g, Shimmer Sensing, Dublin, Ireland). For walking trials, an accelerometer was affixed to the waist at the mid-axillary line bilaterally. Acceleration

Figure 5.1: Laboratory set up for a walk and step down trial

during level walking was determined from the accelerometer that is ipsilateral to the limb stepping on the force platform. For step down trials, subjects wore both accelerometers on the same side as the leading limb. One accelerometer was affixed to the waist at the mid-axillary line just inferior to the iliac crest, and the second accelerometer was secured to the proximal lateral shank just distal to the fibular neck. Recording acceleration proximal to the knee was designed to permit insight into the “stiffened” knee pattern that is emblematic of altered mechanics after ACL reconstruction. Conversely, recording acceleration at the shank was designed to measure the initial impact sustained from the leading limb and to permit greater insight into eccentric control by the trailing limb. Firm-fitting elastic straps secured the accelerometers around the pelvis and shank with the accelerometers aligned as close as possible to the global coordinate system axes while the subjects were standing. This method has previously produced adequate belt tension to reduce noise from poor fixation (Mizrahi et al., 2000). This fixation approach did not interfere with motion capture marker placement, is feasible in a clinical setting, and has been used in other biomechanical research (Rowlands and Stiles, 2012, Rowlands et al., 2013).

Analysis of the frequency spectrum of stepping down was performed via Fast Fourier Transformation. Based on this analysis, accelerometer data was filtered at 50 Hz using a fourth order Butterworth low-pass filter. This cutoff frequency was necessary to preserve the acceleration signal from impact, which was routinely detected at over 40 Hz. This approach reduced the potential for contamination from high-frequency noise (Zijlstra and Hof, 2003, Fong and Chan, 2010). The vertical component of acceleration at the pelvis and shank was the largest component among the three orthogonal planes and was of primary interest in this study. Under ideal settings, the accelerometer would be positioned in such a way as to measure vertical acceleration in one channel. Despite efforts to standardize placement, variations in body shape and limb segment rotation may have resulted in the accelerometer being aligned in a plane that makes vertical acceleration impossible to measure from only one channel. Because of this, the "resultant" acceleration vector was calculated from all three channels using the Euclidean Norm method ($Accel_{Res} = \sqrt{X^2 + Y^2 + Z^2}$). Peak acceleration was recorded from the vertical and resultant axes of acceleration measured at the shank and waist (Figures 5.2 and 5.3, respectively).

a)

b)

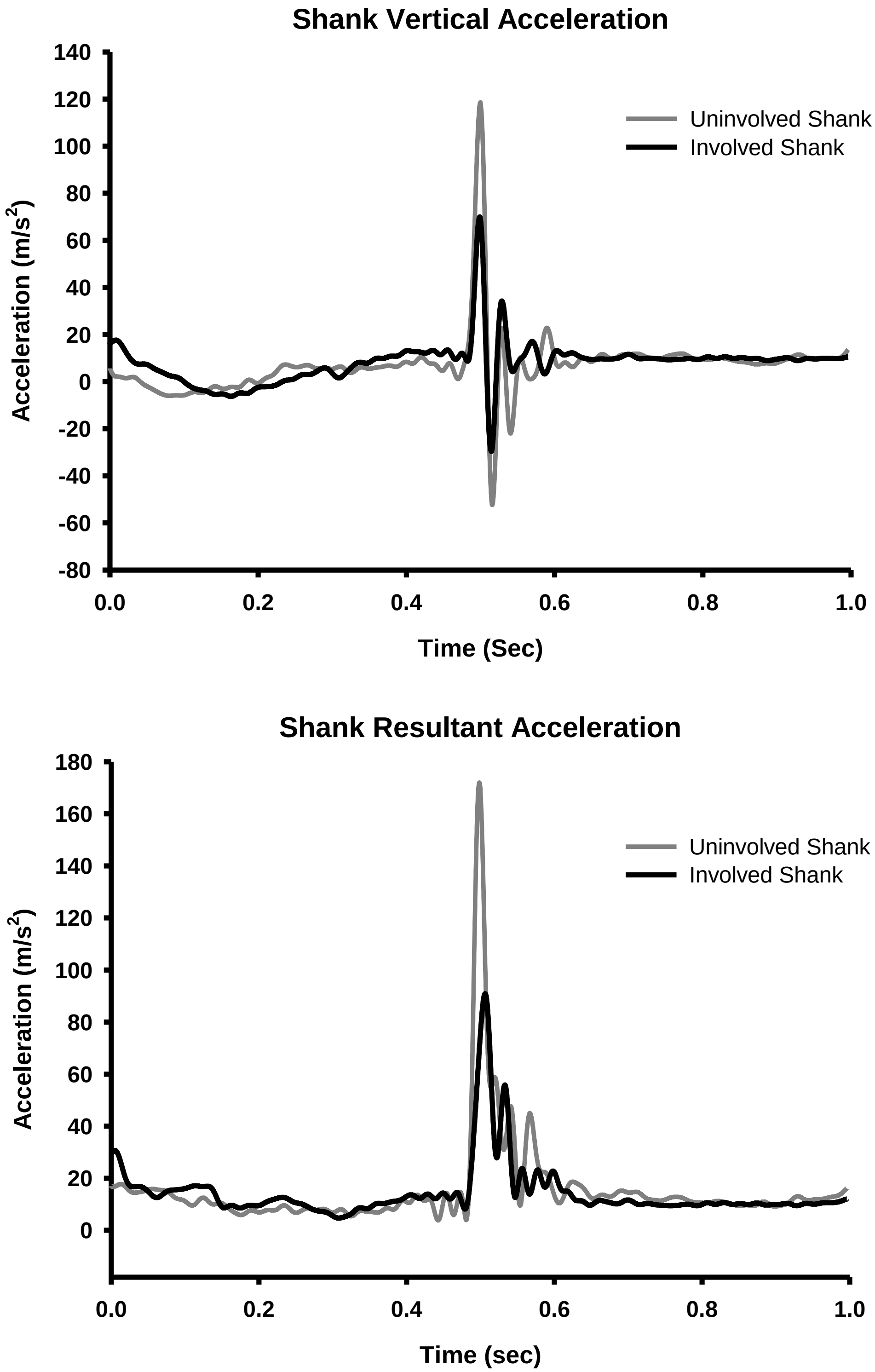


Figure 5.2: Sample acceleration data measured at the shank while stepping down in a subject with asymmetry

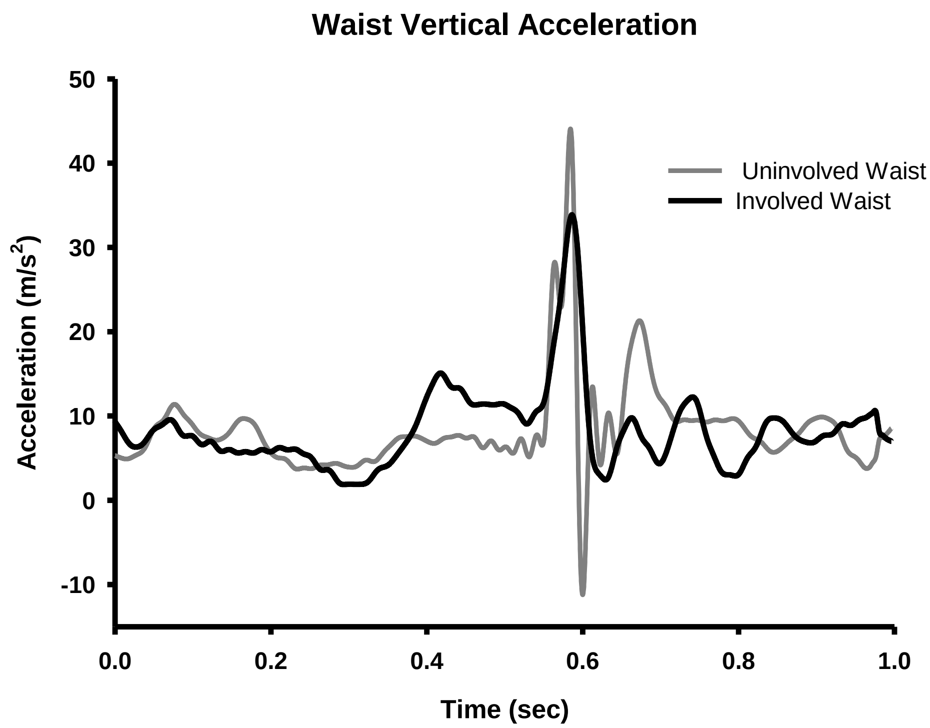


Figure 5.3: Sample acceleration data measured at the waist while stepping down in a subject with asymmetry

<u>*Neuromuscular testing*</u>: For neuromuscular strength testing, subjects sat on a HUMAC NORM Testing and Rehabilitation System (CSMI, Stoughton, MA) with the hips and knees flexed to 85 degrees and 60 degrees, respectively. A seat belt, chest straps, and thigh strap secured subjects to the chair. This approach optimizes the length-tension relationship of the quadriceps and permits insight into side-to-side ratios of both quadriceps and hamstring muscle groups (Krishnan and Williams, 2014). Force data was collected with a custom apparatus consisting of a modified shin guard and load cell affixed to the rigid arm of the HUMAC NORM. The shin guard was attached to the shank with a Velcro strap approximately 5 cm proximal to the medial malleolus. The load cell (model LPU-500, Transducer Techniques, Temecula, CA) was placed in series between the shin guard and rigid arm. The same testing configuration was used bilaterally.

Subjects performed a minimum of 3 maximal voluntary isometric contractions (MVICs) of 5 seconds duration with two minutes of rest in between trials for knee extension and knee flexion. Familiarization trials at 50%, 75%, and 90% of perceived effort, loud verbal encouragement, and real-time feedback of force development were provided to encourage maximal effort. Subjects were required to perform a minimum of two trials within ± 5% of each other to ensure reliability of the data. In addition to MVIC contractions, five trials of rapid knee extension and knee flexion were collected with the instructions to kick out or pull back *as fast* and as hard as possible. Data from these trials was collected to permit secondary analysis of how performance in single leg hopping relates not only to quadriceps and hamstrings strength, but also the maximal rate of force development. Data was sampled at 1000 Hz, collected with an AD Instruments PowerLab 16/30, and processed via LabChart 8 software (AD Instruments, Bella Vista, Australia).

Patient-reported outcomes: Patient-reported measures assessed pain and self-perceived function. A visual analog scale assessed pain on a 100 mm scale during the walk and step-down task. The International Knee Documentation Committee (IKDC) subjective knee evaluation form assessed knee symptoms and self-perceived function during sport and strenuous activity. The IKDC subjective form is highly reliable and shows high construct validity in subjects with ACL reconstruction (Irrgang et al., 2001, van Meer et al., 2013). The Global Rating of Knee Function (also known as the SANE rating) provided a single numerical score between 0 (complete disability) and 100 (no disability) in response to the question "On a scale of 0 to 100, how would you rate the function of your knee (with 100 being complete function)?"(Williams et al., 2000). The Tampa Scale of Kinesiophobia-11 (Woby et al., 2005) measured fear of activity and re-injury, which has been shown to be a significant obstacle to full return to pre-injury activity level (Chmielewski et al., 2008, Flanigan et al., 2013). Similarly, the ACL Return to Sport after Injury (ACL-RSI) measured psychosocial response to returning to sport after ACL reconstruction. This has shown good reliability and ability to help predict return to sport after ACL reconstruction (Webster et al., 2008, Langford et al., 2009, Muller et al., 2014). The Marx activity scale measured participation in athletic activities and has shown high reliability (Marx et al., 2001). These measures of self-reported pain, perceived function, and psychosocial response to activity allow for a comprehensive characterization of the subject pool and permit secondary analysis into the relationship between perceived function, fear avoidance, and hop performance measured in this study.

Data Analysis

Visual 3-D software was used to calculate kinematic and kinetic data. Joint moments were normalized to body mass (kg) and expressed as internal joint moments. Ground reaction force was normalized to body weight (N). For acceleration, kinetic, kinematic, and neuromuscular variables of interest (Table 5.1), mean values for each subject were calculated based on five trials of level walking and stepping down.

Group baseline characteristics and patient-reported outcomes: Descriptive statistics were calculated on continuous variables and the results of dichotomous outcomes summarized. Independent t-tests with the level of significance set to 0.05 were used to evaluate for differences in characteristics between ACL-reconstructed and control subjects.

Differences between limbs and groups: One-way Analyses of Variance (ANOVA) with four levels for Limb were performed to assess for differences between limbs in the ACL-reconstructed and Control groups for acceleration, kinematic, kinetic, and neuromuscular

Parameter	Primary variable	Secondary variable
Pelvic acceleration	Peak acceleration during loading response (2nd peak)	Peak acceleration at heel strike (1st peak)
Shank acceleration	Peak acceleration	--
Ground Reaction Force	Peak vertical ground reaction force at Heel Strike	Peak vertical ground reaction force during Loading Response
Ground Reaction forced timing	Timing of peak ground reaction force (% Stance)	--
Knee kinetics	Peak internal knee extension moment for leading leg	Peak internal knee extension moment for trailing leg
Knee kinematics	Sagittal plane knee excursion during loading response	Peak knee flexion angle during loading response

Table 5.1: Variables of interest for data analysis

variables. Within each group, differences between limbs for acceleration, kinematic, kinetic, and neuromuscular parameters of interest were evaluated with Paired t-tests with the level of significance set to 0.05. Between-limb effect sizes for the ACL-reconstructed group were computed via Cohen's *d* for significant variables of acceleration, kinematic and kinetic parameters, and neuromuscular function. When interpreting Cohen's *d*, a value of 0.2 was considered a small effect size, 0.5 considered a medium effect size, and 0.8 considered a large effect size (Cohen, 1992). Limb symmetry indices (LSIs) were calculated for the ACL-reconstructed group by dividing the result from the involved leg by the result from the uninvolved leg. For control subjects, LSIs were calculated by dividing the result for the nondominant limb by the result for the dominant limb. Differences in limb symmetry between groups were assessed with independent t-tests with the level of significance set to 0.05.

Relationships between acceleration and other variables of variables of interest: Pearson's Product Moment Correlations determined the relationships between several sets of variables. First, Pearson's Product Moment Correlations were computed to identify associations between peak accelerations at the waist and shank and kinematic, kinetic, and neuromuscular parameters of interest. Next, associations between parameters of vertical ground reaction force and kinematic, kinetic, and neuromuscular parameters of interest were identified. Finally, Pearson's Product Moment Correlations were determined for associations between LSIs in peak acceleration, patient-reported outcomes, and LSIs for parameters of kinetics, kinematics, and neuromuscular performance. When interpreting Pearson's Product Moment Correlations, values between 0.25 to 0.50 were considered fair, values between 0.50 and 0.75 were considered moderate-to-good, and values over 0.75 were considered good-to-excellent (Portney, 1993).

Linear Regression: In order to explore the multivariate nature of the relationship between peak acceleration and underlying mechanics, eight iterations of linear regression were performed. Four dependent variables (peak vertical and resultant acceleration at the waist and shank) were evaluated against two sets of predictor variables. Each set of predictor variables contained two variables from a similar domain identified from simple correlations as possessing moderate to good association with peak accelerations. These two set of predictor variables were 1) Leading and Trailing Limb Peak Knee Extension Moments and 2) Peak Vertical Ground Reaction Force at Heel Strike and the Timing of Peak Vertical Ground Reaction Force. SPSS version 23 software was used to perform all statistical analysis (SPSS Inc, Chicago, IL).

Results

Group baseline characteristics and patient-reported outcomes: ACL-reconstructed subjects and control subjects were similar in age, height, and self-reported activity level (Table 5.2). ACL-reconstructed subjects demonstrated significantly higher body mass and BMI when compared to control subjects. ACL-reconstructed subjects scored significantly lower than control subjects on self-reported knee function, readiness to return to sport, pain, and global knee rating.

	Control	**ACL-Reconstructed**
Age (years)	26.60 ± 5.42	26.73 ± 6.30
Sex	8 M, 7 F	8 M, 7 F
Time since surgery (months)	n/a	9.80 ± 3.4 (range 5-19)
ACL graft type		10 STG, 5 PBTB
Mass (kg)	66.61 ± 12.28	80.55 ± 12.22 *
Height (m)	1.69 ± 0.07	1.74 ± 0.08
BMI (kg/m^2)	23.19 ± 3.15	26.41 ± 3.37 *
ACL-RSI (%)	96.11 ± 5.76	56.22 ± 20.86 *
IKDC (%)	98.77 ± 1.72	80.23 ± 11.54 *
Tampa Scale for Kinesiophobia-11	16.53 ± 4.16	19.20 ± 4.80
Pain Visual Analog Scale (mm)	0	4.33 ± 7.36 *
Marx Activity	10.80 ± 4.07	11.47 ± 3.46
Global Knee Rating	99.10 ± 2.58	86.00 ± 8.70 *

Table 5.2: Subject characteristics and self-reported outcomes

STG = Semitendinosus-Gracilis ACL graft

PBTB = Patellar bone-tendon-bone ACL graft

*P < 0.05 from Control group

Differences between limbs and groups

Level walking: For level walking, there were no differences between limbs in either group for peak acceleration at heel strike or loading response, peak knee extension moments, or peak knee flexion angle. Both groups demonstrated between-limb differences in peak vertical ground reaction force (P = 0.018 and P = 0.013 for ACL-reconstructed and Control groups, respectively). The ACL-reconstructed group stepped harder onto the uninvolved limb, and the Control group stepped harder onto the dominant limb. For both groups, the difference between limbs was 4% of body weight, corresponding to about 3% of the mean. The ACL-reconstructed group also demonstrated a between-limb difference in sagittal plane knee excursion while walking, with the involved limb undergoing approximately four degrees less knee flexion, corresponding to about 25% of mean values (P = 0.004).

Walking and Stepping down: For walking and stepping down, significant differences were observed between limbs for waist vertical and resultant acceleration in the ACL-reconstructed group (P = 0.002 and P = 0.005, respectively; Figure 5.4). Higher accelerations were detected when stepping onto the uninvolved limb. These differences were apparent with peak acceleration during loading response (the second peak), but not with the first acceleration peak occurring at heel strike (P = 0.716 and P = 0.804, for vertical and resultant accelerations, respectively). Differences between-limbs were not significant for shank vertical or resultant acceleration while stepping down (P = 0.069 and P = 0.12, respectively), likely due to higher between-subject variability observed with acceleration at the uninvolved shank versus the waist.

Significant differences between limbs while stepping down were detected for many kinematic and kinetic parameters of interest (Table 5.3). Peak knee extension

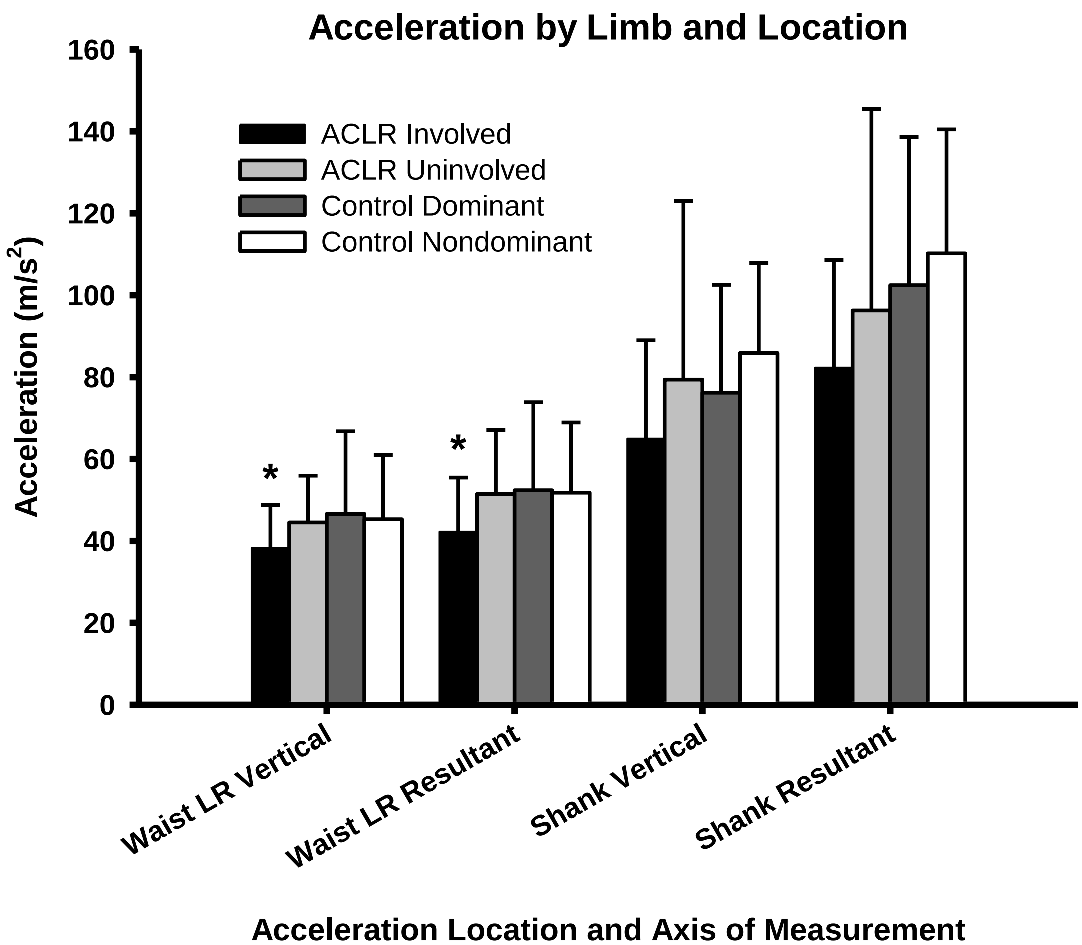


Figure 5.4: Differences between limbs for peak acceleration measured at the waist and shank

*P < 0.05 for ACL-reconstructed group when compared to Control group

LR = Loading Response

moments for the leading limb were lower for the ACL-reconstructed limb versus the uninvolved limb (P = 0.007) and both Control subjects' limbs (P = 0.08 and P = 0.015 for the nondominant and dominant limbs, respectively). Similarly, trailing limb peak knee extension moments were significantly lower in the ACL-reconstructed limb versus the uninvolved limb ($P < 0.001$) and both Control subjects limbs ($P = 0.002$ and $P = 0.011$ for the nondominant and dominant limbs, respectively). Although no differences were

	ACLR Involved	**ACLR Uninvolved**	**Control Nondominant**	**Control Dominant**
Lead Peak KEM (Nm/kg)	1.24 ± 0.38 *†	1.77 ± 0.68	1.85 ± 0.57	1.80 ± 0.73
Trail Peak KEM (Nm/kg)	1.36 ± 0.36 *	1.98 ± 0.37 †	1.87 ± 0.51	1.78 ± 0.46
Lead Peak KFA (degrees)	25.8 ± 10.7	30.1 ± 11.0	30.9 ± 5.7	28.0 ± 12.5
Lead Sagittal Knee Excursion (degrees)	14.8 ± 3.2 *	22.4 ± 6.6	18.6 ± 7.1	17.9 ± 11.1
Peak Heel Strike Vertical GRF (% body weight)	169 ± 30% *	206 ± 36%	190 ± 36%	190 ± 37%
Peak Loading Response Vertical GRF (% body weight)	150 ± 22% *†	165 ± 26%	166 ± 21%	175 ± 25%
Overall Peak Vertical GRF (% Body weight)	175 ± 25% *	209 ± 34%	196 ± 32%	193 ± 29%
Peak Vertical GRF Timing (% Stance)	10.4 ± 5.3%	8.4 ± 4.1%	11.2 ± 5.3%	11.4 ± 5.1%
Lead Stance Time (sec)	0.67 ± 0.07 †	0.68 ± 0.07 †	0.62 ± 0.07	0.62 ± 0.04

Table 5.3: Differences between limbs and groups for kinematic and kinetic parameters of interest

Values are mean ± standard deviation

KEM = Knee Extension Moment   KFA = Knee Flexion Angle

GRF = Ground Reaction Force

*$P < 0.05$ for ACLR subjects' Involved limb when compared to the Uninvolved limb

†$P < 0.05$ for ACLR subjects' Involved limb when compared to Control subjects' limbs

detected in peak knee flexion angle during loading response, the ACL-reconstructed knee demonstrated less sagittal plane excursion during loading response than the uninvolved knee ($P < 0.001$). The Control group did not demonstrate a significant between-limb difference for any kinematic or kinetic parameter of interest.

Significant differences between limbs while stepping down were also apparent for several parameters of interest for vertical ground reaction force (GRF; Table 5.3). ACL-reconstructed subjects stepped onto their uninvolved limb with significantly greater force than their reconstructed limb. These differences were apparent for the overall peak vertical GRF ($P < 0.001$) as well as for GRF corresponding to peak heel strike ($P < 0.001$) and peak loading response ($P = 0.03$). No differences were detected between limbs for the timing of peak vertical GRF or for overall stance time. The Control group did not demonstrate a significant between-limb difference for any parameter of vertical ground reaction force.

Evaluation for between-limb differences in quadriceps and hamstrings function in the ACL-reconstructed group revealed significantly weaker quadriceps MVIC and slower MRFD in the involved limb compared to the uninvolved limb ($P = 0.006$ and $P = 0.004$, respectively; Table 5.4). The involved limb also demonstrated significantly weaker hamstrings MVIC than the uninvolved limb ($P = 0.041$). The control group demonstrated weaker quadriceps MVIC and slower MRFD in the nondominant limb ($P = 0.024$ and $P = 0.026$, respectively) versus the dominant limb.

Effect sizes for between-limb differences in the ACL-reconstructed group were larger for vertical than for resultant acceleration at the waist, but both exceeded 0.80 (Table 5.5). Conversely, effect sizes for acceleration at the shank were small-to-medium with wide confidence intervals that crossed zero. Effect sizes for waist acceleration were similar to those observed for quadriceps normalized MVIC and MRFD and slightly exceeded that observed for the leading limb peak knee extension moment. Effect sizes for the overall peak vertical GRF, peak vertical GRF at heel strike, leading sagittal plane knee excursion, and trailing limb peak knee extension moments all exceeded 1.15 with lower bounds above 0.60. Trailing limb peak knee extension moment demonstrated the largest effect size of any variable, with Cohen's *d* = 1.41.

	ACLR Involved	**ACLR Uninvolved**	**Control Nondominant**	**Control Dominant**
Normalized Quadriceps MVIC (N/kg)	9.28 ± 3.07 *	11.29 ± 2.96	10.71 ± 1.69	11.72 ± 1.73 †
Quadriceps MRFD (N/s)	8048 ± 3238 *	9857 ± 4005	7691 ± 2119	9240 ± 2390 †
Hamstrings MVIC (N)	373.9 ± 102.1 *	422.7 ± 119.7	369.5 ± 115.3	341.2 ± 98.0
Hamstrings MRFD (N/s)	4729 ± 2144	5241 ± 2979	3835 ± 1799	3698 ± 1413

Table 5.4: Differences between limbs and groups for parameters of neuromuscular performance

Values are mean ± standard deviation

MVIC = Maximal Voluntary Isometric Contraction

MRFD = Maximal Rate of Force Development

*$P < 0.05$ for ACLR subjects' Involved limb when compared to the Uninvolved limb

†$P < 0.05$ for Control subjects' Dominant limb when compared to the Nondominant limb

Variable	Effect size (Cohen's *d*)	95% CI
Peak Vertical Acceleration Waist	0.96	0.40, 1.51
Peak Resultant Acceleration Waist	0.87	0.32, 1.42
Peak Vertical Acceleration Shank	0.51	-0.05, 1.06
Peak Resultant Acceleration Shank	0.43	-0.13, 0.98
Peak VGRF at Heel Strike	1.30	0.75, 1.86
Peak VGRF (overall)	1.18	0.63, 1.74
Trailing Limb Peak KEM	1.41	0.85, 1.96
Leading Limb Peak KEM	0.81	0.26, 1.36
Leading Knee Sagittal Excursion	1.24	0.69, 1.80
Quadriceps Normalized MVIC	0.89	0.34, 1.44
Quadriceps MRFD	0.84	0.29, 1.40
Hams MVIC	0.58	0.03, 1.14

Table 5.5: Between-limb effect sizes for Acceleration, vertical ground reaction force, neuromuscular performance variables, and kinetic and kinematic variables of interest in ACL-reconstructed subjects (Cohen's *d*)

VGRF = Vertical Ground Reaction Force KEM = Knee Extension Moment

MVIC = Maximum Voluntary Isometric Contraction

MRFD= Maximum Rate of Force Development

Limb Symmetry: In the ACL-reconstructed group, mean limb symmetry indices for vertical and resultant acceleration at the waist while stepping down exceeded those measured at the shank (84.9% and 83.4% versus 90.5% and 95.0%, respectively). The LSIs at the waist were similar in value to quadriceps index (82.8%) and LSIs observed for quadriceps MRFD (85.2%), peak vertical GRF at heel strike (82.9%), and overall peak vertical GRF (84.4%).

Comparison of limb symmetry indices between groups for the walk and step down task revealed that the ACL-reconstructed group demonstrated significantly less symmetry (i.e., lower LSIs) than the Control group for peak vertical and resultant acceleration measured at the waist (*P* = 0.01 and *P* = 0.006, respectively) and peak vertical acceleration measured at the shank (*P* = 0.006; Figure 5.5). Significant differences in LSI between subject groups were also observed for the trailing limb peak knee extension moments (*P* = 0.003), overall peak vertical GRF (*P* = 0.001), peak vertical GRF at heel strike (*P* < 0.001), and hamstrings MVIC (*P* = 0.009). For all significant differences, the ACLR group demonstrated significantly less symmetry than the control group.

No significant differences were observed between LSIs for leading limb peak knee extension moments when all subjects were analyzed. However, one extreme outlier in each group skewed group means and increased variabilities to an extent that trends from the rest of the groups were obscured (the ACL-reconstructed outlier had LSI of 6.09 and the Control group outlier had LSI of -24.18). Therefore, these two subjects were removed for data analysis pertaining to the limb symmetry index for leading limb peak knee extension moment. After these two subjects were excluded from analysis, the differences between groups in leading limb peak knee extension moments LSIs became significant (*P* = 0.003). After the removal of the outliers, the LSI for leading limb peak knee extension moment became nearly identical to that of the trailing limb.

Relationships between acceleration and other variables of variables of interest: Vertical and resultant acceleration measured at the waist were positively associated with peak knee extension moments of the leading limb and negatively associated with peak knee extension moments of the trailing limbs (Table 5.6). These associations were fair to moderate in strength. Vertical and resultant acceleration measured at the waist were positively and strongly associated with peak vertical GRF at heel strike, and negatively associated with the timing of peak vertical GRF. The strengths of these relationships exceeded those between peak acceleration at the waist and peak vertical ground

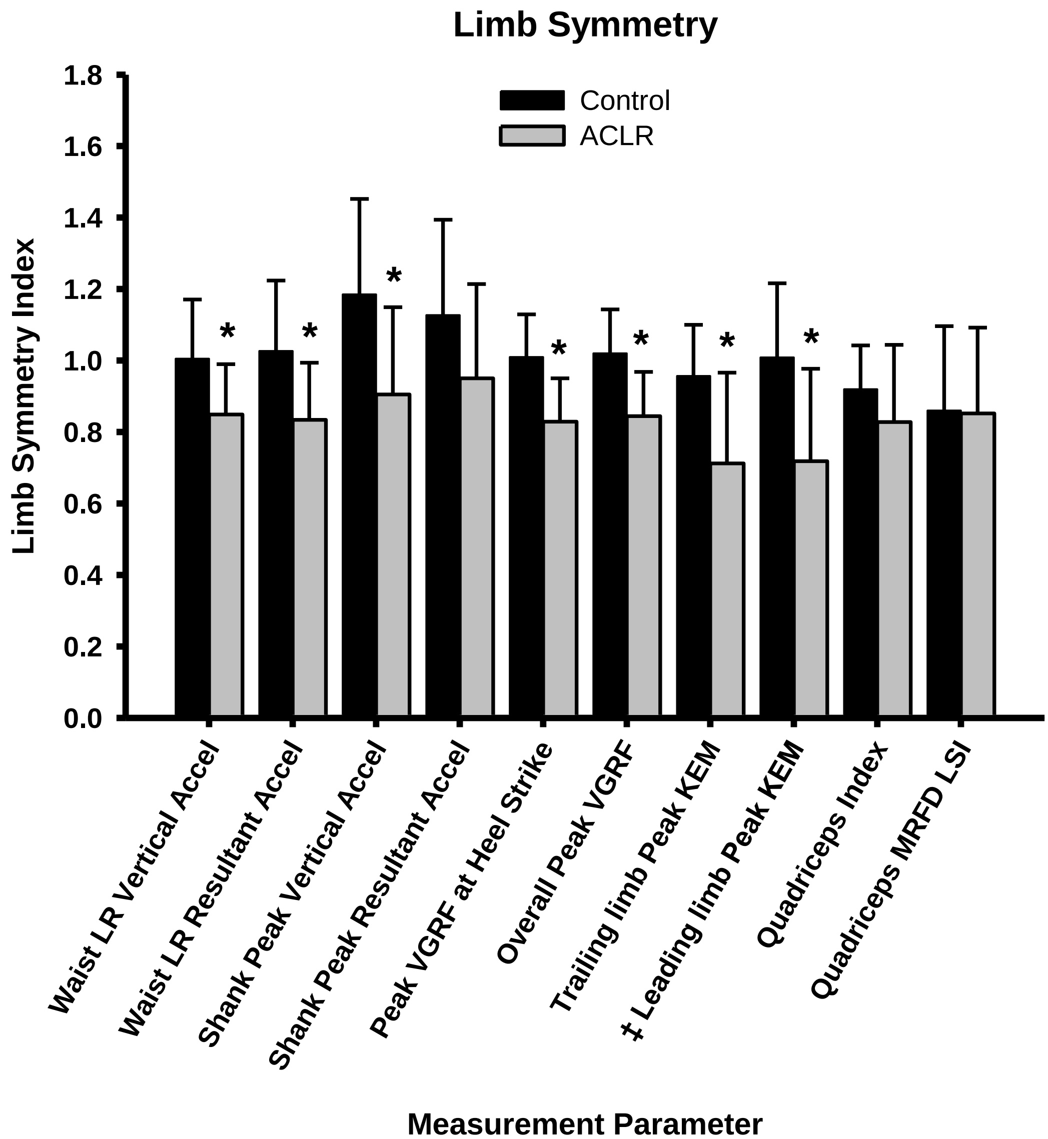


Figure 5.5: Limb Symmetry indices for acceleration, biomechanical, and neuromuscular variables of interest

‡ Outliers removed for leading limb Peak KEM

* P < 0.05 for ACL-reconstructed group when compared to Control group

LR = Loading Response VGRF = Vertical Ground Reaction Force

KEM = Knee Extension Moment

MRFD = Maximal Rate of Force Development

	Waist Vertical Acceleration	Waist Resultant Acceleration	Shank Vertical Acceleration	Shank Resultant Acceleration
Lead Peak KEM	0.536 †	0.531 †	0.189	0.282 *
Trail Peak KEM	-0.444 †	-0.48 †	-0.394 †	-0.481 †
Lead Peak KFA	0.369 †	0.388 †	0.386 †	0.418 †
Lead Sagittal Knee Excursion	0.238 †	0.257 †	0.473 †	0.504 †
Peak VGRF at Heel Strike	0.741 †	0.764 †	0.602 †	0.640 †
Peak VGRF at Loading Response	0.286 *	0.281 *	0.352 †	0.419 †
Overall Peak VGRF	0.664 †	0.682 †	0.558 †	0.584 †
Peak VGRF Timing (% Stance)	-0.466 †	-0.492 †	-0.512 †	-0.470 †
Lead Quadriceps Normalized MVIC	0.115	0.116	0.082	0.074
Lead Quadriceps MRFD	0.103	0.085	0.110	0.185
Lead Hamstrings MVIC	-0.057	-0.064	-0.250	-0.19
Trail Quadriceps Normalized MVIC	0.056	0.058	0.004	-0.026
Trail Quadriceps MRFD	0.168	0.092	0.041	0.089

Table 5.6: Pearson's Product Moment Correlations between peak accelerations at the waist and shank during the walk and step down task, kinematic and kinetic parameters of interest, and neuromuscular performance for all subjects (n = 60 limbs)

MVIC = Maximal Voluntary Isometric Contraction

MRFD = Maximal Rate of Force Development

*P < 0.05 †P < 0.01

reaction force at loading response. In addition, these associations exceeded those detected between acceleration measured at the shank and the same variables, with the exception of peak vertical GRF at loading response. All associations between acceleration at the waist and parameters of neuromuscular performance for the leading and trailing limbs were not statistically significant from zero. Although peak resultant acceleration at the shank demonstrated statistically significant associations with normalized quadriceps MRFD of the leading and trailing limbs, the strength of these relationships were fair.

Peak vertical GRF at heel strike and overall peak vertical GRF were negatively and moderately associated with trailing limb peak knee extension moment ($r = -0.516$ and $r = -0.481$, respectively; Table 5.7). In contrast, peak vertical GRF at loading response was positively associated with leading limb peak knee extension moment and lead peak knee flexion angle ($r = 0.431$ and $r = 0.541$, respectively). Associations between the leading limb quadriceps normalized MVIC and overall peak vertical GRF, and peak vertical GRF at heel strike were fair in strength ($r = 0.328$ and $r = 0.300$, respectively). These were the only relationships between parameters of vertical ground reaction force and thigh neuromuscular performance that were not significantly different from zero.

The limb symmetry index for peak vertical GRF at heel strike of the walk and step down task was significantly and strongly associated with symmetry indices for peak vertical and resultant acceleration at the waist ($r = 0.724$ and 0.791, respectively; Table 5.8). These relationships exceeded those between symmetry indices for overall peak vertical GRF and acceleration at the waist. Associations between symmetry indices in peak vertical GRF and peak acceleration at the shank ranged between 0.40 and 0.50. In addition, the limb symmetry index of peak vertical GRF timing was moderately

	VGRF at Heel Strike	VGRF at Loading Response	Overall Peak VGRF	Peak VGRF Timing
Lead Peak KEM	0.151	0.431 †	0.215	-0.260
Trail Peak KEM	-0.516 †	-0.164	-0.481 †	0.410 †
Lead Peak KFA	0.109	0.541 †	0.148	0.093
Lead Sagittal Knee Excursion	0.345 †	0.254	0.306 *	-0.237
Lead Quadriceps Normalized MVIC	0.300 *	0.133	0.328 *	-0.145
Lead Quadriceps MRFD	0.235	0.061	0.23	-0.164
Lead Hamstrings MVIC	-0.198	-0.013	-0.232	0.083
Trail Quadriceps Normalized MVIC	0.045	-0.011	0.060	-0.137
Trail Quadriceps MRFD	0.004	-0.155	-0.040	-0.095

Table 5.7: Pearson's Product Moment Correlations between parameters of Vertical Ground Reaction Force (VGRF) during the walk and step down task, kinematic and kinetic parameters, and neuromuscular performance for all subjects (n = 60 limbs)

MVIC = Maximal Voluntary Isometric Contraction

MRFD = Maximal Rate of Force Development

*P < 0.05 †P < 0.01

	LSI Waist Vertical Acceleration	LSI Waist Resultant Acceleration	LSI Shank Vertical Acceleration	LSI Shank Resultant Acceleration
‡ LSI Lead Peak KEM	0.397 *	0.590 †	0.535 †	0.637 †
LSI Trail Peak KEM	0.486 †	0.585 †	0.599 †	0.664 †
Quadriceps Index	0.314	0.328	0.242	0.322
LSI Quadriceps MRFD	-0.056	0.098	0.104	0.209
LSI Peak VGRF at Heel Strike	0.724 †	0.791 †	0.495 †	0.402 *
LSI Peak VGRF at Loading Response	-0.03	0.103	0.326	0.368 *
LSI Overall Peak VGRF	0.524 †	0.609 †	0.426 *	0.337
LSI Peak VGRF Timing	-0.525 †	-0.523 †	-0.222	-0.128
LSI Sagittal Knee Excursion	-0.051	0.115	0.267	0.230
ACL-RSI	0.425 *	0.466 †	0.414 *	0.317
IKDC	0.438 *	0.512 †	0.539 †	0.461 *
TSK-11	0.089	0.149	-0.053	0.07
Pain VAS	-0.321	-0.344	-0.238	-0.125
Marx Activity	0.133	0.12	-0.123	-0.101
Global Knee Rating	0.35	0.434 *	0.505 †	0.415 *

Table 5.8: Pearson's Product Moment Correlations between Limb Symmetry Indices (LSIs) for peak accelerations at the waist and shank, kinematic and kinetic parameters of interest, and neuromuscular performance parameters of interest for all subjects observed during the walk and step down task (n = 60 limbs)

‡Outliers removed for leading limb Peak KEM

**P*<0.05 †*P*<0.01

associated with acceleration at the waist, but not the shank. These results indicate that limb symmetry indices for parameters of vertical ground reaction force are more strongly associated with limb symmetry indices in acceleration measured at the waist than at the shank. In contrast to symmetry indices for parameters of vertical GRF, symmetry indices for leading and trailing peak knee extension moments were more strongly associated with limb symmetry in acceleration measured at the shank vs. the waist.

Significant and moderate relationships existed between limb symmetry indices for acceleration and IKDC scores (range 0.438 to 0.539). Fair to moderate relationships also were detected between acceleration and ACL-RSI scores (range 0.317 to 0.466) and Global Knee Rating (range 0.350 to 0.505).

Regression: Results from linear regression analysis with peak acceleration as the dependent variable indicate that more variability in peak acceleration was explained by peak vertical GRF and the timing of peak vertical GRF than for leading and trailing limb peak knee extension moments (Table 5.9). In addition, both sets of predictor variables explained more variability in peak resultant rather than vertical acceleration. These trends were consistent at both the waist and shank locations. When the predictor variables were GRF parameters, the regression model explained more variability in peak waist acceleration than peak shank acceleration. For acceleration measured at the waist, the coefficients of variation when GRF parameters were predictor variables were above 0.50 (R^2 = 0.550 and 0.584 for vertical and resultant acceleration, respectively). The effect sizes of these models were large (Cohen's f = 1.11 and 1.18, respectively). In contrast, when the predictor variables were knee extension moments, the regression model best explained variability in peak resultant acceleration at the shank (R^2 = 0.413, Cohen's f = 0.68).

Regression Equation	R^2	Effect size (*f*)
Waist Vert Accel = 64.20 + [2.24*(Lead PKEM)] – [13.89*(Trail PKEM)]	0.206	0.51
Waist Res Accel = 76.64 + [1.83*(Lead PKEM)] – [17.31*(Trail PKEM)]	0.235	0.55
Waist Vert Accel = -17.24 + [31.52*(PVGRF HS)] + [0.138*(PVGRF Timing)]	0.550	1.11
Waist Res Accel = -21.01 + [36.84*(PVGRF HS)] + [0.09*(PVGRF Timing)]	0.584	1.18
Shank Vert Accel = 105.00 + [9.57*(Lead PKEM)] - [25.37*(Trail PKEM)]	0.196	0.49
Shank Res Accel = 135.23 + [17.20*(Lead PKEM)] - [37.81*(Trail PKEM)]	0.319	0.68
Shank Vert Accel = 15.54 + [39.03*(PVGRF HS)] - [1.22*(PVGRF Timing)]	0.385	0.79
Shank Res Accel = -8.35 + [59.49*(PVGRF HS)] - [0.59*(PVGRF Timing)]	0.413	0.84

Table 5.9: Regression equations with waist and shank accelerations as the dependent variable during the walk and step down task. Effect sizes are expressed as Cohen's *f*

HS PVGRF = Peak vertical ground reaction force at heel strike

PVGRF Timing = Timing within stance of peak vertical ground reaction force

Lead PKEM = Peak knee extension moment of the leading limb

Trail PKEM = Peak knee extension moment of the trailing limb

Discussion

Differences between limbs and groups: The primary aim of this study was to identify the ability of wearable accelerometer sensors to detect movement asymmetry in two functional common daily activities--level walking and walking and stepping down. Our results indicate that accelerometers worn at the waist were able to detect asymmetry between limbs when stepping down during ongoing gait, but not during level walking. Peak acceleration measured at the waist when stepping down onto the uninvolved limb was very similar to peak acceleration at the waist for control subjects. Thus, the adaptation leading to between-limb differences was primarily due to lower acceleration onto the involved limb. Effect sizes in acceleration observed between limbs for ACL-reconstructed subjects when stepping down were large, in particular for peak vertical acceleration measured at the waist during loading response. The peak at loading response is the second peak observed when inspecting data from an accelerometer at the waist during a step down (Figure 4). This also corresponds to the overall peak in vertical acceleration, thus making the identification of this event simple.

During level walking, peak vertical ground reaction force onto the uninvolved limb was greater than onto the involved limb. The direction of this relationship between limbs was similar to that observed during stepping down. But, the differences in peak vertical ground reaction force detected during level walking were relatively small (4% of body weight), and were consistent across both groups of subjects. This difference may represent normal within-subject variability. Thus, the clinical utility of these differences is debatable. In contrast, only the ACL-reconstructed group demonstrated a between-limb difference in sagittal plane knee excursion during level walking, with the involved knee undergoing less motion than the uninvolved knee. The observed difference of approximately 4 degrees was comparatively large (25% of the mean), exceeds recommendations for a clinically-important difference, and is similar to previous reports of knee kinematics in the literature (Roewer et al., 2011, Lewek et al., 2002, Webster et al., 2005). Unlike walking and stepping down, peak acceleration at the waist was similar between limbs during level walking. This finding suggests that accelerometers may not be appropriate for detecting the subtlety of movement asymmetry typically observed during level walking.

The between-limb differences observed in peak knee extension moments and sagittal plane knee motion were large in this study, especially for the trailing limb peak knee extension moment and sagittal plane knee excursion during loading response. The peak knee extension moments we observed for the leading limb were very similar to those reported for healthy and high-functioning ACL-deficient subjects by Houck and Yack (2003), who investigated leading limb mechanics during the walk and step down task from the same height (20 cm). Compared to Houck and Yack, we observed peak knee flexion angles that were systematically lower by about 5 degrees. But, the involved knees in both studies demonstrated 4 to 5 degrees less flexion compared to healthy knees. Thus, the relationship between knees was similar. Our results for leading limb mechanics are comparable with those from two other studies that reported mean internal knee extension moments ranging from 1.3 to 2.0 Nm/kg for healthy subjects stepping down from a 10 cm platform (Barbieri et al., 2014, van Dieen et al., 2008).

The results from this study for trailing limb mechanics indicate that trailing knee extension moments exceeded those from the leading limb. In addition, between-limb differences in trailing peak knee extension moment demonstrated the largest effect size in ACL-reconstructed subjects. No other studies to my knowledge have reported on trailing limb mechanics during the walk and step down task, making direct comparison with previous research difficult. But, previous research has described trailing limb mechanics during step-over stair descent. These studies reported peak internal knee extension moments in healthy subjects during the second half of stance that range from 1.0 Nm/kg to 1.4 Nm/kg for descent of stairs ranging in height from 15 cm to 25 cm (Spanjaard et al., 2008, Cluff and Robertson, 2011, Beaulieu et al., 2008, Novak and Brouwer, 2011, McFadyen and Winter, 1988). These knee extension moments are similar to what we observed in the ACL-reconstructed group (mean 1.36 Nm/kg). But, these estimates were less than what we observed in healthy limbs from control subjects (means 1.87 and 1.78 Nm/kg) and ACL-reconstructed subjects (1.98 Nm/kg). The discrepancy between trailing knee extension moments during step-over stair descent reported in the literature and the walk and step down task in this study suggests that the walk and step down task places a higher demand on the trailing limb than step-over stair descent. An explanation for this is not conclusive from this study, but may be related to the methods prescribed in this study. By mandating a heel strike we increased demand on the trailing limb by effectively increasing the height of the step. By requiring a constant cadence while walking and stepping down, this likely led to higher velocity when compared to descending stairs without such methodological controls.

The between-limb differences in peak vertical ground reaction force observed during the walk and step down task in our study were large. Because we standardized the method of foot strike in this study, we quantified peak vertical GRF in three fashions: at heel strike, during loading response, and the overall peak (defined as the maximum of the two). The relative amplitudes of peak vertical GRF at heel strike and during peak loading response were variable between subjects. However, the majority of the ACL-reconstructed and healthy groups demonstrated higher vertical GRF at heel strike (8 and 11 subjects, respectively). Furthermore, this preference was typically similar across both limbs. Our results indicate that the between-limb differences in vertical GRF were most easily detected by peak vertical GRF at heel strike. The between-limb differences in peak vertical GRF at heel strike in the ACL-reconstructed group appeared related to adaptations bilaterally, although these were statistically insignificant. When compared to the Control group, ACL-reconstructed subjects stepped harder onto the uninvolved limb, and softer onto the ACL-reconstructed limb.

For the ACL-reconstructed group in this study, the between-limb effect size for overall peak vertical VGRF was larger than the effect size observed for peak vertical VGRF in ACL-reconstructed subjects at 6 months from Chapter 3 (Cohen's *d* = 1.18 and 0.89, respectively). This is an interesting finding considering that subjects in this study stepped off a shorter platform (20 cm versus 25.4 cm) and were, on average, several months further out from surgery compared to those in Chapter 3. This phenomenon can be explained by the different methods used between studies. In Chapter 3, subjects were allowed to use their preferred strategy for walking and stepping down. In contrast, we controlled gait cadence and method of foot strike in this study. Thus, leaving these factors uncontrolled seemed to modulate potential between-limb differences in peak vertical ground reaction force.

Limb Symmetry: In the ACL-reconstructed group in this study, mean limb symmetry indices for vertical and resultant acceleration measured at the waist, peak overall vertical GRF, and peak vertical GRF were all between 80% and 85%. This is very similar to the asymmetry observed for quadriceps MVIC and MRFD, adding face validity to these measures. However, limb symmetry indices for peak knee extension moments for the trailing leg and the leading leg (with outliers removed) were significantly lower than for acceleration or vertical GRF. Thus, it appears that neither peak acceleration nor peak vertical GRF are as sensitive in detecting between-limb asymmetry during the walk and step down task as knee extension moments computed from a motion capture system.

Nevertheless, many return-to-sport criteria require greater than 85% or 90% between-limb symmetry in quadriceps strength and functional hop testing to return to sport, yet do not test for movement biomechanics (Ardern et al., 2011, Kvist, 2004, Barber-Westin and Noyes, 2011, Hartigan et al., 2010, Thomee et al., 2011). Therefore, wearable accelerometer sensors may be able to provide much-needed information on functional movement symmetry to clinicians and patients discussing full return to activity.

Controlling the execution of the walk and step down task has drawbacks. First, it decreases the ecological validity of the task by mandating a method of foot strike and/or gait cadence that may be unfamiliar to subjects. In addition, controlling these factors may remove potential differences in the timing of peak vertical GRF and overall stance time. It seems reasonable that both approaches have merit. Early after surgery, differences in peak vertical GRF between limbs were very large with the self-selected approach (Cohen's *d* for ACL-reconstructed subjects from Chapter 3 was 2.41 four weeks after surgery). Thus, a self-selected strategy may offer patients greater comfort with the test and still reveal large and obvious between-limb differences. But, as subjects near completion of rehabilitation and consider a full return to activity, clinicians may want to elicit greater between-limb asymmetry with the more challenging task of controlling cadence and method of foot strike. If the walk and step down task is used to track changes in performance over time with serial measurements, then clinicians also ought to standardize methods across trials.

Compared to acceleration measured at the waist, acceleration measured at the shank appeared to lack the same ability to detect between-limb differences. The lower effect sizes observed at the shank are due to two primary factors. First, there was high between-subject variability in peak shank acceleration, especially when stepping onto the uninvolved limb. Second, group means for between-limb differences were affected by noteworthy variability in technique. Specifically, five ACL-reconstructed subjects (3 males, 2 females) demonstrated lower acceleration at the uninvolved shank despite higher acceleration at the uninvolved waist, higher overall peak vertical GRF and peak GRF at heel strike when stepping onto the uninvolved limb, and lower leading limb peak knee extension moments. Thus, it appeared that in these five subjects, acceleration at that shank failed to follow the same trend as other variables of interest. After probing for an explanation to this unexpected finding, no trends were observed relating to anthropomorphic characteristics, within-subject variability during their five trials (i.e., results were not skewed by one or two outlier trials), peak vertical GRF, the timing of peak GRF, or knee extension moments. Thus, it is possible that idiosyncratic strategies are responsible for this phenomenon.

Relationships between acceleration and other variables of variables of interest: The second aim of this study was to identify relationships between peak acceleration measured at the waist and shank and underlying mechanics in the walk and step down task. Our results for peak accelerations at the waist indicate strong positive relationships with peak vertical ground reaction force at heel strike, and fair to moderate relationships with the timing of peak ground reaction force. Similarly, our results for peak acceleration at the shank indicate moderate to good relationships with peak vertical GRF at heel strike and fair to moderate relationships with the timing of peak ground reaction force. The strengths of all these relationships exceeded hypothesized values. These findings also relate well to the results from Chapter 3, where peak vertical GRF amplitude and timing best quantified asymmetry from force platform data in the walk and step down task.

I hypothesized that the trailing limb knee extension moment would be associated to a greater degree with peak acceleration at the shank than at the waist. I also hypothesized that the leading limb knee extension moment would be negatively associated to a fair extent with acceleration measured at the waist. In contrast to these hypotheses, the results indicated that leading and trailing limb peak knee extension moments were more strongly associated with peak acceleration at the *waist* than peak acceleration at the shank. The direction of the relationship between leading limb peak knee extension moment and peak waist acceleration was positive, opposite to the hypothesized direction.

While formulating these hypotheses, I expected the walk and step down task to be dominated to a greater extent by the leading limb mechanics, in particular knee flexion angle and sagittal plane knee excursion during loading response. Under this hypothesis, if the knee flexes less than normal and the knee extension moment is reduced (hallmark adaptations in ACL-reconstructed gait), less impact force is attenuated by the knee extensor muscles, and more impact force would be transmitted proximally to the pelvis. Contrary to this hypothesis, acceleration at the waist was *positively* associated with leading limb knee extension moments. This finding suggests that under the conditions prescribed in this study (mandated heel strike and cadence) the walk and step down task was dominated more by the trailing limb. Leading limb knee extension moments were determined more by increased peak vertical ground reaction force at impact than decreased knee flexion during loading response. Thus, when knee extension moments of the trailing limb were reduced, subjects demonstrated higher peak vertical ground reaction force that was transmitted proximally to the waist. So even though the uninvolved limb demonstrated higher knee extension moments, the extra force from impact was not fully dissipated by the knee extensor muscles, and instead transmitted proximally to the pelvis.

Relationships between peak acceleration and neuromuscular performance of the thigh muscles were generally poor and not significant from zero. Similarly, relationships between vertical ground reaction force and neuromuscular performance were generally not significant from zero. Such low association of acceleration and vertical ground reaction force parameters with neuromuscular performance was unexpected, especially given the demand that the walk and step down task placed on the leading and trailing limb quadriceps. The relationships observed between vertical ground reaction force and quadriceps performance were even lower in this study than those observed in Chapter 3. This is unexpected, as I anticipated that mandating a heel strike and controlling cadence would increase the demand on the trailing knee extensors while stepping down. I expected this increased demand to be accompanied by stronger relationships between acceleration, vertical GRF, and quadriceps MVIC and MRFD. The difference between studies may be due to how the populations were studied. Subjects in this study were evaluated on one occasion at or after the termination of formal rehabilitation. Their quadriceps function was more homogeneous than subjects in Chapter 3, who were tested on multiple occasions, including early after surgery when quadriceps impairment was highest. Thus, in the cohort of subjects included in this study, altered neuromuscular control strategies, rather than differences in strength, may be responsible for explaining adaptations in performance on this task (Rudolph and Snyder-Mackler, 2004, Alkjaer et al., 2003).

Regression: For linear regression analyses with multiple predictor variables, I hypothesized that peak acceleration at the shank would be better explained by changes in vertical ground reaction force than knee extension moments. Conversely, due to its placement proximal to the knee, I hypothesized that peak acceleration at the waist would be better explained by the combination of leading and trailing knee extension moments.

These hypotheses were partially supported by the results. Parameters of ground reaction force did explain more variability in acceleration at the shank than knee extension moments. But, parameters of ground reaction force explained much more variability in acceleration at the waist than at the shank. In addition, knee extension moments explained more variability in acceleration at the shank than the waist.

Peak resultant acceleration at the waist was the acceleration parameter best explained by vertical ground reaction force. For peak resultant acceleration at the waist, the difference between regression with two predictors (peak vertical GRF at heel strike and the timing of peak vertical GRF) and simple correlation with peak vertical GRF at heel strike was minimal (R^2 values of 0.584 and 0.548, respectively). Peak resultant acceleration at the shank was the acceleration parameter best explained by knee extension moments. For peak resultant acceleration at the shank, the difference between regression with two predictors (trailing and leading limb peak knee extension moments) and simple correlation with the trailing peak knee extension moment was more substantial (R^2 values of 0.413 and 0.231, respectively).

We caution readers with respect to the use of the regression equations reported in this study. These equations are not intended to be used to estimate peak knee extension moments. At this time, we recommend prudent use of accelerometers as an approach to identifying, quantifying, and tracking movement asymmetry in clinical settings, field-based environments, or in large multi-center research studies where motion capture studies are not practical or available.

We found that wearable accelerometers are capable of quantifying movement asymmetry. Data from accelerometers relate strongly to vertical ground reaction force measured by a force platform and moderately well to peak knee extension moments calculated from a 3D motion capture system. However, accelerometers should not be used as substitutes for motion capture to definitively quantify movement asymmetry. Collecting data with these 3D motion capture systems should remain the recommended approach for research studies when three-dimensional mechanics or precision is needed.

This is an important and enabling pre-clinical study. It is one of the first studies to demonstrate that using wearable accelerometer sensors to identify and quantify asymmetry in routine functional movement is feasible.

We acknowledge that our sample size was small and may have limited some analyses; particularly with shank acceleration data due to the higher variability in recordings at that location. This study is an early step in a line of research that will explore the capabilities and value of using portable sensors in mobile health applications and rehabilitation research. Our journey is just beginning and this study gives us and other researchers a solid foundation to build on.

Conclusion

This is the second study in this thesis that leverages recent improvements in wearable technology and investigates novel uses of wearable accelerometer sensors to aid clinicians in efficiently obtaining meaningful biomechanical data in clinical/field environments. Wearable accelerometer sensors have potential to provide a rich and diverse set of data capable of characterizing patient outcome in innovative ways. The specific contribution of this study was to demonstrate that wearable accelerometer sensors were able to detect movement asymmetry in the walk and step down task not typically observable by routine clinical examination. This pre-clinical study is intended to serve as a stepping stone for research that further explores the capabilities and applications of wearable sensors in clinical practice, rehabilitation research, and the broad scope of mobile health. Our goal is to create an application for mobile electronic devices that allows accelerometers and other sensors to be used in testing movement quality. Such a translatable measurement system will permit instantaneous feedback to the patient/client. The ultimate goal of this research is to help physical therapists better perform their hallmark skill — evaluating and treating abnormal human movement to improve patients' quality of life.

CHAPTER 6

SUMMARY

The overall goal of this work was to investigate how portable force plates and wearable accelerometer sensors may identify movement asymmetry in people after knee surgery. As a whole, the studies presented in this thesis make significant contribution to how these pieces of portable technology can accurately characterize movement asymmetry in simple tasks with high functional relevance. Each chapter makes a unique contribution in advancing the use of portable technology in characterizing movement asymmetry.

Walking and Stepping Down as a Simple and Relevant Functional Outcome Measure

The purpose of the study in Chapter 3 was to evaluate how people undergoing two types of common knee surgery perform the walk and step down test onto a portable force platform. The results of this study indicate that people undergoing knee surgery demonstrate movement adaptations that are easily detected with only vertical ground reaction force measurements.

> **Hypothesis 3a:** Performance of the walk and step down test will differ significantly between limbs and across time for ACL-reconstructed and Arthroscopic partial Meniscectomy subjects. When stepping onto the uninvolved limb, higher values will occur for peak vertical ground force during loading response and for the impulse of vertical ground reaction force from initial contact through the peak during loading response. Peak vertical ground reaction force during deceleration will occur earlier in stance when stepping onto the uninvolved

limb. Differences will exist at all measurement points, but will be most pronounced early after surgery.

Supported: As hypothesized, peak vertical ground reaction force was larger and occurred earlier within stance when stepping onto the uninvolved limb. The impulse of vertical ground reaction force from initial contact to loading response was also larger when stepping onto the uninvolved limb. But, due to smaller effect size, this impulse offered no advantage over peak vertical ground reaction force. These between-limb differences were consistent across all measurement points for both cohorts of subjects. Differences were most pronounced early after surgery, but differences across measurement points were only significant in the ACL-reconstructed cohort.

Hypothesis 3b: Quadriceps strength, atrophy, and quadriceps activation of the trailing limb will be moderately to highly associated with and predictive of performance on the walk and step down test ($r < -0.50$). Patient-reported outcomes will demonstrate lower association and be less predictive of changes in ground reaction force ($r < -0.25$).

Not Supported: The direction of the relationships between peak vertical ground reaction force and measurements of quadriceps function (strength, atrophy, and voluntary activation) were opposite to those hypothesized. As the quadriceps of the trailing limb became weaker and less symmetrical with the leading limb, a corresponding decrease (rather than an increase) in peak vertical ground reaction force occurred when stepping onto the leading limb. In addition, peak vertical ground reaction force amplitude and timing were more strongly associated with measurements of quadriceps function of the leading limb rather than the trailing limb. The strength of these relationships rarely exceeded 0.50, indicating that the measures of quadriceps performance in this study did not explain much variability in peak ground reaction force. I hypothesized that self-reported outcome of the trailing limb would be inversely related to peak vertical ground reaction force. But, self-reported function of the trailing limb was not significantly associated with peak ground reaction force. Rather, lower self-reported function of the leading limb was significantly but weakly associated with lower amplitude and symmetry of vertical ground reaction force ($r = 0.22$). The relationships between parameters of ground reaction force, quadriceps function, and self-reported function suggest that quadriceps function and perceived disability of the leading limb may be more important than for the trailing limb.

Single Leg Vertical Hop Measured by Wearable Accelerometer Sensors

The purpose of the study in Chapter 4 was to evaluate the reliability and validity of using wearable accelerometer sensors worn at the waist and lower leg to estimate single leg vertical hop height in healthy people and individuals after ACL reconstruction surgery. The results clearly demonstrate that wearable accelerometer sensors can be reliably and accurately used for this task.

Hypothesis 4a: Determining hop height from flight time measured by wearable accelerometers mounted at the waist or shank will be highly reliable and valid in healthy subjects and subjects after ACL reconstruction. Intraclass Correlation Coefficient values for intra-rater reliability, inter-rater reliability, and concurrent validity will exceed 0.80. Waist and shank locations will demonstrate similar reliability and validity. Analysis with Bland-Altman plots will demonstrate low systematic bias. No differences in reliability or validity will exist between injured and healthy subjects.

Supported: Calculating single leg hop height from wearable accelerometer sensors worn at the waist was highly reliable and valid when compared to the gold standard of a force platform. Coefficients for intra-rater reliability, inter-rater reliability, and concurrent validity all exceeded 0.90. Reliability and validity coefficients were high for estimating hop height at the waist and shank, and were similar for healthy and ACL-reconstructed subjects.

Hypothesis 4b: Systematic error will be less than 2 cm for both accelerometer-based methods of estimating hop height when compared to the criterion standard of a force platform. Bland-Altman 95% limits of agreement, one estimate of random error, will be approximately 6 cm.

Supported: When compared to a force platform, systematic error for calculating hop height from accelerometers worn at the waist and shank were both less than 2 cm as hypothesized. But, estimating hop height from a shank-mounted accelerometer demonstrated virtually no systematic error while estimating hop height from a waist-mounted accelerometer tended to underestimate hop height by about 1 cm. Similarly, 95% limits of agreement from Bland-Altman plots were both less than 6 cm as hypothesized. But, 95% limits of agreement were about 1 cm less for a shank-mounted accelerometer than for a waist-mounted accelerometer. Based on the measurement error obtained from this study, estimates for error range and minimal detectable change were provided in order to assist the end-user in interpreting the results. As a rule of thumb at the waist or the shank, two measurements less than 2 cm apart should not be judged as significantly different from each other, while two measurements greater than 3 cm apart should be deemed significantly different.

Hypothesis 4c: Associations between single leg vertical hop height and quadriceps performance will exceed 0.50. Associations between asymmetry in single leg vertical hop height and patient-reported outcomes will exceed 0.40.

Supported: Hop height estimated by both accelerometer locations and the force platform were most strongly associated with quadriceps strength and rate of force development. These relationships exceeded hypothesized correlation coefficients of 0.50. Strengths of association were moderate with all subjects analyzed together and good-to-excellent with ACL-reconstructed subjects alone. Regardless of the method used, hop height was more strongly associated with quadriceps maximal rate of force development than strength. The relationship between hop height symmetry and two patient-reported outcomes (IKDC scores and Pain visual analog scale) both exceeded 0.70, far exceeding the hypothesized values of 0.40. Surprisingly, these associations even surpassed those between symmetry in hop height symmetry and quadriceps strength.

Lower Extremity Mechanics Measured by Wearable Accelerometer Sensors

The purpose of the study in Chapter 5 was to determine if wearable accelerometer sensors can provide similar insight into movement asymmetry as a system of motion capture cameras and force platforms during the walk and step down test in people after ACL reconstruction surgery. The results indicate that wearable accelerometer sensors are able to detect underlying movement asymmetry when it exists in people after ACL reconstruction.

> **Hypothesis 5a:** Significant differences in acceleration at the waist and shank will exist between the ACL-reconstructed limb, the uninvolved limb, and control subjects during level walking and stepping down after ACL reconstruction. These differences will be more pronounced with stepping down compared to level walking.

Partially Supported: While walking and stepping down, peak acceleration measured at the waist was lower when stepping onto the ACL-reconstructed limb when compared to the uninvolved limb and control subjects. The effect size for peak acceleration observed between limbs for ACL-reconstructed subjects when stepping down was large. Accelerometers worn at the shank lacked the same ability as those worn at the waist to detect between-limb differences during the walk and step down task. This was due to two primary factors: 1) High between-subject variability in shank acceleration, and 2) considerable variability of technique. Acceleration measured at the waist did not differ between limbs while walking on level ground.

> **Hypothesis 5b:** Acceleration at the shank will be strongly associated with peak vertical ground reaction force of the leading leg ($r > 0.50$) and knee extension moment of the trailing leg ($r < -0.50$). Acceleration at the waist will be moderately associated with peak vertical ground reaction force ($r > 0.40$) and internal knee extension moment ($r < -0.40$) of the leading leg. Acceleration measured at the waist will be best predicted by regression analysis that incorporates strategies from both the trailing and leading limbs.

Partially supported: Peak vertical ground reaction force at heel strike was moderately associated with peak acceleration at the shank and strongly associated with peak acceleration measured at the waist. Both of these relationships exceeded hypothesized values. But, peak vertical ground reaction force was more strongly related to acceleration measured at the waist than at the shank, which was opposite to what was hypothesized. The relationship of peak knee extension moment of the trailing limb was weaker than hypothesized with acceleration at the shank but stronger than hypothesized with acceleration at the waist. As hypothesized, peak knee extension moment of the leading limb was moderately associated with acceleration at the waist. Regression analysis indicated that ground reaction force variables better predicted acceleration at the waist than at the shank. Conversely, knee extension moments of both limbs better predicted acceleration at the shank than at the waist. I had hypothesized the opposite in both accounts. Based on the proximity of the shank accelerometer to the ground, I expected shank acceleration would be more strongly related to ground reaction force. Conversely, due to its placement proximal to the knee, I expected waist acceleration to be more strongly related to the combination of leading and trailing limb knee extension moments. Although a definitive explanation for these findings is unclear, I suspect that idiosyncratic strategies were more apparent at the shank than the waist, thus modulating overall trends in the group.

Summary

Wearable sensors have exploded in popularity and have driven a revolution in consumer-oriented personal monitoring. This technology has potential to transform healthcare by providing clinicians, patients, and scientists with unique and complementary information not typically gleaned during routine health care visits. Wearable accelerometer sensors hold promise in characterizing specific movement biomechanics in ways not currently possible in non-laboratory situations. Investigation of wearable accelerometer for this purpose is in its infancy. People after knee surgery demonstrate characteristic adaptations in sagittal plane mechanics that can persist long after post-operative rehabilitation is complete. The main purpose of the series of studies in this thesis is to explore novel uses of technology that can provide clinicians and scientists clinically feasible, low cost approaches to obtain meaningful information about functional limb symmetry in patients with knee injuries.

Chapter 3 demonstrated that vertical ground reaction force from a single force platform can detect differences in performance when people undergoing knee surgery step down while walking, a common functional task throughout the lifespan. Asymmetries were greater in those with higher quadriceps neuromuscular impairment.

Chapters 4 and 5 investigated the use of wearable sensors containing tri-axial accelerometers as a clinically feasible approach to identifying biomechanical events and characterizing asymmetry when performing common functional tasks after rehabilitation from ACL reconstruction surgery. Chapter 4 established that wearable accelerometers can reliably and accurately estimate single leg vertical hop height, a functional test that is highly sensitive for detecting between-limb asymmetry. This is the first study to our knowledge to determine that accelerometers can estimate single leg hop height equally well in healthy and ACL-reconstructed limbs. Chapter 5 demonstrated that accelerometer sensors can detect differences in performance while when stepping down while walking. Peak acceleration is strongly associated with peak vertical ground reaction force of the leading limb, and is moderately associated with peak knee extension moments, especially of the trailing limb.

The use of wearable sensors in mobile health is a rapidly developing topic of interest in medicine. Wearable accelerometer sensors have potential to provide a rich and diverse set of data capable of characterizing patient outcome in innovative ways. The studies in this thesis are among the first to use wearable accelerometer sensors to detect limb symmetry in functional movement. These pre-clinical studies were intended to create a foundation on which future research can build. We expect the studies in this thesis to enable research that further explores the capabilities and applications of wearable sensors in clinical practice, rehabilitation research, sports science, and the broad scope of mobile health. Our goal is to create an application for mobile electronic devices that allows accelerometers and other sensors to be used by clinicians and rehabilitation researchers in testing physical capacity and movement quality in a range of applications including the assessment of an injured athlete’s readiness for return to sports participation.

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