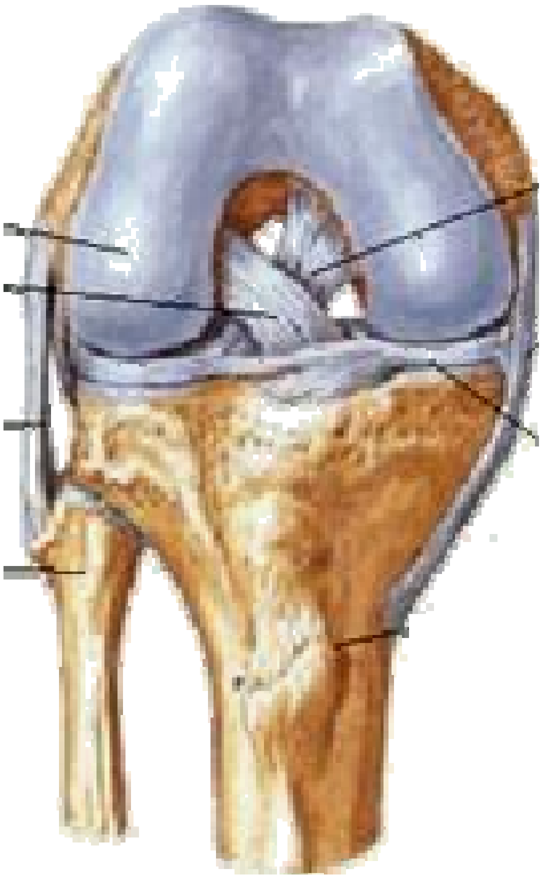
Knee Kinematic and Kinetic Characteristics of Landing after Hop:

ACL injured Subjects Before and After Rehabilitation



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Abstract

Background: Dynamic knee stability strategies of anterior cruciate ligament deficient (ACL) subjects have previously been reported in gait analysis studies. Few studies have investigated these strategies during a more strenuous sport related activity such as a single leg hop. Joint stiffness may be an important functional performance parameter, to our knowledge no one have looked into knee joint stiffness after an ACL injury. Furthermore, too few studies have examined changes in dynamic knee stability strategies after an ACL injury and a specific rehabilitation program.

Objectives: The purpose of this study was to elucidate different movement patterns, in the lower extremity, and knee joint stiffness of injured compared to uninjured side, during landing after a single leg hop in ACL injured subjects classified as non-copers, before and after a 20 session rehabilitation program.

Material and Methods: Twenty-two ACL injured subjects classified as non-copers (8 females and 14 males, no more than 6 months since injury, with a mean age of 25 years, participated in a 20 session rehabilitation program. Prior to rehabilitation, and after completing rehabilitation, subjects were examined using isokinetic quadriceps muscle strength tests at 60˚ pr.sec. (Cybex 6000), and 3-dimensional motion analysis testing (Proreflex) embedded with three forceplates (AMTI) during a single leg hop. Three single leg hop trials were average for subsequent analysis. The average joint stiffness of the knee was determined from the ratio of the change in net muscle moment to joint angular displacement in the sagittale plane between the minimum moment and the first peak moment at initial impact phase.

Results: At baseline there were significant less quadriceps muscle strength, reduced knee ROM and knee extension moment in the injured leg compared to the uninjured leg. Significant higher contribution from the hip and ankle moments and less from the knee moment, to the total support moment in the injured leg compared to the uninjured leg. Significant lower knee joint stiffness in the injured leg compared with the uninjured leg during landing after a single leg hop. After rehabilitation there was significantly improved quadriceps muscle strength, no difference in knee ROM between the legs, but the noncopers still there was reduced extension moment in the knee. After rehabilitation there were more contribution from the knee and less from the ankle moment to the total support moment, and there were no difference in knee joint stiffness between the legs.

Conclusion: The non-copers revealed different movement strategies in the injured leg compared with the uninjured leg during landing after hop. This rehabilitation program significantly improved quadriceps strength, induced changes in the lower extremity, but not all parameters were normalized.

Keywords: ACL injury, non-copers, movement patterns, joint stiffness, single leg hop, rehabilitation

Table of contents

Acronyms

Anterior cruciate ligament	ACL
Activities of Daily Living	ADL
Electromyography	EMG
Central Nervous System	CNS
Initial contact	IC
Knee Outcome survey of Activity of Daily Living Scale	KOS-ADLS
Number Crunches Statistical System	NCSS
Neuromuscular training	NMT
Peak knee flexion	PKF
Randomized controlled trial	RCT
Range of motion	ROM
Repetition of maximum	RM
Vertical ground reaction force	vGRF
Hertz	Hz
Kilogram	kg
Meter	m
Millimeter	mm
Newton	N

Definitions

Biomechanics	The science that studies the influence of forces on living bodies (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004).
Copers	Subjects who have returned to full activity without symptoms of instability for at least 1 year (Rudolph, Eastlack, Axe, & Snyder-Mackler, 1998; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001)
Dynamic knee stability	The ability of the knee joint to remain stable during physical activity (Williams, Chmielewski, Rudolph, Buchanan, & Snyder-Mackler, 2001).
Electromyography	The recording of electrical potentials produced by skeletal muscles (Robertson et al., 2004).
Joint stiffness	The change in joint moment divided by the change in joint angle (Farley, Houdijk, Van, & Louie, 1998b) during initial impact phase.
Kinematics	The study of motion without regard to its causes or quantities of motion, such as velocity, speed, acceleration, angular displacement (Robertson et al., 2004).
Kinetics	The study of the causes of motion; the study

	of forces and moments of force and their characteristics such as work, energy, impulse, momentum, power (Robertson et al., 2004).
Laxity	Slackness or lack of tension, looseness, characteristic of the ACL, referring to a abnormal tibiofemoral translation of the knee joint (Noyes, Grood, & Torzilli, 1989).
Motion analysis system	A system for collecting and processing the motion of sensors or markers attached to a body (Robertson et al., 2004).
Muscle co-contraction	Simultaneous activation of agonist and antagonistic muscles during stance (Chmielewski, Rudolph, Fitzgerald, Axe, & Snyder-Mackler, 2001).
Neuromuscular control	The ability to control movements through co-ordinate muscle activation, depending upon the afferent input to the knee and the subsequent muscular feedback (Williams et al., 2001).
Neromuscular training	As training enhancing subconscious motor responses by stimulating both the afferent signals and central mechanisms responsible for dynamic joint control (Risberg, Mork, Jenssen, & Holm, 2001a).
Non-copers	Subjects who have poor dynamic knee stability with daily activities (Hurd, Axe, & Snyder-Mackler, 2008; Fitzgerald, Axe, & Snyder-Mackler, 2000a)

Potential copers	Subjects who have good dynamic knee stability and the potential to compensate well after ACL injury (Fitzgerald et al., 2000a; Hurd et al., 2008)
Proprioception	The acquisition of stimuli by peripheral mechanoreceptors, and the conversion of these mechanical stimuli into neural signals that are transmitted along afferent pathways to the central nervous system for processing (Lephart & Fu, 2000).
Stiffening strategy	A movement strategy which includes reduced knee motion, reduced internal knee extension moment, distribution of support moment away from the knee, slower muscle activation and generalized co-contraction of the muscles that cross the knee (Rudolph et al., 1998; Rudolph, Axe, & Snyder-Mackler, 2000)
Stiffness	The relationship between the deformation of a body and a given force (Butler, Crowell, III, & Davis, 2003). The combination of all the individual stiffness values contributed by muscles, tendon, ligament, cartilage and bone (Latash & Zatsiorsky, 1993).
Support Moment	The summated extensor moment (hip + knee + ankle moments) (Winter, 1980)

1. Introduction

1.1. Background

Serious knee injuries, such as anterior cruciate ligament (ACL) injuries, are a growing cause of concern. In Norway alone, the yearly incidence of ACL injuries is estimated to be 88 per 100 000 inhabitants (Granan, Engebretsen, & Bahr, 2004). These injuries may have serious consequences for the injured subject in terms of treatment, cost and time lost from activities. On the long term basis, expenses are related to possible disabilities due to an increased risk of early osteoarthritis (OA)(Myklebust & Bahr, 2005).

The principal function of the ACL is to prevent anterior translation of the tibia relative to the femur. Rupture of the ACL typically results in loss of knee joint stability, strength of the surrounding musculature, and function. However, evidence in the literature reports that there are no relationship between the increased anterior knee laxity and the knee stability during dynamic activities (Snyder-Mackler, Fitzgerald, Bartolozzi, III, & Ciccotti, 1997; Lephart et al., 1992; Eastlack, Axe, & Snyder-Mackler, 1999). There are ACL deficient individuals who can maintain high activity levels, experiencing neither instability, loss of function or weakness despite complete rupture of the ACL (Eastlack et al., 1999; Daniel et al., 1994),

A screening examination and classification system has been develop to identify those ACL injured subjects with good dynamic knee stability and the potential to cope well, (potential copers) from those who have poor dynamic knee stability (non-copers) (Fitzgerald et al., 2000a). Those individuals classified as potential copers have passed the screening examination, and have been considered as rehabilitation candidates with the potential to return to preinjury activity level without ACL reconstruction (Daniel et al., 1994; Eastlack et al., 1999). More commonly, subjects do not cope well, fail the screening examination and are termed non-copers (Eastlack et al., 1999; Daniel et al., 1994). Non-copers present greater knee instability and greater alterations in movement patterns when compared to potential copers, and healthy controls (Rudolph et al., 1998; Rudolph et al., 2000; Rudolph et al., 2001), and are, therefore, considered good candidates to study. Non-copers, decrease their injured knee motion, extensor moment and distribute the support moment away from the knee during different movement tasks (Rudolph et al., 1998; Rudolph et al., 2000; Rudolph et al., 2001). The reduced knee flexion and knee extension moment found in non-copers has been shown to relate to quadriceps weakness (Rudolph et al., 2001). Exposing non-copers to a rehabilitation program including both strength and neuromuscular exercises would be of significance, to study if they have the potential to improve their movement dysfunctions (Bruhn, Kullmann, & Gollhofer, 2006; Fitzgerald, Axe, & Snyder-Mackler, 2000b; Chmielewski, Hurd, Rudolph, Axe, & Snyder-Mackler, 2005; Risberg, Moksnes, Storevold, Holm, & Snyder-Mackler, 2009; Hartigan, Axe, & Snyder-Mackler, 2009). Most kinematic and kinetic studies on ACL injured subjects have been performed during walking, while, information regarding movement strategies during strenuous activities, such as hopping, is sparse in the literature (Risberg et al., 2009; Rudolph et al., 2000). Single leg hops represent an activity which places higher demands on the knee than walking. Therefore hopping might provide more information about knee stability during dynamic activities than less strenuous activities.

Another parameter which might give more insight into the non-copers' dynamic knee stability is stiffness. Stiffness plays an important role in determining the dynamics of the interaction between the stance leg and the ground (Farley & Ferris, 1998a), and can be considered as an integrated functional measure of both impacts and displacements. Stiffness represents the elasticity of the musculoskeletal system, and involves movement, position, muscle properties, muscle activation and coordination of the leg (Brughelli & Cronin, 2008). The greater the forces are imparted to the body, the more resistance or stiffness is required to preserve stability (Blickhan, 1989). Little stiffness gives less challenge to the leg, it distributes the forces over a longer period, and may allow excessive motion and increase the risk of soft tissue injuries (Butler et al., 2003). Excessive stiffness is associated with reduced range of motion (ROM) and increased peak ground reaction forces (GRF). The increased peak GRF increases the impact forces, which in turn may increase bone and cartilage injuries (Butler et al., 2003). To our knowledge, no studies have examined knee joint stiffness in ACL injured subjects during dynamic activities. We wanted to examine if joint stiffness would expand our understanding of knee joint biomechanics and lower extremity stability by giving more insight to the functional capacity of the leg. Recent literature has shown that knee extension moment during walking normalizes after ACL rehabilitation, but knee ROM does not (Risberg et al., 2009). It is not known how training could affect stiffness.

This has led us to examine the movement patterns during hop in ACL injured subjects classified as non-copers before and after a rehabilitation program. The data presented in this master thesis is a substudy of a longer project which has followed ACL injured subjects with standardized motion analyzes before and after a 20 session rehabilitation program between 2003 and 2007 (Risberg et al., 2009). This scientific project was performed by our research group NAR, Norwegian research center for active rehabilitation, a collaboration between Orthopaedic Center, Oslo University Hospital, Ullevaal, Hjelp24NIMI and the Norwegian School of Sport Sciences (NIH) in Oslo, Norway.

1.2. Aims and hypotheses of the thesis

The objectives of this study was twofold; (1) to examine differences in quadriceps muscle strength, lower extremity joint motion and moments and knee joint stiffness in injured and uninjured legs during landing after a single leg hop in ACL injured subjects classified as non-copers. (2) To examine changes in both injured and uninjured leg for quadriceps muscle strength, lower extremity joint motions and moments and knee joint stiffness during landing after a single leg hop following a 20 session rehabilitation program in ACL injured subjects classified as non-copers.

Hypothesis developed from objective (1):
First we hypothesized that the quadriceps muscle strength, knee range of motion (ROM) and moments will be reduced in the injured leg compared to the uninjured leg in ACL injured subjects classified as non-copers. Secondly, we hypothesized lower knee and higher hip and ankle contributions to the total support and furthermore that there will be significantly lower knee joint stiffness in the injured leg compared to the uninjured leg in ACL injured subjects classified as non-copers.

Hypothesis developed from objective (2):
First we hypothesized that there will be no significant differences in quadriceps muscle strength, knee ROM and moments between the injured leg compared to the uninjured leg in ACL injured subjects classified as non-copers, after a 20 session rehabilitation program. Secondly, we hypothesized that the total support moment and knee joint stiffness in the injured leg would be more normalized as compared to the uninjured leg in ACL injured subjects classified as non-copers, after a 20 session rehabilitation program.

2. Theoretical background

2.1. Knee joint

The human knee is one of the most complex systems in the body. The knee is the articulation between the femur, tibia, and patella and includes menisci and ligaments. It functions in a way which transmit load between muscle, ligament and bone tissue (Nordin & Frankel, 2001).

The main movement between femur and tibia is flexion-extension, but rotation and varus-valgus movement also take place between the bones (Bojsen-Møller, 1996). Normal function of the knee would be impossible without the complex neurological components providing sensory innervations, including proprioception, as well as active muscle control (Lephart et al., 2000). The stability of the knee joints depend on a fine balance between ligaments, articular contact and muscles (Butler, Noyes, & Grood, 1980; Noyes, Grood, Butler, & Malek M, 1980). The anterior cruciate ligament (ACL) is the most important passive stabilizer in the knee joint (Figure 1).

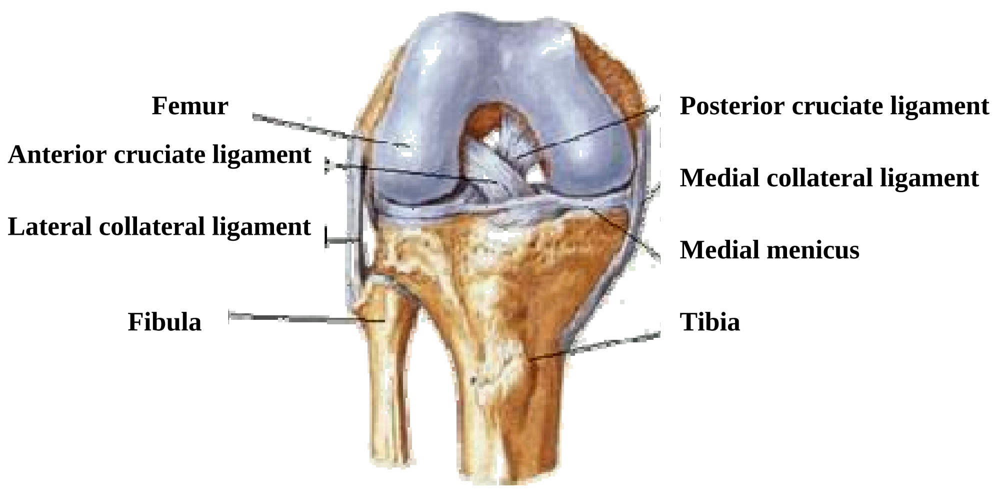

Figure 1 The structure of the knee (the right knee in a frontal view). The position of the knee is straight and the patella and the joint capsule are removed. The figure is derived and modified from Netter (Netter, 1995).

2.1.1. The anterior cruciate ligament

The anterior cruciate ligament (ACL) is one of the four major ligaments of the knee. It is a band of regularly oriented dense connective tissue that connects the femur and the tibia. The ligament is attached to the posterior aspect of the medial surface of the lateral femoral condyle. On the tibia the ligament is attached to a fossa in front of and lateral to the anterior tibial spine. The ACL passes anteriorly, medially, and distally from the femur to the tibia (Odensten & Gillquist, 1985). The ACL structure is externally rotated approximately 90 degrees to its insertion (Samuelson, Drez, Jr., & Maletis, 1996) and has a slight spiral form caused by it`s bony attachments (Odensten et al., 1985). The bony attachments are also responsible for the relative tension throughout the range of motion (Girgis, Marshall, & Monajem, 1975). The ACL ligament consists of two functional synergistic structures: (1) The anteromedial (tight in flexion) and (2) the posterolateral bundle (tight in extension) (Petersen & Zantop, 2007; Zantop, Petersen, Sekiya, Musahl, & Fu, 2006). The ACL is inverted by some fibres from the tibial nerve, and there have been found mechanoreceptors in the superficial tissue of the ligament and its terminal ends (Johansson, Sjolander, & Sojka, 1991; Solomonow & Krogsgaard, 2001; Zantop et al., 2006). These findings confirm that the ACL has two major functions in maintaining normal knee joint kinematics: biomechanical and sensory functions.

2.1.2. Biomechanical function of the ACL

The function of the ACL is to stabilize the knee joint, prevent abnormal movements, and steer the movement of the knee (Nordin et al., 2001). The ACL prevents anterior translation of the tibia relative to the femur, and cadaver studies have shown that at 30 degrees of flexion the ACL represent an average of 86% of the total capsular and ligamentous resistance (Butler et al., 1980; Butler et al., 1980; Olsen, Myklebust, Engebretsen, & Bahr, 2004). In addition, experimental studies have shown that the ACL prevents hyperextension and stabilizes the knee against tibia rotation. These studies have also shown that the ACL act as a valgus and varus stabilizer at 0 to 40 degrees of knee flexion (Girgis et al., 1975; Olsen et al., 2004). The ACL is able to tolerate strain of about 20 % before rupture (Frank & Jackson, 1997; Noyes, Butler, Grood, Zernicke, & Hefzy, 1984; Butler et al., 1992). The force in the intact ACL ranges from approximately 100 N to 2000 N depending upon the tasks and motions of the knee (Johansson, Djupsjobacka, & Sjolander, 1993; Woo, Livesay, & Engle, 1992). The ACL is only loaded to about 20 % of its failure capacity during activity of daily living (ADL). On the other hand, cutting and acceleration-deceleration activities are the most demanding activities for the ACL (Hewett, Myer, & Ford, 2006). The majority of ACL injuries occur during sport activities, when the knee is impacted by unusual combinations of loading patterns for the knee such as valgus collapse, and internal or external rotation when the knee is in a flexed position (Krosshaug, 2006; Beynnon & Fleming, 1998; Beynnon et al., 1995). There has been reported differences in tibiofemoral motion of the ACL injured knee during walking with respect to the uninjured knee or healthy controls (Georgoulis, Papadonikolakis, Papageorgiou, Mitsou, & Stergiou, 2003; Andriacchi & Dyrby, 2005). Chmielewski et al (Chmielewski, Ramsey, & Snyder-Mackler, 2005) found that non-copers have a posterior tibia position relative to femur, and Andriacchi and Dyrby (Andriacchi et al., 2005) found a position of greater internal rotation and posterior translation of the tibia in the ACL injured knee during walking.

2.1.3. Sensory function of the ACL

The mechanoreceptors located in the ACL and other structures in and around the knee joint play an important role in dynamic knee stability and neuromuscular control (Johansson et al., 1991; Solomonow et al., 2001; Hewett, Zazulak, Myer, & Ford, 2005b). The mechanoreceptors provide information regarding tension and compression forces, or rate of loading of the knee (Williams et al., 2001). These mechanical stimuli are processed into neural signals transmitted along afferent pathways to Central Nervous System (CNS) (proprioception). The CNS uses this feedback information, from the muscles and joint proprioception, to obtain dynamic knee stability by processing an efferent feedback to the muscles and adjusting the position of the knee (Hewett, Paterno, & Myer, 2002; Lephart et al., 2000; Williams et al., 2001; Hewett et al., 2005b). Loss of mechanoreceptors after an injury disturbs this sensory system (Friden, Roberts, Ageberg, Walden, & Zatterstrom, 2001), with possible effects on neuromuscular function. Most studies shows decreased neuromuscular control in ACL deficient subjects (Chmielewski et al., 2005; Rudolph et al., 2000; Solomonow et al., 2001).

2.2. Knee function after ACL injury

2.2.1. Dynamic knee stability – potential copers and non-copers

ACL injury usually leads to knee joint instability that often prevents subjects from participating in sports and daily activities. Knee stability is traditionally assessed using static measurements, such as passive anterior knee joint laxity or pivot shift tests. However, studies have shown that passive anterior knee laxity is unrelated to knee stability during dynamic activities (Eastlack et al., 1999). Some individuals with a small degree of passive joint laxity experience huge dynamic instability and vice versa. Dynamic knee stability can be defined as the ability to maintain normal movement patterns while performing high-level activities without episodes of giving way of the knee (subluxation of the tibia relative to the femur) (Lewek, Chmielewski, Risberg, & Snyder-Mackler, 2003). Dynamic knee stability depends upon the interplay between muscles, tendons, ligament, bone and the CNS (Williams et al., 2001). In the absence of the ACL the task of restraint falls upon secondary ligamentous structures and muscles (McNair & Marshall, 1994). The knee joint is forced to rely on the dynamic restraints as muscle function to maintain joint stability because of the lack of bony congruence and the inability of the static restraints to handle the force generated during functional task (Lephart et al., 2000). Therefore, the dynamic knee stability is highly dependent upon neuromuscular control (Williams et al., 2001; Tagesson, Oberg, Good, & Kvist, 2008).

Neuromuscular system responses after ACL injury dictate in large part which patients are able to maintain dynamic knee stability in the absence of ligament support. An ACL rupture does not automatically give functional impairment and instability. Some patients experience major dynamic knee instability problems, while others seem to have a satisfactory dynamic knee stabilization strategy and do not experience giving way during activities (Fitzgerald, Lephart, Hwang, & Wainner, 2001).

As early as 1983 Noyes et al (Noyes, Matthews, Mooar, & Grood, 1983) estimated that approximately one third of the ACL injured subjects became better, one third had to decrease their activity level, and one third experienced instability with non-surgical treatment and had to be treated with ACL reconstruction. Later a screening examination was developed to categorize patients with ACL rupture as potential copers or non-copers based on their dynamic knee stabilization strategies (Fitzgerald et al., 2000a). Those who have the potential to compensate well for the injury are classified as potential copers, and those who experience instability are classified as non-copers (Fitzgerald et al., 2000a).

The screening examination consists of a battery of clinical tests and measures that capture dynamic knee stability (Fitzgerald et al., 2000a). A modified version consists of four functional single leg hop tests (one leg hop, triple jump test, cross-over test, and 6m time hop test), a self-report of knee function survey, the KOS-ADLS (Irrgang, 2003), a global rating of knee function, assessed by a visual analogue scale, and questioning of the number of episodes of giving-way since the initial injury. An episode of giving way is defined as a subluxation event on the knee with pain and subsequent effusion (Fitzgerald et al., 2000a). The KOS-ADLS is a region-specific questionnaire, measuring patients' functional abilities and symptoms during a variety of daily tasks (Irrgang, Snyder-Mackler, Wainner, Fu, & Harner, 1998). The questionnaire consists of 14 items with 6 possible answers (each answer weighted from 0 to 5 points) ranging from not limited or symptomatic to unable to perform the task. Results are expressed as a percentage (answered points / 70 x 100) with higher scores representing a higher level of function. The global rating scale is a single number chosen by the patient from a scale ranging from 0 to 100 %, with 100 % representing the level of function before the injury (including athletics), and 0 % indicating complete loss of function due to the knee injury.

Subjects are classified as a non-coper if they have one or more of the following: (1) less than 80 % of the uninjured leg during the four single leg hop tests, (2) KOS-ADLS less than 80 %, (3) less than 60 % on the Global Rating of Knee Function and (4) more than one episode of giving-way following the initial injury until the time of the screening examination (Fitzgerald et al., 2000b).

2.2.2. Knee joint kinematics and kinetics - movement strategies

Several studies have determined that some ACL injured subjects can adapt to the lack of the ACL over a prolonged period of time. Rudolph et al (Rudolph et al., 2000; Rudolph et al., 1998; Rudolph et al., 2001) have in a series of studies identified movement strategies in potential copers and non-copers. Potential copers had kinematics and kinetics during walking and jogging that were similar to those of uninjured subjects with normal knees. Non-copers, on the other hand, compensated for the absence of the ACL through decreased knee motion and internal knee extensor moment in the injured knee (Rudolph et al., 1998). Electromyography (EMG) suggested that hamstring/quadriceps co-contraction is seen in non-copers but not in potential copers (Rudolph et al., 2000; Rudolph et al., 1998; Rudolph et al., 2001). Winter (Winter, 1995; Winter, 1980) described a covariance of the hip and knee moment where a higher hip extensor moment was associated with lower knee extensor moment, thereby Ferber et al. (Ferber, Osternig, Woollacott, Wasielewski, & Lee, 2003) suggested that non-copers may rely on compensation from the hip musculature to lower the demand in the knee.

Non-copers are characterized by a less flexible strategy across a range of activities, which is referred to a ”stiffening strategy” in the literature (Rudolph et al., 1998; Rudolph et al., 2000; Rudolph et al., 2001). ”Stiffening strategy” involves reducing the range of motion of knee movement and knee moment, distribution of support moment away from the knee, and increasing co-contraction across the knee to statically try to stabilize the injured knee (Chmielewski et al., 2001). The reduced knee motion during weight acceptance reduces flexibility of the knee and suggests that non-copers are using a knee stabilization strategy that may be putting them at risk for future joint degeneration. Co-contraction, less joint range of motion and muscle activity may lead to less shock absorption and increased compressive and shear forces on articular cartilage in all joints during weight acceptance on articular cartilage (Rudolph et al., 2001; Olsen et al., 2004).

Previous researches of ACL injured subjects have focused on movement pattern during gait. Comparatively few researchers have evaluated movement pattern during landing, where limb stability is a major objective.

2.2.3. Landing after hop

Single leg hop tests have been used clinically to examine knee function and stability in patients following ACL rupture (Fitzgerald et al., 2001; Rudolph et al., 2000). Landing from a hop is a task that is demanded frequently in sport and places considerable stress on the ACL-deficient knee. It may reflect the integrated effect of neuromuscular control, strength and confidence in the leg. Risberg (Risberg & Ekeland, 1994) found that the single leg hop, the triple jump and the stairs hop test are sensitive in detecting changes over time in the late rehabilitation phase, and to detect knee instability in ACL injured subjects.

During landing after hop there are numerous forces that act on the body. External forces which includes gravity, friction and contact with other objects and internal forces that are caused by the forces exerted within the body itself, and are the result of muscle activation and restraint provided by ligaments, capsule and bone (Wikstrom, Tillman, Chmielewski, & Borsa, 2006). The lower limb is effectively working like a biomechanical chain, and the joints must function synchronously to distribute loading in space and time (Tillman, Hass, Chow, & Brunt, 2005). The movement of one joint influences all the other joints in the chain (Tillman et al., 2005), and it is the coordination between all of the joints that determines the effectiveness and efficiency of the movement as a whole (Hughes & Watkins, 2008). During landing after hop it is the hip, knee and ankle joints that largely determine the dynamic stability in the leg (Hughes et al., 2008). An injury to the ACL may disrupt this chain system that connects the femur and the tibia and result in increased anterior translation of the knee.

The action of landing from a single leg hop applies forces and moments to the lower extremity, accelerating hip and knee flexion and ankle dorsiflexion, and this way causes the leg to collapse. The goal of a successful landing is therefore to resist this collapse by applying counter extensor moments (deceleration forces) at these joints in such a way that the body's negative velocity is reduced to zero without injury (Tillman et al., 2005). These extensor moments primarily work eccentric to absorb kinetic energy from the skeletal system and to stop the person from falling.

Controlling the moments in the lower extremity joints when the system is influenced by large reaction and intersegmentel forces during landing is a significant challenge to the neuromuscular system. Devita and Skelly (Devita & Skelly, 1992) identified the importance of presetting the joint moments during the descent phase for the successful performance of landing. To provide joint stability and a more erect body posture at initial contact, the subjects increased their net moments at the hip and knee prior to contact. This way the neuromuscular system prepares itself for the load that will occur after foot contact by activating muscles prior to landing (McKinley & Pedotti, 1992). After contact the lower extremity must generate sufficient forces to stabilize the joints, control joint flexion, and reduce total momentum. How the subjects choose the set of muscles to control the reaction force will likely influence the mechanical loading which the lower extremity experiences during landing and their ability to adjust control (Nitt-Gray, Hester, Mathiyakom, & Munkasy, 2001).

2.2.4. Stiffness

Stiffness is a biomechanical parameter that characterises the deformation of the soft tissue structures connecting one bone to another in response to applied load or torque (Schmitz et al, 08). It is thought to be involved in both musculoskeletal performance and injury (Butler et al., 2003). In the human body, stiffness is a measure of the relationship between kinematic and kinetic, and can be related to the choice of movement solution a subject chooses. It can be described from the level of a single muscle fibre, to modelling the entire body as a mass and a spring (Farley et al., 1998b). During running and hopping, the actions of the body's many musculoskeletal elements, including muscles, tendons, and ligaments, are integrated together so that the overall musculoskeletal system acts like a single spring. Hopping can therefore be modelled by using a simple spring-mass system, consisting of a single linear "leg-spring" and a point mass that is representing body mass (Blickhan, 1989; Farley et al., 1998b) (Figure 2).

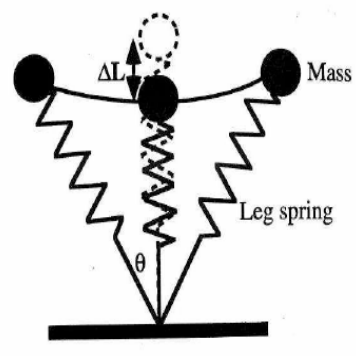

Figure 2 A spring-mass model. The model consists of a linear spring representing the leg and a point mass equivalent to body mass (Farley et al., 1998a).

During landing the lower extremity joints flex and extend, thereby compresses and lengthens the leg spring. The stiffness of the leg influences the way the body interacts with the ground. For example, a greater stiffness leads to a shorter ground contact time and a smaller vertical excursion of the body`s center of mass during the ground contact phase (Farley & Morgenroth, 1999).

By simplifying the human body as a spring, calculation of stiffness is easier. However, it is important to be aware of that the spring-mass model does not represent the real structure, only the behaviour of the whole musculoskeletal system during hopping and running. In real life, the length change of the leg spring that occurs during the ground-contact phase corresponds to joint flexing and extending. For example, during hopping in place, the ankle, knee and hip flex during the first half of the ground-contact phase and extend during the second half of the ground-contact phase (Farley et al., 1998b). For that reason, hopping is not a purely harmonic motion and human joints are not simple mechanical springs.

There are several different calculations of lower extremity stiffness, including vertical, leg and joint stiffness (Butler et al., 2003). With respect to joints, joint stiffness based on joint torque and angular displacement can be determined (Butler et al., 2003). Both vertical and linear stiffness are linear measures, and the joint stiffness correlates to these measures. While linear stiffness is defined as linear force per linear displacement, joint stiffness is most often defined as the change in joint moment divided by the change in joint angle (Farley et al., 1998b).

In the multijointed system in humans, overall stiffness depends on a combination of the joint stiffness and the geometry of the system (Farley et al., 1998b). In this manner, the stiffness in each joint combined with the position of the joints influence the overall leg stiffness.

Dynamic stability requires control of changing torque and stiffness of each joint based on feedback concerning joint angular position (Arampatzis, Bruggemann, & Klapsing, 2001a; Arampatzis, Bruggemann, & Metzler, 1999; Farley et al., 1998b; Farley et al., 1999). Several biomechanical factors control joint stiffness, including muscle activation and force, antagonist muscle coactivation and lower extremity kinematics during ground contact (Farley et al., 1999; McMahon, Valiant, & Frederick, 1987). By changing one of the factors such as loading conditions through different muscle activation and movement strategies, one can adjust lower extremity stiffness (Farley et al., 1999; Farley, Blickhan, Saito, & Taylor, 1991). When humans land following a jump, the stiffness of the landing appears to depend on the knee angle at landing (Devita et al., 1992). If the leg joints are more flexed when the foot hits the ground, the joint moments associated with a given ground reaction force will increase (Farley et al., 1998a). Thus, for a given joint stiffness, the angular displacement of the joints during the ground-contact phase also will increase. As greater displacement with a given force occurs, the result is lesser stiffness. The amount of stiffness required has been reported to increase as the demands of the activity are greater through increased hopping frequency, hopping height, and running speed. As greater forces are imparted to the body, greater resistance to movement is needed to control motion, to resist collapse, and to preserve stability of the lower extremity (Arampatzis et al., 1999; Farley et al., 1991; Granata, Padua, & Wilson, 2002). Studies have shown that subjects can consciously alter their stiffness during landing to change the forces the body experiences (Devita et al., 1992; Zhang, Bates, & Dufek, 2000; Dufek & Bates, 1990), but if training influences stiffness remains uncertain. However, no previous studies have examined knee joint stiffness during hopping in ACL injured subjects.

2.3. Rehabilitation after ACL injury

Treatments for ACL injured subjects include either rehabilitation without surgery or in combination with surgery. In both situations there are four main impairments and disabilities that are important: (1) intra articular effusion, (2) pain, (3) dynamic stability through neuromuscular control, and (4) muscular atrophy and strength (Hewett et al., 2002; Moksnes, 2007). Non-surgical treatment usually require a shorter time of rehabilitation (3-6 months), while rehabilitation after ACL-reconstruction varies from six to twelve months before returning to sports (Beynnon, Johnson, & Fleming, 2002; Kvist, 2004; Myklebust et al., 2005; Myer, Paterno, Ford, Quatman, & Hewett, 2006). Early rehabilitation that normalizes knee motion and reduces joint loading may lower the risk of consequences after an ACL injury (e.g. knee OA).

The ultimate goal of rehabilitation is to restore dynamic knee stability, regain muscular strength and range of motion (ROM), to assist the patient in a safe return to pre-injury level of activity, and reduce the risk factors for developing knee OA (Hewett et al., 2005b; Risberg, Lewek, & Snyder-Mackler, 2004). The lack of adequate neuromuscular control and quadriceps muscle strength is a common explanation for why most subjects with ACL injury fail to cope well after their injury, and are therefore often components of the rehabilitation program after the injury (Chmielewski, Rudolph, & Snyder-Mackler, 2002; Lewek et al., 2003). The literature recommends a combination of neuromuscular and strength training (Bruhn et al., 2006; Trees, Howe, Grant, & Gray, 2007).

2.3.1. Quadriceps muscle strength

Regeneration of quadriceps muscle strength is essential for success after ACL injury. It is well documented that after an ACL injury there is less quadriceps muscle activation and strength (Hurd & Snyder-Mackler, 2007; Wilk, Arrigo, Andrews, & Clancy, 1999). The muscles serve as the primary active stabilizers of the knee and shock absorbers during functional loading. ACL injured subjects with dynamic instability rely heavily on muscle function during functional activities (Keays, Bullock-Saxton, Newcombe, & Keays, 2003). A positive relationship has been found between quadriceps strength and function (Eastlack et al., 1999; Keays et al., 2003; Hurd et al., 2007). Isolated activation of the quadriceps would give a tendency towards an anterior pull of the tibia related to femur, which could lead to subluxations of the knee (Draganich, Andriacchi, & Andersson, 1987; Draganich & Vahey, 1990). Non-copers quadriceps strength has shown to influence the amount of knee flexion used during gait when tested preoperatively (Rudolph et al., 2001; Draganich et al., 1987; Draganich et al., 1990). Furthermore, there is a significant effect of continued quadriceps weakness on diminished knee angles and knee moments during walking following ACL reconstruction (Lewek, Rudolph, Axe, & Snyder-Mackler, 2002). Studies have shown that knee flexion angles greater than 45° allow the quadriceps to be an ACL agonist (decrease strain on the ACL with activation) rather than an ACL antagonist (Draganich et al., 1987; Draganich et al., 1990). Therefore, it is important to use deep knee flexion angles to put the quadriceps into an ACL agonist position. Lower quadriceps muscle activation and decreased neuromuscular control of this muscle, may therefore lead to a changed movement pattern during functional activities (Chmielewski, Hurd, & Snyder-Mackler, 2005; Hurd et al., 2007).

Insufficient quadriceps function is believed to be a consequence of both atrophy and activation failure caused by permanent alteration of muscle activation (Williams et al., 2001; Keays et al., 2003; Ingersoll, Grindstaff, Pietrosimone, & Hart, 2008). Quadriceps weakness does not only contribute to changes in movement pattern and reduced function, but may also be a contributing factor to knee OA (Hurley, 1999).

Figure 3 Squats, example of strength training.

Dose-response considerations are fundamental in muscle strength training. The American College of Sports Medicine (Kraemer et al., 2002; Ratamess et al., 2009) recommends loads for uninjured subjects corresponding to 8-12 repetitions of maximum (RM) (Anderson & Kearney, 1982; Kraemer & Ratamess, 2005), 2-3 days per week, and 1-2 min rest intervals for novice trainers (Kraemer et al., 2005). When training at a specific RM load, it is recommended that a 2-10 % increase in load will be applied when the subject can perform the current workload for one to two repetitions over the desired number (Kraemer et al., 2002; Ratamess et al., 2009). In our study, ACL injured subjects performed 3 sets of 10 repetitions, even though we did not test the ACL injured subjects for 1 RM. Exercises and interventions aimed at increasing muscle strength following ACL injury should support the recommendations based on evidence for muscle strength training. Intensity, frequency, specificity, variation, and compliance are of significance in strength training. Each individual will respond differently and, therefore, should be supervised by a physical therapist during the rehabilitation program (Kraemer et al., 2005; Risberg et al., 2004; von, Henriksson, Holmstrom, & Roos, 2007).

2.3.2. Neuromuscular training

Neuromuscular control is responsible for the dynamic contributions that assist in maintaining joint stability via afferent information and efferent motor response (Griffin et al., 2000). It can be explained by muscular contractions in the proper patterns, sequence, and timing (Haywood, 1993). Neuromuscular training (NMT) aims to improve the nervous system`s ability to generate fast and optimal muscle contraction, better coordination and balance and to relearn movement patterns and skills (Risberg et al., 2001a). Furthermore, provide adequate neuromuscular control and avoidance of subluxations during functional activities (Chmielewski et al., 2005).

The importance of NMT to regain dynamic stability after ACL injury was introduced in the early 1990s (Markey, 1991; Beard, Dodd, Trundle, & Simpson, 1994). NMT exercises have been used for ACL injury prevention and to enhance dynamic stability in ACL injured subjects (Fitzgerald et al., 2000b; Williams et al., 2001; Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2005; Chmielewski et al., 2005; Hewett, Ford, & Myer, 2005a; Williams, Buchanan, Barrance, Axe, & Snyder-Mackler, 2005). Studies have shown that neuromuscular training programs induce changes in knee and hip biomechanics, as well as changes in muscle activation patterns, and clinical outcomes for ACL injured subjects(Fitzgerald et al., 2000b; Williams et al., 2001; Olsen et al., 2005; Chmielewski et al., 2005; Hewett et al., 2005a; Williams et al., 2005). Bruhn et al (Bruhn et al., 2006) have suggested that introducing NMT prior to muscular strength training enhances the effect of strength training, which may have an impact on the design of training protocols. Neuromuscular training consists of balancing on wobble boards or uneven surfaces. NMT includes balance exercises, dynamic knee stabilization exercises, plyometric exercises and perturbation training (Risberg, Mork, Jenssen, & Holm, 2001b).

Perturbation training involves manipulation of an unstable support surface while the subject maintains balance (Fitzgerald et al., 2000b). Perturbation training has been successful in improving kinematics, muscle recruitment strategies and dynamic knee stability among ACL injured subjects classified as potential copers (Chmielewski et al., 2005; Myer, Ford, Brent, & Hewett, 2006; Fitzgerald et al., 2000b). Neuromuscular training may therefore improve control of abnormal joint translation during functional activities by inducing compensatory alterations in muscle activity patterns and reducing the risk of continued episodes of giving way of the knee (Lephart, Kocher, Harner, & Fu, 1993; Risberg et al., 2001a).

Plyometric exercises involve pre-stretching the muscle and activating the stretch-shortening cycle to produce a subsequent stronger concentric contraction. When the elastic components of the muscle are being stretched, a certain amount of potential energy is stored. The potential energy is released during a concentric contraction (Schibye & Klausen, 2000). Hop training is often used to describe plyometric training and are recommended to improve jump ability (Kraemer et al., 2002; Ratamess et al., 2009). Plyometric exercises involves neuromuscular adaptation and deceleration of the body's centre of mass in an optimal and efficient manner (Hewett, Myer, Ford, & Slauterbeck, 2007). Drop jump (Figure 4), is one of the most popular plyometric exercise, involves jumping down from a height and, upon landing, performing a maximal jump (Hewett et al., 2007).

Figure 4 Example of plyometric exercise.

3. Concept of Measurements

The usefulness of measurement in clinical research depends on the extent to which clinicians can rely on data as accurate and meaningful indicators of what you want to measure. There are many criteria for scientific use of research methods.

3.1. Reliability

The first criterion is reliability, or the degree to which a measure is consistent and free from random error (Portney & Watkins, 2009). Reliability reflects reproducibility. Evaluation of reliability is accomplished by repeatedly measuring the same subjects and by repeated measurements by independent testers (Domholdt, 2000). There are three sources of measurement errors: 1: the instrument itself, 2: the tester and 3: the subjects being tested. A reliable instrument is an instrument that will perform consistently and predictably under the conditions which are set. Similarly, a reliable examiner will be able to measure repeated outcomes with consistent scores. Methods of establishing reliability vary depending on the instrument which has been used. Test-retest reliability illuminates the degree of stability in a measurement over time or between different testers. Test-retest reliability is divided into intra- or inter-test reliability. Inter-test reliability is the consistency between different testers doing the same measurements, and intra-test reliability is the consistency between the same tester during repeated measurements (Portney et al., 2009). The degree of relative reliability is often expressed by an intraclass correlation coefficient (ICC), ranging from 0.00 to 1.00. Values approaching 1.00, indicate a higher reliability (Portney et al., 2009).The degree of absolute reliability is often expressed as standard error of measurement (SEM).

3.2. Validity

The second criterion regarding scientific use of research methods is validity. Validity assures that a test is measuring what it is intended to measure (Portney et al., 2009). Validity is necessary for the ability to make a conclusion from data, and determining how the result of a test can be used. An essential part of validity is reliability, a test cannot be considered valid if it is not reliable. That means if the test is not consistent, if you cannot depend on successive trials to give the same results, then the test cannot be trusted.

Measurement validity is an expression which is used to determine whether the conclusions from the measurement are relevant, meaningful and useful (Domholdt, 2000). There are two main forms of validity, internal and external validity. Internal validity displays the quality of the study, including if the design was suitable, if the variables examined were relevant and if the variables were examined appropriately (Domholdt, 2000). External validity is used to determine if the results can be generalized (Domholdt, 2000). In this thesis the focus is on internal validity of the measurements.

Measurement validity can be categorized as logical, content, criterion, or construct validity (Portney et al., 2009). Logical validity indicates that an instrument appears to test what it is supposed to. Content validity indicates that an instrument adequately samples valid information based on the content of the test. It demands that the test is not influenced by factors that are irrelevant to the purpose of the measurement. Criterion validity is the degree to which scores on a test are related to some recognized gold standard or criterion. There are two main types of criterion validity: concurrent and predictive validity. Concurrent validity is the validity of the test related to a criterion, or a gold standard. Predictive validity establishes that the outcome of the test can predict a future criterion outcome. Construct validity is about the meaning of the variables in the study (Domholdt, 2000). It refers to which degree a test measures a hypothetical construct, and is usually established by relating the test to some established gold standard. In cases where there are no gold standards construct validity is usually considered most important (Portney et al., 2009). Knee function is an abstract human characteristic which we are unable to directly measure, and may be called a hypothetical construct (Domholdt, 2000).

In the following section I will present the scientific documentation for the use of the main outcome measurements included in this study and the reliability and validity of these measurements.

3.3. Motion Analysis

Motion analysis is a means of obtaining quantitative measurements of human motion and performance in 2 or 3 dimensions. It provides unique information not attainable through standard clinical examination. There are two important concepts of motion analysis, kinematics and kinetics. Kinematics describes the motion we see. It is defined as “the study of bodies in motion without regard to the cause of the motion”(Robertson et al., 2004). Kinetics is used to understand why the motions occur. It is defined as “the study of the forces and torques that cause motion of a body”(Robertson et al., 2004). Inverse dynamics is the specialized branch of mechanics that bridges the areas of kinematics and kinetics. It is a process by which forces and moments of forces are indirectly determined from the kinematics (position data) and inertial properties of moving bodies. Inverse dynamics determines the internal forces and moments that act across the joints in response to such external forces as ground reaction forces (GRF)(Robertson et al., 2004).

There are many factors that contribute to obtaining reliable and valid measurements using a motion analysis system. The determination of joint kinematics, for instance, is influenced by the instrumental errors, soft tissue artefacts, and anatomical landmark misplacement. Instrumental errors are of two types: systematic errors and random errors in the process of reconstructing marker trajectories (Chiari, Della, Leardini, & Cappozzo, 2005). The systematic errors depend on the size of the measurement volume and on the position that the marker assumes within (Chiari et al., 2005). Random errors may be due to electronic noise, marker flickering, i.e. merging of markers with each other or with phantom signals (Chiari et al., 2005). The motion analysis system used in the present study (Qualisys, ProReflex) can achieve very high accuracy under optimal conditions (an absolute accuracy better than 0.2mm in a $1m^3$ volume, Qualisys Inc., Gothenburg, Sweden). The magnitude of the instrumental errors is small when compared to other sources of errors.

Soft tissue artefacts have been proven to be a major source of errors in skin marker-based joint motion analysis (Leardini, Chiari, Della, & Cappozzo, 2005). A major concern is the assumption that the body segments behave as rigid bodies during movements. This assumption is obviously not valid because bones bend, blood flows, ligaments stretch, and muscles contract. Therefore, markers placed on the skin surface will move relative to the underlying bone during activities and with the deformation of soft tissue (Robertson et al., 2004). This represents an artefact which can disturb the accuracy of the measurement. Leardini et al. (Leardini et al., 2005) concluded that the pattern of the artefact is task dependent, it is reproducible within, but not among subjects, it introduced systematic as well as random errors, and the soft tissue artefacts associated with the thigh is greater than any other lower extremity segment. Skin markers can therefore be used to determine reliable joint flexion/extension movement (Leardini et al., 2005).

The gold standard of marker placement is an array of three noncollinear markers on a pin that is inserted directly into the bone (Robertson et al., 2004). This is a painful and invasive method, it may influence movement pattern and it likely would be difficult to obtain research subjects. Therefore, skin markers, which are a non-invasive and radiation-free technique, is the most widely used technique for human motion studies. There are different techniques for minimizing the soft tissue artefact (Leardini et al., 2005). We have used a cluster of three markers on a rigid plate on the segment, which have been reported to give less displacement than skin markers (Robertson et al., 2004).

The disadvantage of inverse dynamics in estimating segmental acceleration is that differentiation of position data amplifies errors (Bisseling & Hof, 2006). The ankle kinetics and shank acceleration are used to calculate knee kinetics, and the knee kinetics and thigh acceleration are used to calculate hip kinetics. At each step errors are introduced by the acceleration estimates as well as the estimates of joint centers and the center of mass of each of the segments.

Despite the various issue of reliability and validity, we find the motion analysis to be appropriate to our study. Errors might have arisen at different levels of the motion analysis capturing, but in the present study we tried to minimize the errors by using several compensation techniques. Procedures for a standardised set-up of the laboratory (number and placement of the cameras, the size of the measurement volume, the outcome of the calibration procedure, marker placement), calibration of the cameras to minimize systematic errors, filtering and smoothing techniques for the data to reduce random errors and the software takes care of the marker image processing and handling of missing markers are all means to minimize errors.

3.4. Muscle strength testing

The isokinetic muscle strength test is widely used in follow-up and outcome studies to evaluate the quadriceps and hamstrings muscle strength. It has shown to give reproducible and comparable assessment measurements of muscle performance (Dvir, 1995). The principle of isokinetic muscle strength testing is that the angular velocity is constant, and the load range of the performance is represented only under this condition (Dvir, 1995).

Isokinetic muscle strength can be tested during different velocities. Lower velocity (30-60°/s) are usually performed by few maximal repetitions (4-5 repetitions) while higher velocity (180°/s-300°/s) is performed with many maximal repetitions (20-30 repetitions) (Dvir, 1995). Low velocities have been reported to be more representative for muscle strength, while higher forces may be generated (Gaines & Talbot, 1999). The results are normally reported as total work or peak torque at high and low angular velocities. In this study we used total work as outcome measure in order to evaluate muscle performance, or the ability of the muscle to produce force throughout the movement cycle (Holm, 1996).

The use of modern isokinetic equipment results in minimal device errors; if the calibration routines are closely followed, the system is very reliable (Holm, 1996). Errors of the tester refer to, among others, the position and fixation of the test subject. Human variability refers to patient motivation, performing the test at the same time of the day, and giving all the test subjects the same type of encouragement. The reliability for isokinetic muscle testing for knee extension has previously been reported to be adequate (Impellizzeri, Bizzini, Rampinini, Cereda, & Maffiuletti, 2008). For isokinetic testing of the knee extensor muscle performance at 60 ˚ the intraclass correlation coefficients (ICC) have been shown to be above 90, both for healthy subjects and for subjects with an ACL injury (Impellizzeri et al., 2008; Sole, Hamren, Milosavljevic, Nicholson, & Sullivan, 2007). Absolute reliability (standard error of measurement) from 4% to 10% for peak torque and average work measurements (Impellizzeri et al., 2008; Sole et al., 2007).

Concerning the validity of isokinetic muscle strength measurements, three questions are of importance. First, does isokinetic testing measure muscle strength? Second, is there a relationship between muscle strength and functional outcome? And, third, to what degree are isokinetic measurements influenced by confounding factors, e.g pain (Holm, 1996). Dvir has stated (Dvir, 1995), in his book concerning validity, that isokinetic testing procedures are valid for measuring muscle performance in ACL injured subjects. Isokinetic testing at 60˚ is frequently referred to as a relevant and valid measurement for quadriceps muscle performance in ACL injured subjects (Dvir, 1995; Keays et al., 2003).

The disadvantage of using isokinetic muscle strength, as a definition of muscle strength, is that the movements performed in the isokinetic device rarely occurs in daily activities (Kannus 1994). Isokinetic muscle strength is performed in an open kinetic chain (OCK), while most daily life activities are performed in a close kinetic chain (CKC). The degree of validity and clinical relevance of isokinetic muscle test is therefore often discussed (Holm, 1996). However, one argument in favour of isokinetic muscle strength testing is that the test allows isolation of the muscle concern. Several studies have shown a positive correlation between isokinetic muscle strength and functional tests measures such as running, cutting and hopping (Risberg et al., 1994; Noyes, Barber SD, & Mangine, 1991; Risberg, Holm, & Ekeland, 1993). Nevertheless, it is important to keep in mind that landing after hop requires eccentric muscle work, while isokinetic testing is concentric muscle work.

4. Material and Methods

The present data collection was undertaken at the Sport Medicine Clinic Hjelp24NIMI, the biomechanics laboratory at the Norwegian University of Sports Sciences (NIH) and the biomechanics laboratory at the Rikshospitalet University Hospital during 2003-2007.

4.1. Subjects

Thirty-three consecutive ACL patients between the ages of 15 to 40 years who were referred to our institution (Hjelp24NIMI Ullevaal) in the period august 2003 until October 2005 were enrolled in the study. They were included if the ACL rupture was confirmed with magnetic resonance imaging, Lachmans test and a KT 1000 test of >3mm side to side difference with a manual maximum test, level 1 or 2 on the activity level score (Appendix I) (Hefti, Muller, Jakob, & Staubli, 1993), and no more than 6 months since the injury. Inclusion criteria for the screening examination, quadriceps muscle strength- and the motion analysis testing were: resolution of knee effusion, restoration of full knee range of motion, and the ability to hop on one leg without pain. Physical therapy treatment was implemented to resolve impairments prior to inclusion, if necessary. Subjects were excluded according to the following criteria: concomitant ligamentous injury, bilateral involvement, symptomatic meniscal or cartilage injury and/or fractures.

All subjects participated in a screening examination in order to be classified as a potential coper or a non-coper. Isokinetic muscle strength test and motion analysis during landing after hop were collected before and after intervention. In this master's thesis only those who were classified as non-copers by the screening examination were included (n=22). The uninjured leg was used as the control for isokinetic muscle strength and motion analysis during landing after hop.

4.1.1. Classification of Non-copers

Subjects were classified as either a potential coper or a non-coper using an established screening examination (Fitzgerald et al., 2000a). The screening examination consisted of hop testing, self-assessment questionnaires, and the number of giving way episodes during Activity of Daily Living since the injury (Fitzgerald et al., 2000b). The hop tests described by Noyes et al. (Noyes et al., 1991) consisted of four tasks: 1) a single legged hop for distance, 2) a cross-over hop for distance, in which the subjects crossed a 15 cm wide tape for three consecutive hops, 3) a straight triple hop for distance, and 4) a timed 6 meter hop test. Patients performed each hop test on the uninjured leg first, followed by the injured leg. Patients performed two practice trials, and then two trials that were averaged to gain a representative value for that leg. For the single, cross-over and triple hops, the involved side average scores were divided by the uninjured side average score and multiplied by 100. For the timed hop test, the uninjured side average scores was divided by the injured side average score and multiplied by 100. After completing the hop testing protocol patients completed two self-assessment questionnaires: The Knee Outcome Survey-Activities of Daily Living Scale (KOS-ADLS) and a global rating scale of knee function (earlier described in the theory chapter).

Subjects were determined to have knee instability and classified as non-copers if they had one or more of the following: (1) less than 80 % of the uninjured leg during the four hop test, (2) KOS-ADLS less than 80 %, (3) less than 60 % on the Global Rating of Knee Function and (4) more than one episode of giving-way following the initial injury until the time of the screening examination (Fitzgerald et al., 2000b).

4.2. Motion Analysis data collection

Kinematic data were collected using Qualisys pro-reflex motion analysis system (Qualisys Inc., Gothenburg, Sweden) with eight cameras at a sampling frequency of 240 Hz. The motion analysis system was synchronized with three force platforms sampling at a rate of 960 Hz (AMTI LG6-4-1, Watertown, MA 02472, USA). Attempts were made to position the cameras such that all markers could be seen optimally throughout the measurement volume. Before each data collection, the motion analysis system was calibrated according to the recommendations from the manufacturer. Six control points were used, four fixed and two movable located on a wand.

Anatomical markers defining joint centers and segment lengths were placed bilaterally over the following landmarks: iliac crest, greater trochanter, medial and lateral femoral condyle, medial and lateral malleolus, and the base of the first metatarsal. Clusters of three retro-reflective markers (15.5 mm diameter) attached to rigid thermoplastic shells were secured to the posterior aspect of the sacrum, and bilaterally to the posterior aspects of the thigh and shank to track three-dimensional movement. Foot segment motion was tracked by placing two markers on the heel of each shoe and a marker at the base of the fifth metatarsal. After a static calibration, the anatomical markers were removed and three successful single leg hop trials were collected.

Figure 5 Marker placements.

Subjects were instructed to perform a single leg hop for distance, from one force plate to another. They were told to stand on one leg, hop forward as far as possible and land on the same leg. There were no restrictions for the arms or trunk. They had to land controlled and balanced, without taking help steps. Two practice trials were conducted, before three successfully test trials with adequate landing on the force plate were performed. Failure to maintain the landing position resulted in a disqualified hop, and the measurement was repeated. Both limbs were tested, the uninjured side first. The single leg hops were performed wearing shoes and without a brace.

4.2.1. Motion Analysis Data Management

From the standing trial a kinematic model was generated by defining four skeletal segments (foot, shank, thigh, and pelvis) with 6 degrees of freedom. Each segment was defined by a four-marker system: a proximal medial marker, proximal lateral marker, distal medial marker, and a distal lateral marker. The midpoint between the medial and lateral marker was defined as the joint center. For areas where it was difficult to place the anatomical markers, for instance a proximal medial marker at the thigh; a virtual marker was calculated in Visual 3D software. The static subject calibration file was used to define a local coordinate system for each segment and to align them to the laboratory coordinate system. The raw coordinate data were filtered using an optimized cutoff frequency. Marker data were low pass filtered at 10 Hz with a fourth-order, zero lag, Butterworth filter. The ground-reaction forces were filtered using a fourth-order, Butterworth filter at a cutoff frequency of 45 Hz. Hip, knee, and ankle motions and moments were calculated using rigid body analysis with Euler angles, and joint moments were calculated using inverse dynamics (C-Motion, Inc. Rockville, MD, USA). Only sagittal plane kinematics and kinetics during landing were reported. Segmental weight was determined from the total body weight and Dempsters anthropometricdata. The joint resultant moment and forces calculated using this procedure were the estimated internal moments and forces and were based on the ground-reaction forces and segment inertial forces. Joint resultant forces were normalized to body weight, and joint resultant moments were normalized to body weight and height to allow comparison among subjects of different sizes.

All kinematic and kinetic data were post-processed using custom made software (Labview Version 6.0; Norway ML 2003). Furthermore, the kinematic and kinetic calculations were performed using the software Visual 3D (C- motion Inc, Crabbs Branch Way Rockville MD). Visual3D is a movement analysis program designed for the determination of 3D trajectories and of analog signals into kinematics and inverse dynamics reports.

The kinematic variables of interest included sagittal plane hip, knee and ankle angles at initial contact (IC) (when the vertical GRF exceeded 25N) and at peak knee flexion (PKF), and sagittal plane hip, knee and ankle range of motion (ROM) during landing. The magnitudes of ROM at the lower extremity joints were calculated from initial contact to peak knee flexion. Kinetic variables of interest included the internal sagittal hip, knee and ankle moment at IC and PKF, and percent contribution of individual joint moments to the total support moment of the leg. Hip, knee, and ankle extensor moments were summed to calculate the total support moment. The relative contribution of each joint to the support moment (percentage support moment) was calculated by dividing the individual joint extensor moment by the total support moment.

All data were analyzed for both legs and averaged across 3 single leg hop trials, and the average of the three trials was used for further analysis.

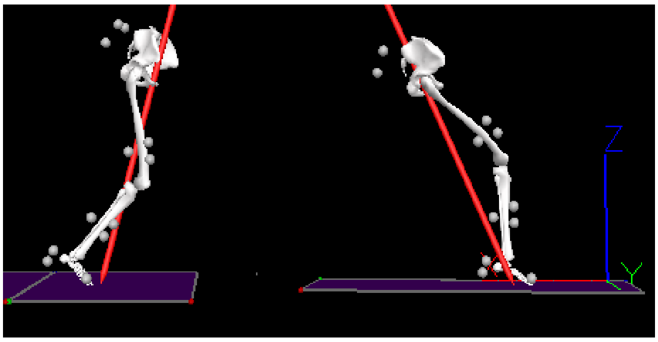

Figure 6 Visual 3D picture of push off and landing during a single leg hop.

4.3. Quadriceps muscle strength test

Quadriceps strength was measured using a Cybex 6000 Isokinetic Dynamometer (Cybex, Division of Lumex, Inc.,Ronkonkoma, NY,USA). Prior to the test the principles of isokinetic strength testing were explained to the subjects. Subjects were seated with hips and knees in 90° of flexion. Stabilizing straps were placed around the chest, pelvis, thigh and ankle. The shin pad was centred 2 cm above the medial malleolus. The axis of rotation of the dynamometer was aligned with the lateral epicondyle of the femur. The ROM for the knee was from 90° of flexion to full knee extension (0°). Before testing, patients were warmed up for 8 minutes on a stationary bicycle. There were given no verbal motivations during the test, only counting. Both limbs were tested, the uninjured side first. The test protocol consisted of 5 repetitions at an angular velocity of 60°/sec. The parameter used was total work (Joule) at 60°/sec. Relative total work was calculated as (injured side/uninjured side*100).

4.4. Calculation of joint torsional stiffness

A spring model was used to analyze the mechanics of landing after hop. The average joint stiffness of the knee was determined from the ratio of the change in net muscle moment to joint angular displacement in the sagittal plane between the minimum moment and the first peak moment at initial impact phase. In this phase, the joint behaved like a spring, the moment/angle graph was relatively linear. Three hop trials were collected for each subject. Mean characteristics across the three trials were used in subsequent analyses. Six subjects had one or two problematic trials where the joint stiffness could not be calculated, and they were excluded from the analysis. Mean of the two or simply one trial was used in these cases. One subject did not have data from the injured knee after rehabilitation due to trouble with the analysis.

4.5. Rehabilitation program

The rehabilitation program included both neuromuscular and strength training. Frequency was based on patient scheduling availability and the training response; twice a week was a minimum number of weekly exercise sessions. The subjects underwent 20 sessions of training, where each session lasted for about 1 hour and 15 minutes, over a time period of 10 weeks. Each component of the training focused on comprehensive biomechanical analysis by the instructor, with feedback being given to the subject both during and after training. The rehabilitation program focused on alignment, knee over toe position and softer landing, and consisted of balance training, dynamic knee stabilization exercises and strength training. The protocol for strength training was based on American College of Sports Medicine guidelines (Kraemer et al., 2002; Ratamess et al., 2009), and consisted of hamstring- quadriceps-, and calf exercises. Subjects were instructed to perform three series of 10 repetitions for all the strength exercises (Kraemer et al., 2005). The resistance was individualized, and increased when subjects were able to perform three extra repetitions at the third set of each exercise. Subjects were instructed to exert maximal effort for all three sets. Progression in balance and dynamic knee stabilization exercises were applied according to the responsible physical therapists' clinical judgment. Before attending the program, subjects warmed up on a stationary ergometer cycle, a step machine or a treadmill for at least 10 minutes.

The rehabilitation program included a bench press on one leg, squats on two legs, one leg exercises on balance mat, leg curls, leg raising exercises, step up and step down exercises, launch exercises, jumping exercises including jumping from height (drop jump), side jumps, and cross over jumps.

4.5.1. Compliance

Each subject filled out a weekly diary that documented their compliance with the rehabilitation program (Appendix II). The training diary included both number of visits for physical therapy intervention and hours spent exercising during the rehabilitation program, in addition to number of other exercise sessions and hours spent doing other exercise activities. Completion of at least 70% of the training sessions was required to be judged adherent to the rehabilitation program.

4.6. Ethical aspects

Before participation, all subjects signed a written informed consent form (Appendix III). The rights of the subjects are protected by the principles outlined in the Declaration of Helsinki, and the study was approved by the Regional Ethical Committee for eastern Norway (Appendix IV) and The Data Inspectorate (Appendix V). Prior to giving their written confirmation, the subjects were explained the criteria for participation, the purpose of the research and the procedure involved, potential risk, assurance and confidentiality, that their participation was voluntary, and that they could withdraw from the study at any time without giving any reason

4.7. Statistics

All statistical analyses were performed using NCSS 2004 (Number Crunches Statistical System, NCSS, Kaysville, Utah, USA). A skewness normality test was used to determine whether the data variables met the assumption of normality. Mean, standard deviation (SD) was used when a normal distribution was presumed, and median, first (1.) and third (3.) quartiles were calculated when a normal distribution was rejected. For comparison between the injured leg and the uninjured leg at baseline and after rehabilitation, and comparison between injured at baseline and after rehabilitation and uninjured at baseline and after rehabilitation, a paired Student's t-test was used for comparison when a normal distribution was presumed. Similarly, where the skewness normality test was rejected, Wilcoxons Signed-Rank Test for paired differences was used. Student's t-tests were also performed to compare the difference between injured and uninjured leg at baseline and after rehabilitation. The level of significance was established at $p<0.05$.

5. Results

At baseline 22 of the 33 subjects were classified as non-copers, 36% (n=8) females, and 64% (n=14) males. One subject did not complete the rehabilitation program and post test due to illness, and another three did not perform the post strength test (n=18). Four subjects had no knee joint laxity resulting from the KT-1000 manual maximum test (n=18). One subject did not have stiffness data after rehabilitation (n=20). Subjects' characteristics are shown in Table 1.

Table 1 Subject characteristics at baseline (n=22) Mean and standard deviation (SD)

	Mean (SD)
Age (years)	25 (5.2)
Days since injury (days)	64 (39.5)
KT-1000 manual maximum test (Millimetre difference between injured and uninjured knee)	8.03 (3.5)

5.1. Baseline

The non-copers had significantly less quadriceps muscle strength in the injured leg (744±201 Joule) compared to the uninjured leg (885±230 Joule) at baseline ($p< 0.001$). The injured leg had 85±13% quadriceps muscle strength of the uninjured leg.

Average kinematic and kinetic data from the three single leg hop trials are shown in Tables 3 and 4. Non-copers landed with significantly higher knee flexion and plantar flexion angles in the injured leg compared to the uninjured leg at IC (Table 2). There was a significantly reduced knee ROM in the injured leg compared to the uninjured leg ($p=0.01$) (Table 2).

The peak vGRF in the injured leg (34.7±6.6 N/kg) were significantly reduced compared to the uninjured side (38.6±6.7 N/kg), ($p=0.01$) at baseline. There was a significantly higher hip extension moment and knee flexion moment in the injured leg compared to the uninjured leg at IC (Table 3). There was a reduced plantar flexion moment in the injured leg compared to the uninjured leg at IC. At PKF there were reduced knee extension moment, and higher plantar flexion moment in the injured leg compared to the uninjured leg (Table 3). A typical knee moment curve is shown in Figure 7.

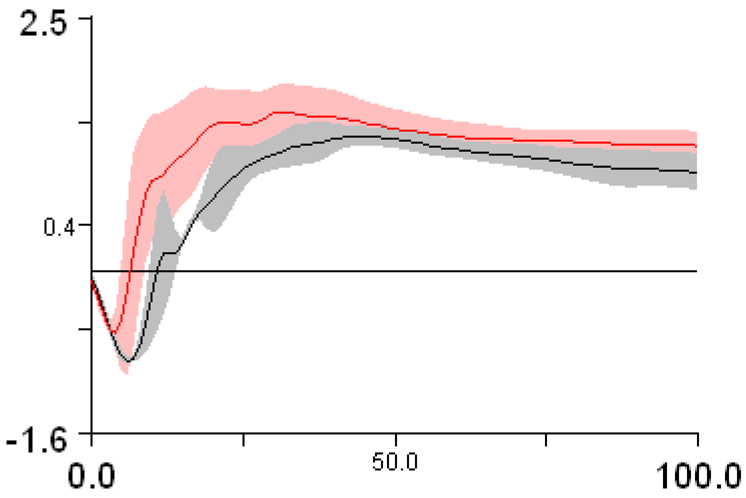

Figure 7 Typical knee moment curve during landing. The grey line is the injured knee and red line is the uninjured knee. The horizontal axe is the % of landing phase from initial contact to peak knee flexion, and the vertical axe is moments in Nm/kg*m. Positive value is extension and negative value is flexion moment.

The distribution of support moment showed that non-copers used a higher contribution from the hip and ankle and a lower contribution from the knee to the total support moment in the injured leg compared to the uninjured leg at baseline (Figure 9).

Non-copers had lower knee joint stiffness in the injured leg compared to the uninjured leg at baseline (p=0.04) (Table 2). The change in knee moments during initial impact phase (Δ moments) were significantly reduced (p=0.01) in the injured leg compared to the uninjured leg at baseline (Table 2). A typical knee joint stiffness curve is shown in Figure 8.

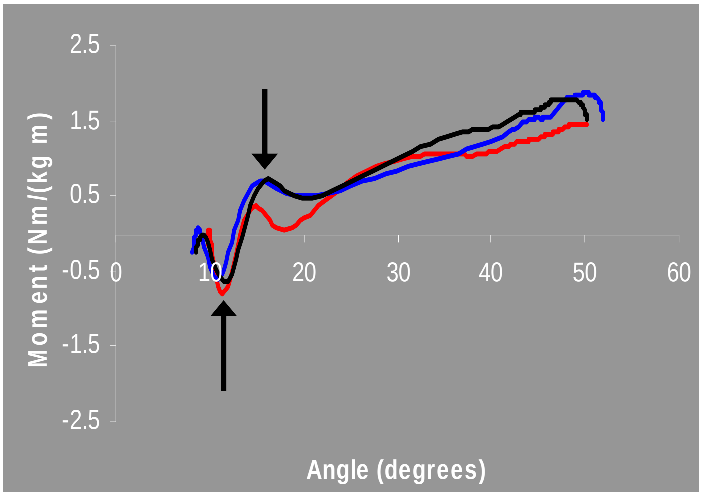

Figure 8 Typically knee joint stiffness curve, the Δ Angle /Δ Moment. The horizontal axe is angles in degrees and the vertical axe is moments in Nm/kg*m. The arrow shows where minimum and maximum moments, Δ Moment, and corresponding angles, Δ Angle, were picked at initial impact phase.

5.2. *After rehabilitation*

After rehabilitation the quadriceps muscle strength in the injured leg had significantly improved from 85% to 93±18% of the uninjured leg after rehabilitation ($p=0.02$) (Injured = 811 (SD=238) Joule, Uninjured = 884 (SD=218) Joule. But a difference in quadriceps muscle strength between the legs persisted ($p=0.03$). There was no difference in hip flexion or ankle angles between the legs at IC after rehabilitation. There was still higher knee flexion angle in the injured leg compared to the uninjured leg (Table 3). At PKF the hip flexion increased in both legs (Table 3).

After rehabilitation there was no significant difference in knee ROM of the injured leg compared to uninjured leg (Table 3). The hip ROM increased in the injured leg after rehabilitation (Table 3). For the joint loading, there was no longer a difference in moments at IC for all the three joints (Table 4). However, there were still a significantly higher hip and reduced knee extension moment in the injured leg compared to the uninjured leg during PKF (Table 4). There was no difference in ankle plantar flexion moment between the legs after rehabilitation (Table 4).

The distribution of support moment showed that the injured leg still used a significantly higher contribution from the hip and less contribution from the knee after rehabilitation to the total support momen, but there was no difference in the contribution from the ankle between the legs (Figure 9). The contribution from the ankle had significantly decreased ($p=0.03$).

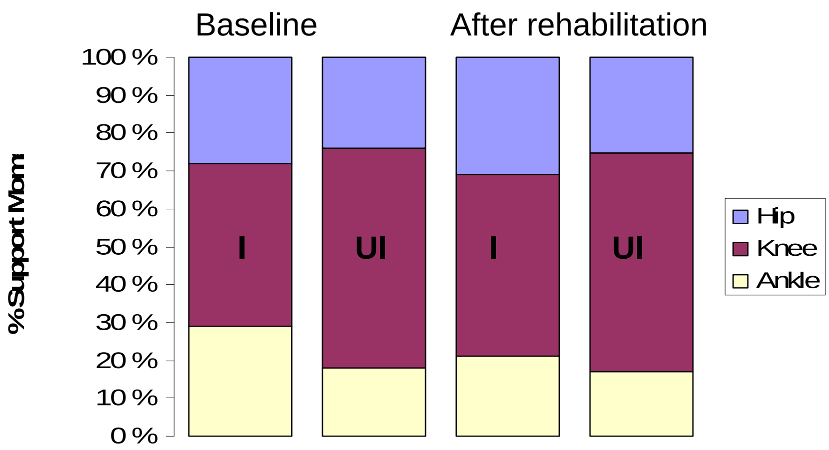

Figure 9 Distribution of support moments on the injured and uninjured side during landing after a single leg hop. The injured leg had significantly greater hip (p=0.02), less knee (p<0.001), and greater ankle (p=0.001) extensor moment at baseline. After rehabilitation, the injured leg had still significantly greater hip (p=0.001) and less knee (p=0.01) extensor moment, but there were no difference at the ankle between the legs (p=0.20) Support moments is total support moments as described by Winter et al (Winter, 1980). I=injured knee, UI=uninjured knee.

The peak vGRF did not change after rehabilitation, the injured leg (33.3±5.7 N/kg) had still lower peak vGRF than the uninjured leg (37.9±5.7 N/kg) (p=.

No statistical difference were observed in knee joint stiffness between the legs after rehabilitation (p= 0.53). The knee joint stiffness in the injured leg had significantly increased during landing after hop compared to baseline (Table 2). The change in knee angles (Δ angles) during initial impact phase significantly decreased in the injured leg compared to baseline (p= 0.02). The change in knee moments (Δ moments) during initial impact phase were still reduced in the injured knee compared to uninjured knee after rehabilitation (p=0.03) (Table 2).

Table 2; Knee joint stiffness during initial impact phase (ΔMoment/ΔAngle). Mean and standard deviation (SD).

	Baseline			**After rehabilitation**		
Joint stiffness	**Injured**	**Uninjured**	**P-value**	**Injured**	**Uninjured**	**P-value**
Knee	0.41 (0.24)	0.55 (0.21)	0.04	0.71 (0.87)*	0.88 (1.04)	0.53
Δ Angle	5.2 (2.9)	3.9 (1.7)	0.11	3.8 (2.3)*	3.5 (1.7)	0.79
Δ Moment	1.60 (0.55)	1.90 (0.73)	0.02	1.58 (0.40)	1.89 (0.75)	0.03

*p<0.05 between involved side and ** p<0.05 between uninvolved side from pre-to post test

Table 3 Joint angles (degree) at baseline and after rehabilitation, at initial contact (IC) and peak knee flexion (PKF) and range of motion (ROM) during landing after a single leg hop. Injured and uninjured leg n=22 at baseline and n=21 after rehabilitation. A positive value for hip and knee is flexion angle and for ankle dorsi flexion angle, a negative value is the opposite direction. Mean and standard deviation (SD).

	Baseline n=22			**After rehabilitation n=21**		
Angles	**Injured**	**Uninjured**	**P-value**	**Injured**	**Uninjured**	**P-value**
IC						
Hip flexion	36.2 (7.4)	34.2 (6.9)	0.05	37.6 (9.0)	36.0 (9.5)	0.25
Knee flexion	13.9 (5.3)	11.3 (3.7)	0.02	14.2 (4.5)	12.2 (4.1)	0.03
Ankle plantar /dorsi flexion	-1.9 (13.6)	3.5 (8.0)	0.04	-1.0 (9.7)	2.2 (6.9)	0.27
PKF						
Hip flexion	45.7 (10.7)	45.0 (11.1)	0.73	50.2 (12.1)*	48.8 (11.2)**	0.36
Knee flexion	53.4 (9.6)	54.0 (8.4)	0.80	53.8 (7.6)	55.0 (8.6)	0.51
Ankle dorsi flexion	1.7 (5.0)	2.4 (5.5)	0.25	2.9 (5.4)	2.6 (5.3)	0.81
ROM						
Hip	9.5 (5.9)	10.8 (6.4)	0.41	12.6 (7.2)*	12.9 (6.9)	0.88
Knee	39.5 (8.6)	42.6 (7.2)	0.01	39.6 (6.7)	42.8 (6.7)	0.07
Ankle	3.5 (12.5)	-1.09 (6.86)	0.20	3.9 (10.4)	0.4 (7.2)	0.19

*p<0.05 between injured side and ** p<0.05 between uninjured side baseline -to after rehabilitation

Table 4 Joint moments (Nm/kg*m) at baseline and after rehabilitation, at initial contact (IC) and peak knee flexion (PKF) during landing after a single leg hop. Injured and uninjured leg n = 22 at baseline, n= 21 after rehabilitation. A negative value means hip extension, knee flexion and ankle dorsi flexion, a positive value is the opposite direction. Normal distribution: Mean and standard deviation (SD). Rejected normality: Median (1. ; 3. quartile)#

	Baseline n=22			**After rehabilitation n=21**		
Moments	**Injured**	**Uninjured**	**P-value**	**Injured**	**Uninjured**	**P-value**
IC						
Hip extension	-0.22 (-0.85-0.81)#	-0.12 (-0.50-0.58)#	0.03	-0.15 (-0.31 ; 0.001)#	-0.13 (-0.29 ; -0.02)#	0.29
Knee flexion	-0.17 (-0.40-0.13)#	-0.10 (-0.29-0.16)#	0.01	-0.15 (-0.20 ; -0.02)#	-0.09 (-0.20 ; -0.03)#	0.26
Ankle plantar flexion	0.01 (0.04)	0.04 (0.05)	0.03	0.05 (0.05)*	0.05 (0.05)	0.97
PKF						
Hip extension	-0.56 (-0.85 ; -0.43)#	-0.46 (-0.60 ; -0.30)#	0.06	0.59 (-1.22-0.35)#	-0.52 (-1.06-0.11)#	0.05
Knee extension	0.88 (0.31)	1.15 (0.36)	<0.001	0.88 (0.37)	1.04 (0.37)	0.03
Ankle plantar flexion	0.58 (0.28)	0.36 (0.22)	<0.001	0.39 (0.24)*	0.31 (0.21)	0.21

*p<0.05 between injured side and ** p<0.05 between uninjured side baseline -to after rehabilitation

Median (1. ; 3. quartile)

Comparisons between baseline and after rehabilitation of the difference between the injured and uninjured knees were also done, but the only significant differences that were found were in quadriceps muscle strength (at baseline 136± 116 Joule, after rehabilitation 60 ± 128 Joule), ankle dorsiflexion moment (at baseline 0.21± 0.25 Nm/kg*m, after rehabilitation 0.08± 0.28)Nm/kg*m and displacement of knee Δ angles during initial impact phase (baseline -1.4± 3.5 degree, after rehabilitation -0.2± 2.6 degree.

6. Discussion

6.1. Baseline

Our first hypothesis was confirmed for all variables as the ACL injured subjects classified as non-copers demonstrated significantly reduced quadriceps muscle strength, knee ROM and knee extension moment (at PKF) in the injured leg compared to the uninjured leg at baseline. These results are consistent with those reported by other investigators during the stance phase of gait and hop (Rudolph et al., 1998; Rudolph et al., 2000; Hurd et al., 2007; Rudolph et al., 2001). The lower knee ROM and knee extensor moments in the injured leg compared to the uninjured leg may be a strategy to minimize the risk of anterior knee instability induced by the quadriceps muscle (Shelburne, Torry, & Pandy, 2005b). Long term consequences of such changes in knee biomechanics and in quadriceps muscle strength will be discussed later in this chapter.

The reduced knee motion during landing makes the non-copers injured knee less flexible during weight acceptance. With greater ROM in lower extremity joints the lower extremity musculature has more room to perform necessary work during the shock absorption process during landing, thereby reducing the peak loads across the lower extremity (Zhang et al., 2000). Reduced knee ROM has also shown to result in or be the result of muscular co-contraction (Chmielewski et al., 2001; Rudolph et al., 2001). This explanation has been supported by others who have reported higher muscle co-contractions during gait (Hurd et al., 2007; Chmielewski et al., 2005). However, these strategies during landing after hop have rarely been studied.

Knee angle at initial contact (IC) is believed to have a significant relationship to impact reduction capacity. Elvin et al (Elvin, Elvin, Arnoczky, & Torry, 2007) showed that during landing after hop the impact forces decreased with increased knee flexion. A small increase in knee flexion angle at near full knee extension had a greater effect on impact forces than the same angle at larger knee flexion angles (Elvin et al., 2007). We found reduced peak vGRF in the injured leg compared to the uninjured leg. The reduction in the peak vGRF may therefore be a consequence of the non-copers' movement strategy. The non-copers landed with greater knee flexion angle at IC and a more plantar flexion position at the ankle in the injured leg compared to the uninjured leg.

The amount of knee flexion has also been reported to be influenced by quadriceps strength at IC during walking for non-copers (Rudolph et al., 2001). Jump-landing tasks require the body to utilize various movement strategies in order to absorb the body's energy and forces during landing. Muscles are considered as a major source for reducing shock because of their ability to absorb kinetic energy during movements (Lafortune, Lake, & Hennig, 1996). Reduced quadriceps strength in the non-copers may influence the shock during landing after hop, as quadriceps is well known as the main absorber. Profound quadriceps weakness among subjects with poor function has been reported by numerous investigators (Wojtys & Huston, 1994; Eastlack et al., 1999; Williams et al., 2005; Rudolph et al., 1998; Rudolph et al., 2001). The strong influence of quadriceps strength on knee function in non-copers, suggests that muscle weakness is of clinical significance for these individuals. Previous studies suggest that non-copers have reduced peak knee flexion angles however, there were no statistical difference between the PKF angles in this study. The knee moment though, showed a significantly higher flexion moment at IC. The flexed knee adaptation revealed in the non-copers may increase their angle between the tibia and hamstrings and thus improve the hamstrings ability to pull backward on the tibia (Shelburne et al., 2005b) and thereby reduce the anterior drawer to protect the knee.

Our second hypothesis was also confirmed, the non-copers showed significantly lower knee extension moment and higher plantar flexion moments, with a tendency of higher hip extension moment in the injured leg compared with the uninjured leg. They also revealed significantly lower knee joint stiffness in the injured leg compared to the uninjured leg at baseline. The lower knee moment, in conjunction with higher hip moment and ankle plantar flexion moment during landing, may reflect a greater compensating strategy to knee stability from the hip and ankle extensors. The total support moment in this study indicated that the non-copers use a compensation strategy previously described as a hip-ankle strategy in the injured leg (Winter, 1980; Houck, Duncan, & De Haven, 2005). Winter et al (Winter, 1980) found that the summated extensor moments were consistent in injured and uninjured subjects during walking. When a deficit existed in one joint, he found that other lower extremity joints would compensate and increase their extension moment to maintain the summated extension moment throughout the lower extremity (Winter, 1980). Our results are consistent with this pattern, where lower contributions from the knee are compensated by higher hip and ankle contribution to the total support moment. This indicated that landing from a hop challenges knee stability for the non-copers. By using the aforementioned strategy, the control of the knee stability is transferred away from the unstable knee to the hip and ankle to protect the leg from collapsing. The same pattern has been observed during walking and jogging (Rudolph et al., 2000).

The reduced knee extension moment may be an adaptive change in quadriceps muscle force, to reduce anterior tibial translation (Shelburne et al., 2005b). The higher plantar flexion moment combined with the reduced knee extension moment suggests the importance of using the plantar flexors muscles to effectively absorb the shock during landing. By using the plantar flexiors muscle the quadriceps muscle contraction demands may be reduced, and thereby give lower GRF, and less anterior shear forces through compression (McNair et al., 1994). The tendency of a higher hip extension moment combined with significantly reduced knee extension moment may indicate that the non-copers lean their trunk more forward, thereby helping to decrease the strain placed upon the quadriceps during landing. It could also indicate an increase in or maintenance of hamstring muscle contraction demand (Shimokochi, Lee, Schultz, & Schmitz, 2009). Shelburne et al (Shelburne, Torry, & Pandy, 2005a) suggested that hamstrings facilitation is more effective than quadriceps avoidance in reducing the anterior pull. Thereby, the neuromuscular system may recruit the hip extensors and the ankle plantar flexors muscles to assist knee extension. However, we did not include electromyography (EMG) data in this study to verify these assumptions.

The reduced knee extension moment appears to be the hallmark of the ACL injured subject classified as a non-coper during both lower level activities (walking) and more strenuous activities (jogging and hopping). The moments we were measuring are net moments, which means the total moments of the leg. Therefore, individual tissue forces acting on the segment and their moments arm cannot be determined. The reduced knee extensor moment can more likely reflect a greater relative contribution of the knee flexors, including hamstrings and gastrocnemius muscles than less quadriceps force (Rudolph et al., 2001). These speculations cannot be confirmed without EMG. The lower knee extensor moment in non-copers may be explained by some combination of decreased knee ROM during landing and lower vGRF. The knee ”stiffening strategy” or altered biomechanics seen in non-copers may increase cartilage contact stress through inadequate load distribution, and dynamic knee joint stability (Chaudhari, Briant, Bevill, Koo, & Andriacchi, 2008). Perhaps their movement pattern shifts the loading location on the cartilage. Such a shift may cause new regions of the cartilage to become loaded, be exposed to altered levels of compression and tension, or become unloaded.

This could in the long run induce degenerative metabolic changes in regions of cartilage and development of premature knee OA if the tissue cannot adapt to the new loading pattern (Slemenda et al., 1998; Thorstensson, Petersson, Jacobsson, Boegard, & Roos, 2004; Chaudhari et al., 2008).

Although the significant differences in knee angles and knee moments between the injured and uninjured leg were small, it is consistent with an immature compensation strategy for ACL injury because people who can fully cope with the injury have been shown to have movement patterns that are no different from uninjured people in a variety of tasks (Rudolph et al., 1998; Rudolph et al., 2001). The question is, do the significant differences that we found reflect clinical relevance? To answer such a question, a cohort of ACL injured non-copers need to be followed prospectively for many years.

The injured leg also showed lower knee joint stiffness during landing after hop compared to the uninjured knee. It is possible that the non-copers have less ability to resist compression of the injured leg compared to the uninjured leg during landing after hop. Lower stiffness may therefore be important when impulsive loads are transmitted across the joints during the landing after hop. Research has shown that during the transition from non-weight bearing to weight bearing, there is an anterior translation of the tibia relative to the femur that normally is restrained by the ACL (Schultz et al., 2006). During landing from a jump, decrease in knee joint stiffness may be associated with alterations in early joint positioning that could produce corresponding effects on the sequence and magnitude of muscle contractions and thereby influence anterior translation. This lower knee joint stiffness might be a strategy that non-copers use to reserve them self against the fast and forceful activation of quadriceps during the landing, which may in turn lead to an anterior draw of the tibia. They choose a more gradual slope, leading to less knee joint stiffness in the injured knee, so that they may have more time to control the knee.

Our findings at baseline showed that the landing activity was challenging for the knee. The non-copers used a similar strategy during landing after hop compared to movement strategies reported earlier for walking and hopping (Hurd et al., 2007; Rudolph et al., 1998). They used a strategy which might give them a feeling of a safer landing. In summary, our non-copers showed less quadriceps muscle strength, both reduced knee ROM and knee extension moment at PKF in the injured leg compared to the uninjured leg during landing. Furthermore, the total support moment distribution in the injured leg was higher at the hip and ankle and lower at the knee compared to the uninjured leg. The knee joint stiffness in the injured leg was reduced compared to the uninjured leg during landing. The reduced knee joint stiffness may disclose lower demand on the injured leg. The unloading of the injured knee may be due to lower quadriceps muscle force, lower stiffness, and poor neuromuscular control with the goal of protecting the site of injury.

6.2. After Rehabilitation

Our first hypothesis regarding aim 2 was only partly confirmed; the non-copers increased their quadriceps strength in the injured leg from 85% to 93% of the uninjured leg after the 20 session rehabilitation program. There were no longer differences in knee ROM between the legs, but the knee extension moment (at PKF) still had not reached a value similar to the uninjured leg after the 20 session rehabilitation program. Previous attempts at improving the quadriceps strength among ACL injured non-copers have been unsuccessful (Williams et al., 2005), although other studies without dividing the ACL injured in subgroups have shown to increase strength after rehabilitation. The rehabilitation program in our study may have significantly increased the quadriceps muscle strength, and thereby improved the ability to decrease impact forces during landing after hop. However, due to the design of the study we do not know if these changes observed are an effect of time since injury or an effect of the actual intervention

No differences in the knee ROM between the legs after rehabilitation may indicate the adoption of a movement pattern that is more consistent with clinical findings of improved dynamic stability after rehabilitation (Fitzgerald et al., 2000b). The non-copers may utilize a greater knee ROM during landing after hop in order to allow the knee joint to continuously absorb the energy throughout a greater movement, and stabilize their knees in a manner similar to the uninjured leg. This is supported by others who have found increased knee ROM and decreased co-contraction during the stance phase of gait after a rehabilitation program (Chmielewski et al., 2005; Hartigan et al., 2009). This may be more relevant in more strenuous activities, such as landing after hop.

The hip ROM and the hip flexion angle at PKF had significantly increased in both the injured and uninjured side. There were no differences in ankle angles after rehabilitation, but the knee flexion angle was still significantly higher in the injured leg compared to the uninjured knee at IC. All the moments in the three lower extremity joints at IC were more normalized after rehabilitation at IC, making their movement pattern similar to the uninjured leg. However, they still landed with more flexion in the knee, probably due to minimizing quadriceps drawer.

Our second hypothesis was also partly confirmed. The total support moment distribution during landing after hop was higher at the knee (from 42% to 48%) and lower at the ankle, but still there were a significant reduced knee extension moment in the injured leg compared to the uninjured leg. There was no difference in the knee joint stiffness between the injured and the uninjured leg after rehabilitation.

The persistent lower knee extension moment may indicate that the landing after hop is too challenging for the injured leg in non-copers. Recent literature has shown that rehabilitation can balance the knee extensor moment during walking (Risberg et al., 2009). However, landing leg hop put greater demands on the knee. Therefore, to resist the greater forces acting on the lower extremity during landing, the reduced knee extension moment may be a more suitable strategy for the non-copers to deal with the forces in landing after hop. In this manner they reduce the anterior drawer influenced by the quadriceps muscle, which again may influence the dynamic stability during landing (Shelburne et al., 2005b). It could also be as Huck et al (Houck, De Haven, & Maloney, 2007) suggested that the non-copers fail to learn to modulate knee extensor loads due to neuromuscular control deficit.

There were no differences in knee joint stiffness between the legs after rehabilitation. The knee angle displacement had significantly decreased after rehabilitation, but the change in knee moments during initial impact phase (Δ moment) stayed the same, it was still significantly reduced in the injured leg. The results showed a steeper slope between the knee Δ moments, and Δ angles, which could mean that the non-copers have more resistance to the displacement during landing. Maybe they can tolerate the fast and forceful activation of quadriceps during landing better, thereby relying more on their injured knee. On the other hand, it could also mean that they may be able to activate the muscle fibres more adequate to achieve the same force and moment (Δ moments did not change), and therefore are able to land with less excursion during initial impact phase. This is only speculation, and has to be confirmed by EMG data. However, the quadriceps strength significantly increased and the ability to modulate muscle force is one key attribute of neuromuscular control (Williams, Barrance, Snyder-Mackler, Axe, & Buchanan, 2003).

Our findings after rehabilitation revealed that there were more symmetrical landing strategies after rehabilitation between in the injured leg compared to the uninjured leg, although there were still significant differences. Whether or not these changes have clinical meaning remains uncertain. In summary the non-copers had improved quadriceps strength after rehabilitation. They had normalized knee ROM, but still they landed with a reduced knee flexion angle and knee extension moment (at PKF) in the injured leg compared with the uninjured leg. They had greater contribution from the injured knee moment to the total support moment, and knee joint stiffness showed no difference between the injured knee and uninjured knee after rehabilitation.

6.3. Methodological considerations

6.3.1. Study design

We recognize the inherent limitation of the present study. First and foremost, the study was a nonrandomized prospective study design. Because of the lack of a control group, we do not know if the biomechanical changes observed were an effect of time since injury or an effect of the actual intervention (the rehabilitation program). Including a control group with no intervention would be unethical because of our knowledge of the probable outcome after rehabilitation. On the other side, a different training regime could have been performed but this study did not have as its aim the investigation of the potential effects of different rehabilitation programs.

6.3.2. Methods

We have only examined the sagittal-plane motion at the hip, knee and ankle. Movement pattern alterations may have occurred in the transverse or frontal planes of motion at these joints that could provide further insight into dynamic stability in the lower extremity. Although, the literature have shown greater errors in transverse or frontal planes of motions compared to the sagittale plane (Della, Leardini, Chiari, & Cappozzo, 2005). Another limitation is that we did not obtain EMG data to assess muscle activation. Discussion of muscle contraction and co-contraction demands were based only on net joint internal moments data and basic biomechanical concepts and other studies (Chmielewski et al., 2005; Rudolph et al., 1998; Rudolph et al., 2001; Williams, Snyder-Mackler, Barrance, & Buchanan, 2005).

Stiffness is a parameter that can be calculated in a number of ways, and different methods for calculating stiffness will likely produce different results. This makes it difficult to make comparisons with other studies, and caution should therefore be taken in the interpretation of the data. Latash and Zatsiorsky (Latash et al., 1993) suggest that an accurate model must account for all the components that contribute to stiffness (tendons, ligaments, muscles, cartilage and bone). A model that accounts for all the components that influence motion is very complicated, and the mathematical model is not yet been developed. Thus most researchers return to the simpler mass-spring model (Farley et al., 1998b).

This is the first study to investigate knee joint stiffness in non-copers during landing before and after a rehabilitation program. Therefore, the interpretation is made more difficult. It is thought that stiffness has a major influence on performance (Butler et al., 2003; Arampatzis et al., 1999; Granata et al., 2002; Arampatzis, Schade, Walsh, & Bruggemann, 2001b; Arampatzis et al., 2001a), but the optimal amount of stiffness required for movement such as single leg hop remains controversial. A limitation of this study is that we only included knee joint stiffness. The total stiffness of the leg is dependent on every joint in the chain. The relative joint contributions to stiffness during an activity appear to be task-related (Farley et al., 1998b; Arampatzis et al., 1999; Arampatzis et al., 2001b) and should therefore be calculated. The hip and ankle joint in our study did not show spring-like behaviour during landing after hop, no linear response which made it difficult to calculate stiffness in these joints. The human body doesn’t always move like a simple mass and spring model, the movement is more complex. Another limitation in calculating the knee joint stiffness is the defining of the period during landing the relative linear relationship between angle and moments occured. We have only called it initial impact phase due to difficulties in deciding the point in time it occurred. We do not know if this linearity happens at the same time in different subjects, and this may also question the method used. We tried to calculate the knee joint stiffness from IC to PKF, but the results made no sense. The single leg hop task in this study may not be a good choice for using joint stiffness computations as the challenge for the body is not to restore energy elastically like a spring, but rather to absorb energy like a damper. So it does not come as a surprise that the stiffness results are not conclusive.

6.3.3. Statistics

The statistical methods used in this study were dependent upon the sample size included to reach statistical significance. A total of 22 subjects were used for statistical comparison, an adequate sample size compared with other biomechanical studies. However, no statistical power analyzes were carried out before or after intervention. This could have provided insight as to how many participants should have been included in our study to elucidate further dynamic knee stability differences. Due to the variability in most of the variables it would have been reasonable to include a greater number of subjects. Although hopping differs from the cyclic walking pattern. Each individual has different motor organization patterns to adjust to various tasks, which may indicate that strategies might be related to individual landing styles. Each individual seems to have a very specific landing pattern that influenced their lower extremity motion strategies based on preference and tasks demands. It could be closely related to sports background, training and confidence in the leg.

We only included non-copers due to their more stereotype movement patterns, and more prominent deficits. However, our results revealed that the non-copers had very different individual movement patterns during landing after hop, a large variance was seen. This resulted in a more heterogenic group than was earlier described. Perhaps some of the non-copers turned into potential copers after the rehabilitation program. Even so, subjects often have their own unique landing

A possible confounding factor is that we compare the injured with the uninjured leg, and differences present before the ACL injury were not taken in account. But, van der Hars et al (van der Harst, Gokeler, & Hof, 2007) found no important differences between the dominant leg and contralateral leg in healthy subjects, and Petschning (98) found that regardless of the leg injured, the uninjured leg could be used as an outcome standard from rehabilitation. As a consequence, the uninjured leg of ACL injured subjects may be used as a reference (van der Harst et al., 2007). However, our study showed that the uninjured leg also was influenced by the intervention, even though not significantly. Recent studies have suggested that neuromuscular dysfunction and quadriceps strength loss after ACL injury also affects the uninjured side (Konishi, Aihara, Sakai, Ogawa, & Fukubayashi, 2007; Ingersoll et al., 2008; Konishi et al., 2007). This makes it difficult to show any significant change in the difference between injured and uninjured leg. Even though comparisons of the magnitude of the difference and a within comparison are better statistical tests to determine the influence of an intervention, the baseline and after rehabilitation comparisons may tell us more clinically. There were only three variables which showed a significant difference from baseline to after rehabilitation (within comparison); quadriceps strength, ankle plantar flexion moment and Δ knee angles at impact.

7. Clinical implications

In the current study, non-copers showed kinematic and kinetic differences in landing characteristics after hop in the injured leg compared to the uninjured leg. The non-copers in this study still have impairments after the rehabilitation program that could affect the load on the cartilage during landing after hop, and thereby a possible degeneration of the knee joint over time. The normalized knee joint stiffness in the injured leg after rehabilitation may be a promising first indication that rehabilitation may result in a more mature and refined stabilization strategy in the injured leg in non-copers. Investigations that include EMG are warranted to determine the neuromuscular changes. Before conclusions can be made about the influence of joint stiffness to dynamic stability in landing after hop and the effect of rehabilitation more studies are necessary. To our best knowledge, this is the first study that has attempted to quantify knee joint stiffness during landing after hop in ACL injured subjects classified as non-copers.

A future question would be: are the altered movement patterns necessary compensating strategies to cope with an ACL injury for non-copers, or should the ACL injured subject aim at normalizing these movement strategies?

8. Conclusion

Aim 1

The non-coper had significantly less quadriceps muscle strength, both reduced knee ROM and knee extension moment in the injured leg compared to the uninjured leg during landing after hop. The support moment was distributed higher at the hip and ankle, and lower at the knee. They had less knee joint stiffness in the injured leg compared to the uninjured leg.

Aim 2

After a 20 sessions rehabilitation program, the quadriceps muscle strength had significantly increased, there were no difference in knee ROM in the injured leg compared with the uninjured leg, but they still landed with greater knee flexion and lower knee extension moment in the injured leg compared to the uninjured knee. There was no difference in the knee joint stiffness between the legs after rehabilitation.

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Figures

Figure 1 The structure of the knee (the right knee in a frontal view). The position of the knee is straight and the patella and the joint capsule are removed. The figure is derived and modified from Netter (Netter, 1995).

Figure 2 A spring-mass modell. The model consists of a linear spring representing the leg and a point mass equivalent to body mass (Farley et al., 1998a)

Figure 3 Squats, example of strength training.

Figure 4 Example of plyometric exercise.

Figure 5 Marker placements.

Figure 6 Visual 3D picture of push off and landing during a single leg hop.

Figure 7 Typical knee moment curve during landing. The grey line is the injured knee and red line is the uninjured knee. The horizontal axe is the % of landing phase from initial contact to peak knee flexion, and the vertical axe is moments in Nm/kg*m. Positive value is extension and negative value is flexion moment.

Figure 8 Typically knee joint stiffness curve, the Δ Angle /Δ Moment. The horizontal axe is angles in degrees and the vertical axe is moments in Nm/kg*m. The arrow shows where minimum and maximum moments, Δ Moment, and corresponding angles, Δ Angle, were picked at initial impact phase.

Figure 9 Distribution of support moments on the injured and uninjured side during landing after a single leg hop. The injured leg had significantly greater hip ($p=0.02$), less knee ($p<0.001$), and greater ankle ($p=0.001$) extensor moment at baseline. After rehabilitation, the injured leg had still significantly greater hip ($p=0.001$) and less knee ($p=0.01$) extensor moment, but there were no difference at the ankle between the legs ($p=0.20$) Support moments is total support moments as described by Winter et al (Winter, 1980). I=injured knee, UI=uninjured knee.

Tables

Table 1 Subject characteristics at baseline (n=22) Mean and standard deviation (SD)

Table 2 Knee joint stiffness during initial impact phase (ΔMoment/ΔAngle). Mean and standard deviation (SD).

Table 3 Joint angles (degree) at baseline and after rehabilitation, at initial contact (IC) and peak knee flexion (PKF) and range of motion (ROM) during landing after a single leg hop. Injured and uninjured leg n=22 at baseline and n=21 after rehabilitation. A positive value for hip and knee is flexion angle and for ankle dorsi flexion angle, a negative value is the opposite direction. Mean and standard deviation (SD).

Table 4 Joint moments (Nm/kg*m) at baseline and after rehabilitation, at initial contact (IC) and peak knee flexion (PKF) during landing after a single leg hop. Injured and uninjured leg n = 22 at baseline, n= 21 after rehabilitation. A negative value means hip extension, knee flexion and ankle dorsi flexion, a positive value is the opposite direction. Normal distribution: Mean and standard deviation (SD). Rejected normality: Median (1. ; 3. quartile)[#]

Appendix

I. The activity level scale

II. Log sheets

III. Consent form

IV. The Regional Ethical Committee for Eastern Norway

V. The Data Inspectorate

APPENDIX I

Navn:

AKTIVITETSSKALA

Hva slags idrett eller trening drev du på med (FØR SKADE)?

____

Hvor mange ganger pr uke (i gj.snitt) (FØR SKADE)?

____

Hva slags idrett eller trening drev du på med (FØR treningsstart)?

____

Hvor mange ganger pr uke (i gj.snitt) (FØR treningsstart)?

____

Nivå 1 (deltar 4-7 dager pr uke)

100 Hopp, brå vridninger og vendinger (håndball, fotball, basketball, volleyball, turn, squash)
95 Løp, vridning, vending (tennis, alpinski, ishockey, friidrett)
90 Ingen løping, hopping, eller vridning (sykling, svømming)

Nivå 2 (deltar 1-3 dager pr uke)

85 Hopp, brå vridninger og vendinger (håndball, fotball, basketball, volleyball, turn, squash)
80 Løp, vridning, vending (tennis, alpinski, ishockey, friidrett)
75 Ingen løping, hopping, eller vridning (sykling, svømming)

Nivå 3 (deltar 1-3 ganger i mnd)

65 Hopp, brå vridninger og vendinger (håndball, fotball, basketball, volleyball, turn, squash)
60 Løp, vridning, vending (tennis, alpinski, ishockey, friidrett)
55 Ingen løping, hopping, eller vridning (sykling, svømming)

Nivå 4 (ingen idrett)

40 Jeg utfører daglige gjøremål uten problem
20 Jeg har moderate problemer med daglige gjøremål
0 Jeg har store problemer med daglige gjøremål (krykker, full disability)

TRENINGSDAGBOK FOR UOPERERTE KORSBÅNDSKADDE

Navn:______________________ Pas.nr:_____ Uke: ___

ØVELSER	Reps/tid	Belastning	Man	Tir	Ons	Tor	Fre	Lør	Søn
Ergometersykkel									
Eliptisk stepmaskin									
Tredemølle									
Benpress eksentrisk (E)									
Leg curl									
Knebøy på røde puter									
Knebøy med belastning									
Ettbensknebøy på blå pute									
Oppsteg trapp på pute(P) og kneløft									
Utfall (P)									
Nedsteg trapp på pute (P)									
Tåhev eksentrisk (E)									
Balanse fram/tibake i trekkapp. (P)									
Balanse sideveis i trekkapparat (P)									
Fallhopp (2), (1), (P)									
Sidehopp (R)									
Hink over planke (3 hink og stopp)									
Nordic Hamstring									
Treningstid pr.dag									

Annen trening	Reps/tid	Belastning	Man	Tir	Ons	Tor	Fre	Lør	Søn
Treningstid pr.dag									
Total treningstid denne uken									

(E) = større eksentrisk belastning
(P) = airexpute
(R) = retningsforandring
(2) = tobenslanding
(1) = ettbenslanding

Gjennomsnittlig smerte ved aktivitet denne uken:

Smertefri ———————————— **Verst mulig smerte**

RETNINGSLINJER FOR UTFYLLING AV TRENINGSDAGBOK

- **Første test gjennomføres så snart pasientens kne ikke er hovent og hun/han kan hinke på ett ben**

- **Deretter skal skjema ”Treningsdagbok for uopererte korsbåndskadde” fylles ut for hver trening**

- **Pasienten skal trene 2-3 organiserte økter pr.uke**

- **Nå pasienten har trent 20 organiserte økter gjøres test nr.2.**

- **De 20 øktene skal gjennomføres på maksimum 10 uker**

- **Det skal inkluderes minimum 5 økter med pertubation trening**
    - Kan gjøres som egen økt eller i tillegg til ordinær trening

- **I tidlig fase gjøres alle øvelser med lett belastning og mange repitisjoner (3x20)**

- **I de øvelsene der det er mulig skal belastningen gradvis økes og repitisjonene ned mot 3x6 (spesielt knebøy) med tilnærmet maximal belastning**

- **Alle øvelser gjøres med så stort bevegelsesutslag som mulig (knebøy kun ned til horisontale lår)**

APPENDIX III

Informasjon til pasienter som har en isolert fremre korsbåndskade og som ønsker å vurdere deltagelse i prosjektet:

«Dynamisk stabilitet etter fremre korsbåndskade - evaluering av et treningsprogram»

Alle pasienter som pådrar seg fremre korsbåndskade får tilbud om et rehabiliteringsprogram, og det blir informert om at rehabiliteringen er helt sentral for å få tilbake en god kne funksjon. En del av de som pådrar seg en slik skade må gjennom en operasjon der man lager et nytt korsbånd ved hjelp av en sene fra det samme kneet. Men det er også en del personer med korsbåndskade som kan fungere veldig bra uten å operere. Uansett om dere vurderes for operativt inngrep eller ikke går de aller fleste igjennom et treningsprogram før evt. operasjoner.

Formålet med dette forskningsprosjektet er å undersøke om en screeningtest bestående av 4 kne funksjonstester (hinketester), to spørreskjemaer og en muskelstyrketest kan benyttes for å vurdere kne funksjonen etter en korsbåndskade, og for å evaluere effekten av et treningsprogram, enten som en konservativ behandling eller som treningsprogram før en eventuell operasjon.

Det vil bli gjennomført bevegelsesanalyse i et laboratorium på Norges idrettshøgskole. Denne testingen vil ta ca 2 timer. Ved elektromyografiske målinger fester man små elektroder på huden over muskulaturen for å registrere muskulaturens aktivitet. Dette er elektroder som *registrerer* hva som skjer i musklene. Bevegelses-analyse-testingen foregår ved at man fester små reflekskuler på huden og på skoene. Disse refleksene fanges opp av 8 kameraer som "fotograferer" hvordan benet beveger seg ved gange og ved hopping. Denne "fotograferingen" foregår ved at kameraene sender ut lys som fanger opp refleksene som sitter festet på bena. Ved hjelp av denne testingen vil man kunne finne ut mer om hvilke mekanismer som er med på å påvirke kne funksjonen og hva som bør gjøres for å endre på eventuelle funksjonsproblemer. Testingen av kne funksjonen vil gjentas etter endt treningsprogram. Før testene kan gjennomføres skal du ha et smertefritt kne, uten hevelse og med full bevegelighet.
Du vil få dekket reiseutgifter til testingen.

Bakgrunnen for å gjøre disse analysene er at tidligere undersøkelser har vist at pasienter som har et velfungerende kne på tross av sin korsbåndskade har et normalisert bevegelsesmønster, mens de som ikke har god kne funksjon beveger seg stivere ved gange og hinking.

For å kunne belyse spørsmålene knyttet til hvordan du opplever stabiliteten i kneet ditt etter skaden ønsker vi å intervjue deg i forbindelse med screeningstesten og underveis i rehabiliteringsprogrammet. Intervjuene vil dreie seg om erfaringer med å være skadet og med å delta i rehabiliteringsprogrammet.
Dataene fra intervjuene skal brukes i forskning om rehabilitering av personer med korsbåndskader. Intervjuene vil foregå i forbindelse med screening testen og en gang under treningsperioden på 8-10 uker (dvs 2 intervjuer).

Intervjuene vil ta ca 1 time og vil foregå på NIMI eller hvor du ønsker det. Intervjuene blir tatt opp på bånd. Båndene vil ikke være mulig å identifisere til deg personlig.

De som deltar i intervjuene vil ikke kunne identifiseres i rapporter og artikler.

Dersom du ønsker å være med i forskningsprosjektet må du være innstilt på å gjennomføre rehabilitering 2-3 ganger i uken hos fysioterapeut på Norsk Idrettsmedisinsk Institutt, NIMI Ullevål stadion eller Ekeberg, totalt 20 treningsøkter. Oppfølging hos fysioterapeut og treningen er uten ekstra kostnader. Du kan på et hvert tidspunkt trekke deg fra undersøkelsen.

Det er ingen kjent risiko ved å delta i disse testene eller rehabiliteringsprogrammet

Du har nøyaktig de samme rettighetene og forsikringsvilkårene som du ville hatt dersom du ikke deltok i denne undersøkelsen. Du har rett til å trekke deg fra undersøkelsen når som helst, og du har da rett på å kreve dataene slettet. Dersom feil oppdages har du rett på å få korrigert opplysningene.

Dataene som innhentes på kne funksjonen din vil lagres i manuelle arkiv med personidentifikasjon som låses inn, og du har til enhver tid full innsynsrett i dataene. Dataene avidentifiseres ved elektronisk lagring på PC for statistiske analyser. Elektronisk lagres dataene kun med nummer. Ingen av dataene sammenholdes med elektroniske registre. Lagringen av data vil foregå i henhold til personsopplysningsloven. Datatilsynet og Etisk komité har godkjent prosjektet.

Prosjektet planlegges avsluttet i 2006, og alle sensitive persondata vil bli slettet innen 2 år etter at studien er ferdig. Dersom nye studier basert på de innsamlede opplysninger blir aktuelle, ber vi om tillatelse til å henvende oss for nytt samtykke for slik bruk.

Dersom du har spørsmål underveis, kan du ringe forskningsleder/fysioterapeut May Arna Risberg tlf 41312776 eller fysioterapeut Håvard Moksnes, 2326 5640.

Samtykkeerklæring

Jeg har lest og blitt forklart informasjonen på medfølgende informasjonsskriv om prosjektet, og sier meg villig i å delta i undersøkelsen.
Jeg har forstått at deltakelsen er frivillig.

______ ______

Sted Dato

____________________________ ____________________________

Underskrift

Underskrift av foresatt
(dersom pasienten er under 18 år)

Consent form

The written consent states that the data will be deleted within two years after the end of the study (2008). Due to maternity leaf I had to extend my master period to spring 2009. The data inspectorate (personvernombud, Heidi Thorstensen) has improved the extended use of the data with a new passive consent form that will be sent to all subjects by our research group (NAR) by the en of 2009.

REGIONAL KOMITE FOR MEDISINSK FORSKNINGSETIKK

Øst-Norge (REK I)

Forskningsleder May Arna Risberg
Ortopedisk senter
Ullevål universitetssykehus

Deres ref.: Vår ref.: 544-03165 Dato: 24. november 2003

Dynamisk stabilitet etter fremre korsbåndskader – evaluering av et treningsprogram

Vi viser til Deres brev av 27.10.03 med svar på komiteens merknader til prosjektet.

Saken ble behandlet på nytt på komiteens møte 18.11.03

Komiteen tar brevet med vedlegg til orientering, og finner å kunne tilrå at prosjektet gjennomføres.

Dersom denne komiteen ønskes nevnt i pasientinformasjonen, bes "godkjent" endret til "vurdert". Komiteen er ikke forvaltningsorgan, og har således ikke godkjenningsmyndighet.

Med vennlig hilsen

Knut Engedal
Professor dr.med.
Leder

Ida Nyquist
Ida Nyquist
sekretær

Postboks 1130, Blindern, 0318 Oslo. Tlf.: 22 84 46 67. Fax: 22 84 46 61
Besøksadresse: Frederik Holsts Hus, Ullevål sykehus

Datatilsynet

KOPI

Ullevål universitetssykehus HF
Saksbeh: PoPA
Saknr/Jnr: 04/0267-1
Dato: 09 FEB. 2004
Arkivkode: 529
Merkn: Kopi Mai Anna Risberg

Ullevaal universitetssykehus HF
Kirkeveien 166

0407 OSLO

Deres ref	Vår ref (bes oppgitt ved svar)	Dato
M. A. Risberg - FUS	2003/1708-4 CBR/-	05.02.2004

KONSESJON TIL Å BEHANDLE HELSEOPPLYSNINGER

Datatilsynet viser til Deres søknad av 11. november 2003, om konsesjon til å behandle helseopplysninger.

Datatilsynet har vurdert søknaden og gir Dem med hjemmel i helseregisterloven § 5, jf. personopplysningsloven § 33, jf. § 34, konsesjon til å behandle helseopplysninger i forskningsprosjekt vedrørende evaluering av knefunksjonen hos pasienter med fremre korsbåndskade og effekten av et rehabiliteringsprogram.

Konsesjonen er gitt under forutsetning av at behandlingen foretas i henhold til søknaden, vedlagte merknader og de bestemmelser som følger av helseregisterloven med forskrifter.

Dersom det skjer endringer i behandlingen i forhold til de opplysninger som er gitt i søknaden, må dette fremmes i ny konsesjonssøknad.

I medhold av helseregisterloven § 5, jf. § 36, jf. personopplysningsloven § 35, fastsettes i tillegg følgende vilkår for behandlingen:

1. Den databehandlingsansvarlige skal hvert tredje år sende Datatilsynet bekreftelse på at behandlingen skjer i overensstemmelse med søknaden og helseregisterlovens regler.
2. Personidentifiserbare data skal slettes/anonymiseres straks oppbevaring ikke er nødvendig, og senest to år etter prosjektavslutning, 31.12.2008.

Med hilsen

Hanne P. Gulbrandsen
Hanne P. Gulbrandsen (e f)
rådgiver

Cecilie Rønnevik
Cecilie L. B. Rønnevik
rådgiver
(saksbehandler,
telefon 22 39 69 00)

Vedlegg: merknader
Kopi: UUS, Heidi Thorstensen