Aleksander Killingmo

Is postural sway in a one-leg static balance task a determinant for frontal-plane knee joint loads and excursions in a vertical drop-jump?

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Abstract

Background: Team handball and soccer world have a high incidence of anterior cruciate ligament (ACL) injuries. This injury has severe consequences such as long term rehabilitation, sport disability and osteoarthritis. The mechanism of injury is still debated, and some, but not all risk factors are known. Preventive training programs have shown to reduce the incidence of ACL injury; many of these programs include balance training, yet it is not known whether poor balance is a risk factor for ACL injury. A comprehensive knowledge of the risk to sustain an ACL injury would increase the ability to identify players at risk and to improve existing preventive training regimes. Frontal plane knee joint kinetics and kinematics of a vertical drop-jump (VDJ) have in a previous study been used to identify athletes at risk. **Purpose:** This study is part of an extensive cohort study that aims to identify risk factors of ACL injuries in female team handball and soccer players. The objective of this study is to explore the possible association between postural sway (PS), of one-leg static stance and knee joint kinetic and kinematics of a VDJ by 3D motion analysis? **Study design:** Cross sectional study. **Methods:** Out of 184 female team handball players and 187 female soccer players from the top divisions in Norway that attended the VDJ and one-leg static balance test, 151 team handball results and 156 soccer results were valid and analyzed. The one-leg static balance was recorded as mediolateral and anterioposterior velocity (m/sec), total distance (mm) and 95% ellipse area (mm^2) of PS, while the frontal plane knee joint kinetics and kinematics was measured as frontal plane projection angle (°), valgus angle (°) and valgus moment (Nm/kg). **Results:** There were only four statistically significant associations of PS measures and knee joint kinetics and kinematics out of 48 possible. The significant associations were between frontal plane projection angle and mediolateral sway, total distance and 95% ellipse area, and between valgus angle and 95% confidence area. The R^2 coefficients for the statistically significant associations were small. **Conclusion:** The results of this study did not reveal any consistent association between PS in one-leg static balance and FPPA, valgus angle and valgus moments in VDJ. Considering the limitations in this study, a possible association between postural sway and knee joint kinetic and kinematics should not be dismissed.

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Preface

This study was carried out at Norwegian School of Sport Sciences as part of a large scale ACL risk factor study carried out between 2007 and 2011, which was financed by Oslo Sport Trauma Research Center.

I would like to thank my supervisors Tron Krosshaug and Agnethe Nilstad for constructive feedback and helpful advice around the clock. They were always available and in good spirit, whenever I requested their help.

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This thesis could not have been written without the unconditional support from my family, and Rikke's amazing ability to patiently keep together, the things that I let fall apart. Now it is your turn.

Sognsvann, Oslo, 2011

Aleksander Killingmo

1. Introduction

Many people attend sports at some time in life. A Norwegian survey from The Norwegian Olympic and Paralympic Committee and Confederation of Sports show that more than 40 % of the Norwegian population is member of a sports organization, and the number is rising (www.idrett.no). For the last decade there has been a focus on health benefits of sports participation and physical activity in media, but higher sport participation gives more sports injuries. The high number of sports participation have great health benefits, but also a downside in sport injuries (Bahr, 2004).

Team handball and soccer are sports where there are reported frequent injury incidences (Andersen, Tenga, Engebretsen, & Bahr, 2004; Junge et al., 2006). Junge and colleagues (2006; 2009) found that team handball and soccer was the two sports with highest injury risk among the team sports in the 2004 and 2008 Olympic Games. When Langevoort and colleagues (2007) investigated the injury incidence of eight major team handball tournaments and found 108 injuris per 1000 player hours, while Andersen and colleagues (2004) registered 425 injury incidents in 174 Norwegian male national league soccer games in 2000 and Myklebust and colleagues (1997) found

Team handball and soccer are sports with many participants in Norway and globally. In Norway there are approximately 115,000 team handball players and 365,000 soccer players (www.handball.no, www.fotball.no). Internationally, the International Handball Federation organize about 800,000 team handball teams (www.ihf.info), and a survey from 2006 by Fédération Internationale de Football Association found that approximately 265million people play soccer (www.fifa.com).

One relatively common and severe injury in team handball and soccer is rupture of the anterior cruciate ligament (ACL). Silvers and Mandelbaum (2007) estimated in, a review, that the annual number of ACL ruptures in the USA is about 250,000. In Scandinavia, Granan and colleagues (2009) reported from the Scandinavian ACL reconstruction registries 2004-2007 that the incidence of ACL reconstructions was 32, 34 and 38 per 100,000 inhabitants for respectively Sweden, Norway and Denmark. However, looking at the inhabitants between 16 and 39 of age (defined as population at risk) the incidence figures increased to 71, 85 and 91 per 100 000 inhabitants respectively. The most recent figures from the Norwegian cruciate ligament injury registry, reported an average of 1744 registered reconstructions annually in the past five years in Norway (Nasjonalt register for leddproteser, 2010). ACL reconstruction figures represent approximately 50% of ACL injuries (Granan, Engebretsen, & Bahr, 2004). A great proportion of the ACL ruptures occur in team handball and soccer. The Norwegian cruciate ligament injury registry report that out of 1790 ACL reconstructions in 2009, 700 injuries were caused by soccer and 240 by team handball (Nasjonalt register for leddproteser, 2010) and Arendt and Dick (1995) investigated ACL rupture incidence rates and found that the average ACL injury rate was 0.31 in female soccer and 0.13 male soccer, measured in injuries per 1000 athlete exposures.

Arendt and Dick (1995) also found that there are more ACL injuries in female athletes compared to male. A German study found a high incidence of ACL injuries in female soccer players in the national league, with eleven injuries in 165 players in one season (Faude, Junge, Kindermann, & Dvorak, 2005). In Norway, Myklebust and colleagues (1997) followed male and female team handball players at the three highest levels of national series for two consecutive seasons and found 87 ACL injuries. The ACL injury rate was calculated to 1.8 % of the female players and one % of the male players, or an injury rate of 0.97 per 1000 playing hours. A year later, the same research group (Myklebust, Maehlum, Holm, & Bahr, 1998) showed a prevalence of 0.31 injuries per 1000 hours of training, and a sevenfold higher prevalence of ACL injuries in women compared to men in top league team handball. Both these studies also found that non-contact injuries account for 70-95% of all ACL injuries, and that contact injuries do not play an important role on the number of ACL injuries. Also the risk of suffering an ACL injury is much greater in competition than in training (Myklebust, Maehlum, Engebretsen, Strand, & Solheim, 1997; Myklebust et al., 1998). Most of the ACL injured athletes are between 15-19 years of age (Nasjonalt register for leddproteser, 2010). Additionally, players with a previously ruptured ACL are at significantly increased risk of sustaining a new ACL injury (Faude et al., 2005). It is well established that ACL rupture is a relatively common sports injury, that female athletes sustain ACL injury more often than males, and that the ACL injuries occur during competition and in non contact situations.

The consequences of an ACL injury are severe. Myklebust and Bahr (2005) highlighted the high rate of recurrent injuries and high prevalence of early onset osteoarthritis after an ACL injury in a literature review. Recurrent injuries and osteoarthritis are severe consequences and may force the athlete to stop participating in sports, or worse, to experience pain and loss of function later in life. The Norwegian ministry of Labor recognizes ACL injury as a significant lifelong disability and ascribes at least five % disability for the rest of the working career for individuals that have sustained an ACL injury (The Norwegian ministry of Labor, 2006) The cost of treatment and rehabilitation was estimated to be about $17,000 per injury in the US in 1999 (Hewett, Lindenfeld, Riccobene, & Noyes, 1999). This estimation does not count costs related to loss of scholarship funding, future disabilities and pain. There is no doubt that rupturing the ACL is a severe injury that has an even broader perspective than the individual athlete, the injury and its short-term rehabilitation.

To efficiently prevent an injury, it is compelling to identify the mechanisms of injury, but also consider he risk factors involved (Bahr & Krosshaug, 2005). There are several theories concerning what causes ACL injury, but there is no consensus on the main mechanism of injury (Bahr, 2009). One main theory of ACL injury mechanism is valgus collapse, in combination with shank rotation and little knee flexion can cause ACL injury (Hewett, Torg, & Boden, 2009; Olsen, Myklebust, Engebretsen, & Bahr, 2004).

In spite of the controversy concerning injury mechanisms, several research-groups have succeeded in reducing the incidence of ACL injuries by preventive training programs (Alentorn-Geli et al., 2009; Caraffa, Cerulli, Projetti, Aisa, & Rizzo, 1996; Grindstaff, Hammill, Tuzson, & Hertel, 2006; Hewett et al., 1999; Hewett, Myer, & Ford, 2005; Mandelbaum et al., 2005; Myklebust et al., 2003; Silvers & Mandelbaum, 2007).

However, there is still need to further develop prevention programs and find tests that helps the clinicians to identify athletes at risk. Hewett and colleagues (2005) showed that an athlete at risk of sustaining an ACL injury possibly could be identified when conducting a vertical drop-jump (VDJ), by looking at knee flexion angle, knee abduction angle and valgus moment in a 3D motion laboratory. Stensrud and colleagues (2010) found that the athletes displaying high valgus angles could be identified through VDJ and single leg squat in clinical tests.

The risk of sustaining an ACL injury is dependent on several risk factors, which more or less contribute to injury risk. Some risk factors are non-modifiable, such as gender, age and anatomical modalities. These factors are important to know of, because they identify groups of athletes where preventive measures should be emphasized. Some risk factors are modifiable such as playing surface, technique, rules and strength. These are important to identify in the individual athletes and alter.

One such risk factor could be balance. There have been shown associations between knee injuries and balance (Paterno et al., 2010; Zazulak, Hewett, Reeves, Goldberg, & Cholewicki, 2007), and balance training as part of preventive training programs has showed to injury incidence (Caraffa et al., 1996; Hewett et al., 1999; Myer, Ford, McLean, & Hewett, 2006; Myklebust et al., 2003). Balance is modifiable as it has shown to improve by training (Granacher, Gollhofer, & Kriemler, 2010; Holm et al., 2004). There are few studies investigating the role of balance as a determinant for ACL injury, and to our knowledge, no study has focused on the possible association between balance and knee joint kinetics and kinematics in a VDJ. Balance training has been shown to reduce valgus moments (Cochrane et al., 2010; Myer et al., 2006), and a Croatian study suggest that a balance index may be used as an effective risk predictor for ACL injury (Vrbanic, Ravlic-Gulan, Gulan, & Matovinovic, 2007).

The rational leading to balance as a possible risk factor is explained through valgus moments. The valgus moment may be simplified into three components. The magnitude of the Ground Reaction Force (GRF) translated through the foot, ankle and shank, the direction of this force (given by the direction of movement by the player and the foot placement) and the position of the knee in relation to that force. The latter component, the knee position could be influenced by the balance. It is reasonable that those with high balance skills, defined by little excursion of the center of pressure (COP)(Riemann and Lephart, 2002a), are better at keeping their knees aligned in relation to the ankle, hip and torso, and thereby possibly avoid extensive valgus loads and joint excursions, and in some situations avoid valgus collapse.

The purpose of this study was to investigate the possible relationship between balance during a one-leg static test and frontal plane knee joint kinetics and kinematics in a VDJ.

2. Objective

We want to explore whether there is an association between balance and frontal plane knee joint kinetics and kinematics that can be discovered by measuring balance as postural sway in one-leg static balance test, and frontal plane knee joint kinetic and kinematics by 3D motion analysis of a vertical drop-jump.

From this objective we formed the hypotheses:

H_1. There is a statistical association between postural sway in a one-leg static balance test, and frontal-plane projection angle, valgus angle and valgus moment in a vertical drop-jump in Norwegian elite female team handball and soccer players.

H_0: There is no statistical association between postural sway in a one-leg static balance test, and frontal-plane projection angle, valgus angle and valgus moment in a vertical drop-jump in Norwegian elite female team handball and soccer players.

3. Theoretical background

3.1 Anatomy of the knee

The knee is the largest joint in the body, usually regarded as a hinge joint, but the motion is more complex (Gray, 2005). It consists of two articular facets connecting tibia and femur, and a third articulation between the femur and patella (Gray, 2005). The joint works as a fulcrum for propulsive muscles, and allows for great mobility at the price of instability (Gray, 2005).

Axis and planes

The knee is considered to have six degrees of freedom: flexion/extension, internal-/external rotation, valgus/varus and anterio-posterior translation, medio-lateral translation and traction/compression (Kaltenborn, 2002). The primary movement of the knee is flexion, extension, internal and external rotation (Gray, 2005)

Range of motion

Normal range of motion is considered to be five to ten degrees of extension and 120°-140° of active flexion and 160° of passive flexion (Gray, 2005). The total passive range of rotation (internal and external) in the transverse plane is approx 25° dependent on the knee flexion angle (Markolf, Mensch, & Amstutz, 1976). The frontal-plane rotation has been shown to vary through flexion grades, allowing most movement at 0-30° (Markolf et al., 1995). At 20° Shultz and colleagues (2007) applied ten Nm torque to the knee and measured 5.5° abduction and 4.5° adduction.

Stabilizing structures

The human movement requires great movement and stability of this joint during gait, running and jumping. In flexion and extension of the knee the femur condyles shifts backward and forward because of the shape of the femur condyles, further there is low congruence of the articular faces of the femur and the tibia. These properties allow for great movement, but compromise the stability of the joint. Thereby, most of the static stability of the knee is provided by the menisci, the collateral ligaments and the cruciate (Gray, 2005)(figure 3.1).

The two menisci are fibrocartilaginous discs situated on the two articular surfaces of tibia, and are attached to the articular surface; the medial meniscus is also attached to the medial capsule and the collateral ligament (Gray, 2005). The tibial collateral ligament runs between the medial condyles of femur and tibia. It is flat and deltoid shaped, with the broad attachement on tibia, and is connected to the medial capsule and the tendon of m. Semimembranosus (Gray, 2005). The fibular collateral ligament runs from the lateral femur condyle to the head of fibula, is circular and has no attachments to the capsule or other tendons (Gray, 2005). The joint capsule is a strong fibrosus capsule enclosing the joints (Gray, 2005). The cruciate ligaments are attached to femur and tibia and are situated between the two femur condyles, in what is called the intercondylar notch (figure 3.7). The ACL is attached to the anterior aspect of the tibia, and the posterior cruciate ligament is attached to the posterior aspect of the tibia (Gray, 2005).

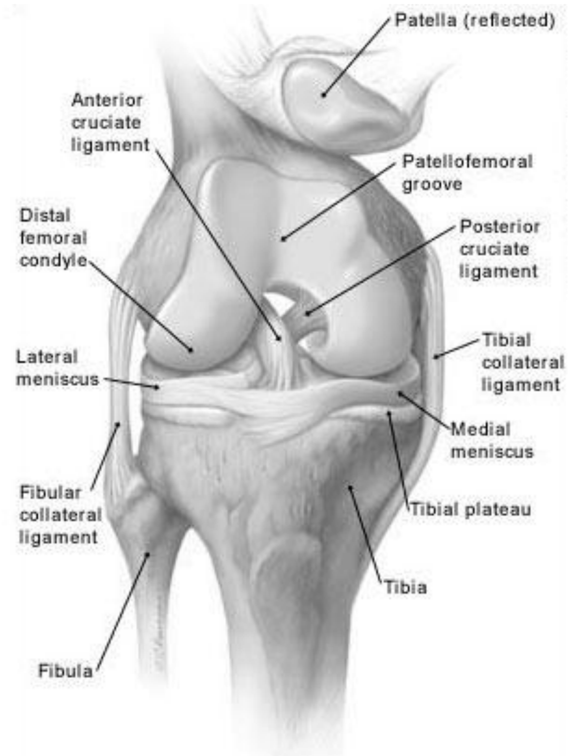

Figure 3.1: *Anatomy of the knee. Frontal view of a partly flexed knee with labels of major structures. The patella tendon has been cut and reflected (www.medchrome.com).*

The anterior cruciate ligament.

The ACL is attached on the posteromedial aspect of the lateral femoral curve, and anteriorly on the interchondylar area of tibia slightly anterior and lateral to the medial tibial eminence, partly attached to the anterior horn of the lateral meniscus (figure 3.1) (Gray, 2005). The ACL twist partly around its own axis and fans out towards the attachments (Gray, 2005). The function of the ACL is to provide stability to the knee joint. It limits the tibial anterior translation and rotation in relation to femur (Amis & Dawkins, 1991; Matsumoto et al., 2001). The length of the ACL ranges between 22 to 41mm and the width range from seven to 12mm (Amis & Dawkins, 1991; Duthon et al., 2006). The ligament consists of two or three bundles which are not isometric, the anteromedial bundle tightens with flexion and the posterolateral bundle tightens with extension (Amis & Dawkins, 1991; Duthon et al., 2006). This implies that the ACL has some stabilizing function in the knee joint during both extension and flexion. The complex organization of proteins, glycoproteins, elastic systems and glycomsamionglycans allow the ACL to withstand multiaxial strains (Duthon et al., 2006). Harner and colleagues (1994) measured the ACL cross sectional area five places from the femur origin towards the tibial insertion and showed that the ACL has an hourglass shape. The cross sectional area of the ACL decreases the first one fourth of the length from the femoral origin before it fans out towards the tibial attachment (Bernard, Hertel, Hornung, & Cierpinski, 1997; Harner, Paulos, Greenwald, Rosenberg, & Cooley, 1994). Woo and colleagues (1991) estimated ultimate load to the ACL to 2160N ± 157N and a linear stiffness of < 242 +/- 28 N/mm in 22 – 35 year olds. A direct strain of higher magnitude than the ultimate load will lead to ACL tear (figure 3.2) The mechanoreseptores in the ACL, innervated by the tibial nerve have proprioceptive function by providing afferent signals to induce postural changes, the “ACL reflex” (Duthon et al., 2006). An injury to the ligament may reduce this function and increase the time for the postural change to occur after stimulus (Duthon et al., 2006).

Figure 3.2: *A ruptured ACL (www.thefemaleatheltefirst.com).*

3.2 Theoretical framework for injury prevention

In 1992 Van Mechelen and colleagues (1992) recognized the need for a theoretical framework in research leading to injury prevention in sports. In their review they presented a four step sequence model (figure 3.3). The first step concerns identification of the magnitude by finding the incidence and severity of the magnitude of the injury. Secondly the etiology and the mechanism of injury need to be determined. Thereafter preventive measures have to be introduced and lastly the effect of the prevention should be assessed by repeating step one. The sequence should be repeated until satisfactory results are achieved.

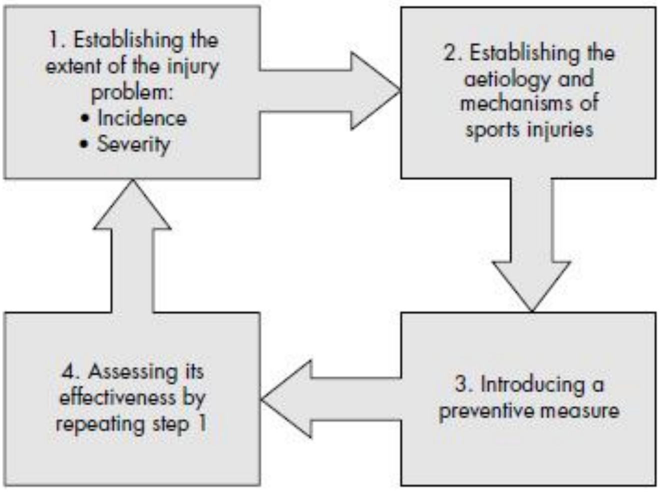

Figure 3.3: *The sequence of injury prevention (Van Mechelen et al., 1992).*

3.3 Injury mechanism research methods

Van Mechelen and colleagues (1992) identified the understanding of the etiology and injury mechanism (step two) as crucial in the sequence of injury prevention (figure 3.3). In 1994 Meeuwisse (1994) developed a multifactorial model for assessing the etiology and injury mechanisms. This was further refined by Bahr and Krosshaug (2005) (figure 3.4). They emphasized that a thorough understanding of internal risk factors, external risk factors and mechanism of injury is required to fully understand the etiology of a sport injury. In later manuscript and editorial Meeuwisse (2009; 2007) emphasized the cyclic factor of repeated participation as a source of adaption, or maladaption and the mechanism of no injury as additional factors in the multifactorial model.

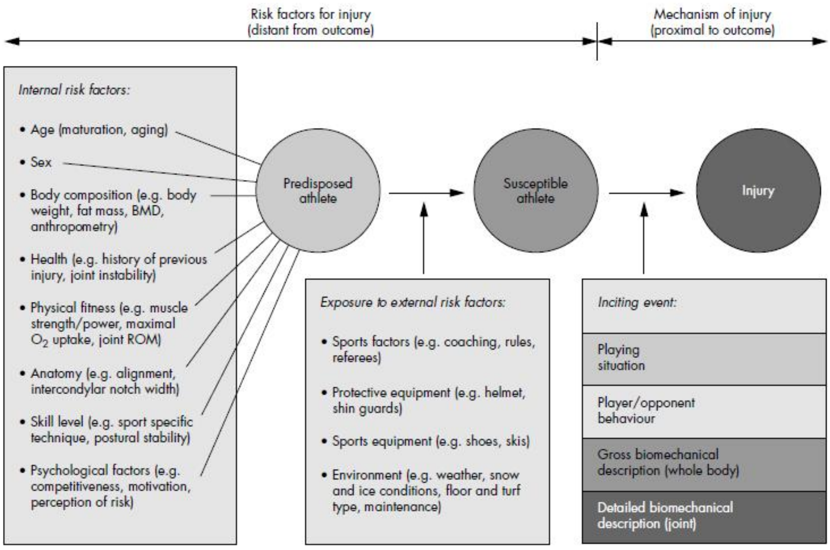

Figure 3.4: *A comprehensive model for injury causation (Bahr and Krosshaug, 2005).*

3.4 ACL injury mechanisms

Quatman and colleagues (2010) state that compression is the translation of highest magnitude that occurs in a movement causing ACL injury because of GRF, and muscle contractions. This is supported by the association of bone bruise following ACL injury. However, compression puts little strain on the ACL, and is not likely the cause of rupture (Quatman, Quatman-Yates, & Hewett, 2010).

The planar contributions of ACL injury mechanism are still debated. The common debate centers around in which plane the movement that causes the ACL to rupture occurs; whether it is in the transversal, frontal or saggital plane (figure 3.5). In the following a short resemblance of the planar theories are explained.

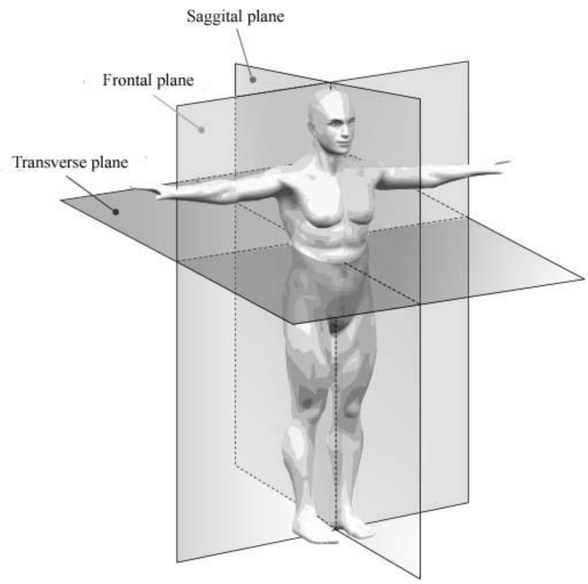

Figure 3.5: *The anatomical planes of the human body (www.thehumananatomydiagram.com).*

Transverse plane

The knee allows small rotations in the transverse plane (rotations and translations) as described earlier. Several studies have found that external, but most of all internal rotations increase the strain on the ACL (Markolf et al., 1995; Miyasaka, Matsumoto, Suda, Otani, & Toyama, 2002). In a study directly measuring ACL strain in 18 cadavers it was found that the strain was greatest when the knee was close to full extension (Wascher, Markolf, Shapiro, & Finerman, 1993). In another in vivo study, Myer and colleagues (2008) demonstrated that high compressive and torsional forces can rupture the ACL without substantial damage to other ligamentous structures. Quatman and colleagues (2010) state that only five % of the studies they reviewed support a sole transverse plane mechanism. However, it is plausible that ACL injuries sustained form torsion is underestimated, because the injury rapports are mainly based on video analysis. Estimations of knee rotations from video analysis have proven to contain great sources of error and should be interpreted with caution (Krosshaug et al., 2007).

Saggital plane

A recent review identified 32% of the studies reviewed to support saggital plane injury mechanism (Quatman et al., 2010). The saggital plane is the plane where the knee has largest range of motion and a more erect knee posture are theorized to increase risk of injury. Females display less knee flexion in jumping, landing and cutting tasks compared to males and several studies find low knee flexion angles during injury events

(Boden, Dean, Feagin, Jr., & Garrett, Jr., 2000; Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Krosshaug et al., 2007; McNair, Marshall, & Matheson, 1990; Olsen et al., 2004).

The ACL provides about 85% of the restraint against extensive tibial anterior translation (Butler, Noyes, & Grood, 1980; Markolf et al., 1976) and so this translation should be considered. Studies have found ACL deficient knees to have more tibial anterior translation than knees with intact ACL (Daniel, Stone, Sachs, & Malcom, 1985; Fukubayashi, Torzilli, Sherman, & Warren, 1982). Further, knees with intact ACL display greater range of anterio-posterior translation at low flexion grades (Fukubayashi et al., 1982). The quadriceps muscle creates a pull on the tibia through the patella ligament insertion. This has been suggested to create a significant anterior shear force on the ACL (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004). This anterior shear force produces direct ACL loading in combination with low flexion grades (Markolf et al., 1995; Sell et al., 2007). Because of this, it has been theorized that a powerful quadriceps pull, as required in activities like jumping, landing and cutting could cause the ACL to rupture (Li et al., 1999; Markolf et al., 1995; Yu, Lin, & Garrett, 2006). It has been shown that the force produced by quadriceps at low flexion angles produce sufficient strain to cause the ACL to rupture (DeMorat et al., 2004). These hypotheses are supported by MRI studies of knees after an ACL rupture where the tibial bone bruises are found more posterior than bone bruise on the femur (Mink & Deutsch, 1989; Speer, Spritzer, Bassett, Feagin, & Garrett, 1992). This bone bruise pattern may indicate that the tibia has been pulled anteriorly before the bone bruise is caused.

Frontal-plane

The passive frontal-plane knee joint movement are restrained by ligaments and knee joint articulation, which results in little range of motion (Gray, 2005). Excessive movement beyond this range of motion can have catastrophic result for the knee joint. Video analysis and interviews have displayed that ACL ruptures often are associated with excessive movement in the frontal-plane (Hewett et al., 2009; Kobayashi et al., 2010; Krosshaug et al., 2007; Olsen et al., 2004). Valgus moment has been suggested to lead to high ACL strain, especially close to full extension of the knee and with weight loading (Fleming et al., 1999; Krosshaug et al., 2007; Wascher et al., 1993). A prospective study by Hewett and colleagues (2005) that investigated the joint kinetics and kinematics during a VDJ in 205 female adolescent college athletes showed that the nine athletes that suffered ACL injuries the following season displayed 7.6° greater maximum knee abduction angle, 10.5° less knee flexion angle, and 2.5 times greater peak knee abduction moment than the uninjured group. Video analyses of ACL injury occurring during sport competition have some common elements in an injury situation. Commonly the knee is loaded and in ground contact, the athlete is performing some form for deceleration, while the knee is close to fully extended and the tibia is rotated either internally or externally, this followed by a valgus collapse (Boden et al., 2000; Krosshaug et al., 2007). This injury pattern theory is enhanced by Olsen and colleagues (2004) video analysis which found that valgus collapse is the most common injury mechanism for team handball players. Further, a video analysis of ten female team handball and basketball players Koga and colleagues (2010) identified immediate valgus motion and internal rotation after initial contact (IC) and a change to external rotation at approximately 40milliseconds. This suggests that frontal-plane valgus, coupled with internal rotation is a contributing factor of ACL rupture, and that the rupture occurs at approximately 40milliseconds.

McLean and colleagues (2004) motion analysis and mathematical model showed that abduction loads in a cutting movement can produce strain powerful enough to rupture the ACL while magnitude of saggital plane forces was too small to cause ACL injury MRI studies that found that bone bruising was most likely to occur on the lateral condyle of the femur, and the posterolateral portions of the tibial articular face (Mink & Deutsch, 1989; Speer et al., 1992; Viskontas et al., 2008). This may indicate compression of the lateral joint compartment following a loaded valgus movement.

Kobayashi and colleagues (2010) interviewed 1,700 people who suffered ACL rupture and found that the most common answer to cause of injury was dynamic valgus, and in that matter support this theory. Additionally, out of 198 studies reviewed by Quatman and colleagues (2010), ten % supported a solely frontal-plane injury mechanism, while 80% of the studies supported an injury mechanism where frontal-plane displacement was a central factor .

Investigations of his injury mechanism have revealed gender differences that possible can account for the gender disparity in injury occurrence. In a study of 81 high-school athletes performing a drop jump, female athletes displayed greater valgus motion and maximum valgus angle than male athletes (Ford, Myer, & Hewett, 2003). This coincides with the conclusion of Russell and colleagues (2006), that healthy women tend to land in more knee valgus compared to men form a single leg landing, and the study of Ford and colleagues (2006) that found female athletes to display greater frontal-plane excursions in hip, knee and ankle in a medial directed one-leg drop landing. A video analysis study of ACL injuries, Hewett and colleagues (Hewett et al., 2009) found that female athletes display greater knee abduction in an injury situation than their male counterparts.

Summary

Because of the intensity and movement patterns of team handball and soccer, players are prone to experience high loading conditions in any plane of the knee. 82% of 198 studies included in a literature review by Quatman and colleagues (2010) support the idea of a multi-planar injury mechanism. This supports the conclusion from Shimokochi and Shultz (2008) review that ACL injury usually is a result of multi-planar loading. Several injury mechanisms and combinations of these may be at play in different injury situations. In team handball and soccer, the valgus collapse is often cause the ACL injury (Krosshaug et al., 2007; Olsen et al., 2004; Quatman et al., 2010).

3.5 Known risk factors for ACL injury

Gender differences

Arendt and Dick (1995) found there was a higher incidence of ACL ruptures in females than in males performing the same sport, and most of the injuries were non-contact. Following this, in a retrospective study by Arendt and colleagues (1999) females were found to rupture their ACL significantly more often than males, but no physical or historical measurements were found to cause this difference. Simultaneously, Myklebust and colleagues (1998) examined gender differences of the incidence in ACL injuries across three seasons of Norwegian team handball males and females. They found a significantly higher risk of ACL injury for females (0.31 injuries per 1000 player hours) than males (0.06 31 injuries per 1000 player hours). Further, they described increased risk along with the menstrual cycle as a factor for gender difference. In a more recent study of 93 ACL injured soccer players, Brophy and colleagues (2010) found that females tend to injure their supporting leg, while males injure their kicking leg. This suggests that leg dominance is an etiological factor in the gender difference of ACL injuries. Several anatomical and physiological gender specific risk factors have been suggested (Ford, Myer, Toms, & Hewett, 2005; Griffin et al., 2006; Huston, Vibert, shton-Miller, & Wojtys, 2001; Lephart, Ferris, Riemann, Myers, & Fu, 2002). However there is no consensus of the cause of ACL injury gender difference.

3.5.1 Intrinsic factors

Quadriceps angle

The quadriceps angle is the angle between the imagined line from the anterior superior iliac spine trough the patella and the line between the center of patella and the center of the tibial tubercle (figure 3.6). In a study to determine the relationship between lower extremity alignment and quadriceps angle Nguyen and colleagues (2009) measured 219 subjects and found that greater femoral anteversion and tibiofemoral angle results in greater quadriceps angle. This angle is often reported as a possible risk factor for ACL injury. A recent review showed that there is conflicting evidence to support this theory and that further investigation is necessary (Posthumus, Collins, September, & Schwellnus, 2011). It is suggested that because the quadriceps angle represents a frontal-plane alignment measure it would be a poor predictor of ACL injuries occurring in a combination of planes (Nguyen, Boling, Levine, & Shultz, 2009).

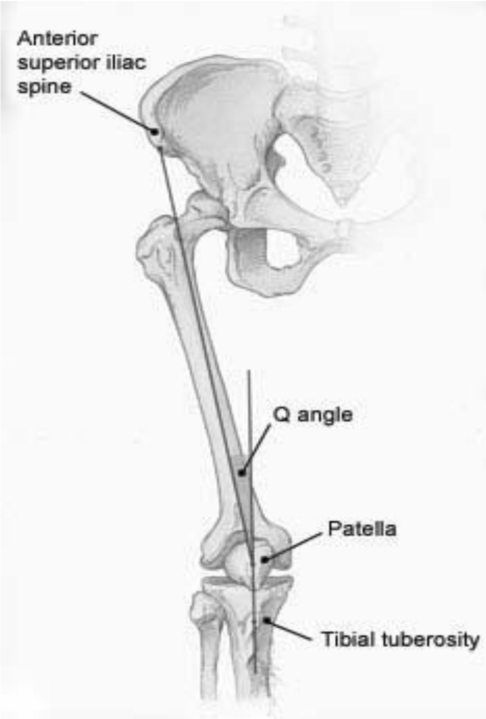

Figure 3.6: *Quadriceps angle (www.aafp.org).*

Interchondylar notch.

The gap between the two femoral condyles is called the interchondylar notch (figure 3.7). The width of the notch, and the notch width index (the ratio of the notch with and total condylar width) are both suggested to be risk factors for ACL injury (Hashemi et al., 2010; Renstrom et al., 2008; Uhorchak et al., 2003). Shelbourne and colleagues (1998) identified that females have a narrower interchondylar notch compared to males. They also concluded that a narrow interchondylar notch is a gender dependent risk factor for ACL injury based on the observations that the rate of re-ruptures after surgery did not display any gender difference. These studies suggest that a narrow notch has the potential to squeeze the ACL and cause rupture. Recently Everhart and colleagues (2010) identified a bony ridge on the anteriomedial part of the interchondylar notch with the potential to cause damage to the ACL.

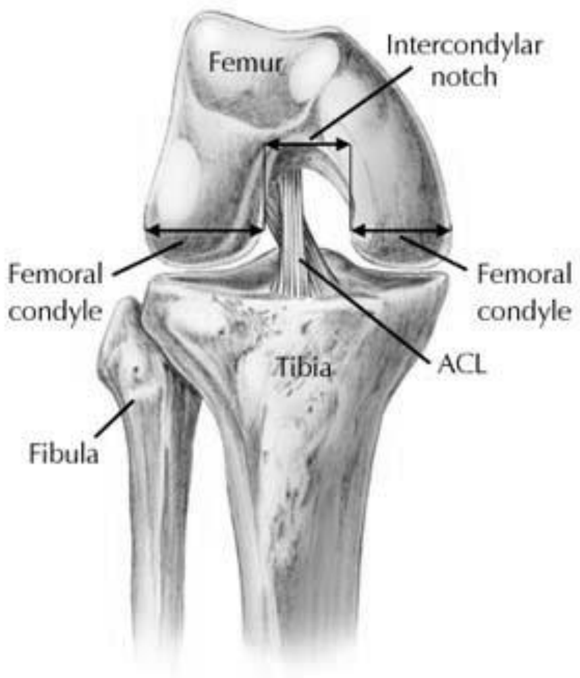

Figure 3.7: *Interchondylar notch (www.hughston.com).*

The tibial plateau geometry

The shape of the tibial plateau is discussed as a possible risk factor for ACL injury. Dejour and Bonnin (1994) found that there was a significant association with the tibial anterior translation and the angle between a horizontal line and the tibial plateau in the saggital plane. Following that study, several modalities of the tibial geometry have suggested as risk factors for ACL injury in addition to the degree of posterior inclination of the whole tibial plateau (tibial slope) (Simon, Everhart, Nagaraja, & Chaudhari, 2010; Todd, Lalliss, Garcia, DeBerardino, & Cameron, 2010), the medial tibial slope, lateral tibial slope (Hashemi et al., 2010; Simon et al., 2010) and the depth of the concavity of the medial plateau (Hashemi et al., 2010) all have been investigated as potentially contributing to the risk of ACL injury.

Shultz and colleagues (2010) state in their 2010 ACL research retreat that we know that compared to healthy controls, ACL injured have greater lateral tibial slope and shallower medial plateau. This conclusion is supported and extended by Posthumus and colleagues (2011) who concluded in their review that there is moderate level of evidence supporting that some factors of tibial plateau geometry increase the risk of ACL injury.

ACL geometry

The diameter of the ACL can influence the material strength of the ligament and essentially the risk of rupturing, assuming a strong correlation between ACL diameter and strength. There are few studies investigating the ACL diameter as a risk factor. There are studies suggesting that a narrow intercondylar notch indicates small ACL diameter, and that ACL diameter might be the risk factor in their study (Shelbourne, Davis, & Klootwyk, 1998; Uhorchak et al., 2003). The one study measuring the ACL diameter directly found significant association between diameter and injury risk only in females (Chaudhari, Zelman, Flanigan, Kaeding, & Nagaraja, 2009).

Navicular drop

Excessive pronation of the foot, measured by navicular drop has been investigated by several studies as a possible risk factor for ACL injury (Allen & Glasoe, 2000; Smith, Szczerba, Arnold, Perrin, & Martin, 1997; Woodford-Rogers, Cyphert, & Denegar, 1994). Most of the studies support the link between ACL injury risk and increased navicular drop, however the studies are designed as case-control and the number of participants are inadequate/ low (Allen & Glasoe, 2000; Woodford-Rogers et al., 1994). On the contrary, Smith and colleagues (1997) found that there was no significant relation between ACL injury and navicular drop . In conclusion, there is a possible link between navicular drop and ACL injury, but the knowledge in this area is not sufficient.

Joint laxity

There are several studies investigating general laxity, genu recurvatum, hamstring flexibility and knee anterior-posterior translation that have identified high levels of laxity and ACL injury risk (Boden et al., 2000; Myer, Ford, Paterno, Nick, & Hewett, 2008; Uhorchak et al., 2003; Van Mechelen, Hlobil, & Kemper, 1992; Woodford-Rogers et al., 1994). However, because of their cross-sectional design and different outcome measures no confident conclusion can be stated.

Hormonal factors

Yu and colleagues (1999) showed in a lab study that increased estrogen levels change the material properties of the ACL through reduced fibroblast proliferation and procollagene synthesis while progesterone has the opposite effect. This may imply that fluctuations of estrogen and progesterone throughout a menstrual cycle influence the risk of ACL injury. In a study with 14 college students Hertel and colleagues (2006) found no significant correlations between menstrual cycle and knee laxity, passive position sense, strength or postural control. In a three year prospective follow up, Arendt and colleagues (2002) displayed a significant relationship of increased ACL injuries in the follicular and luteal faces of the menstrual cycle. Myklebust and colleagues (1998) collected a reliable menstrual history of 17 ACL injured handball players, and found that there was a concentration of injury prior to or in the menstrual period. In a literature review by Hewett and colleagues (2007) that investigated the association between menstrual cycle and ACL injury, the conclusion was that female athletes are predisposed to ACL injury in the preovulatory phase. However, in a more recent review Posthumus and colleagues (2011) state that there have been a significant number of studies finding clustering of ACL injuries according to the menstrual cycle, but the time of the clustering is conflicting, and so there is no confident conclusion in how the menstrual cycle influences the injury risk.

Body weight

Uhorchak and colleagues (2003) found that female participants with body weight 1 standard deviation (SD) above mean had 3.2 times greater chance of sustaining an ACL injury, while the female participants with BMI 1 SD above mean had 3.5 greater chance of sustaining an ACL injury.

Muscle and joint stiffness

Hewett and colleagues (2005) proposed that athletes with dynamic valgus and high abduction loads are at increased risk of sustaining an ACL injury. This may partly be explained by difference in muscle and joint stiffness. Muscle stiffness is the ratio of force per length, while joint stiffness is the sum of resistance to a certain motion in the joint produced by all structures loaded in that motion (muscles, tendon, skin, subcutaneous tissue, fascia, ligaments joint capsule and cartilage) (Riemann & Lephart, 2002). Increased muscle stiffness can enhance joint stiffness, and this may be beneficial to functional joint stability. Stiffer muscles should be able to resist sudden joint displacements more effectively and thereby reduce destabilizing forces (Riemann & Lephart, 2002). Results that supported that explanation theory were displayed by Lloyd and Buchannan (Lloyd & Buchanan, 2001) who showed that hamstring and quadriceps co-contraction contribute most muscular support to reduce impact of varus and valgus moments. In their study the co-contraction supported 11-14% of the external moment in pure varus and valgus respectively. The co-activation will compress the joint, and apply structural stability to the knee because of the cavities on the tibial plateau. This could protect the knee from anterior translation, dynamic valgus and torsional loading (Hewett, Myer, & Ford, 2006; Posthumus et al., 2011). When weak hamstring muscles are unable to contribute sufficiently to the co-activation, the quadriceps pull has to be reduced to allow net flexion movement, and this would reduce the structural stability provided by the compression. This mechanism was proposed by Markolf and colleagues (1976) and displayed and refined by Hewett and colleagues (1996) in their study of plyometric training in volleyball players.

Hamstring/quadriceps ratio

Hamstring/quadriceps ratio may play a significant role as risk factor for ACL injury. There has been shown a threefold higher flexion moment in male athletes compared to female athletes implying that increased hamstring strength relative to quadriceps strength could reduce the risk of injury (Hewett, Stroupe, Nance, & Noyes, 1996). There are several studies suggest that the ACL is strained by quadriceps contraction within a range of 30° – 45° of knee flexion, implying that a high hamstring/quadriceps ratio could inflect a possible hazardous strain to the ACL (Beynnon et al., 1995; Lloyd, Buchanan, & Besier, 2005; McNair & Marshall, 1994). A prospective study of Mayer and colleagues (2009) found that females that suffer an ACL injury had stronger quadriceps but similar hamstring strength compared to uninjured controls. Sell and colleagues (2007) found that there is a gender difference in hamstring – quadriceps ratio where females have relative higher quadriceps activation which may generate more anterior tibial shear force. Additionally a study of Myer and colleagues (2005) found that EMG measures of quadriceps displayed higher lateral quadriceps readings relative to medial quadriceps in female than male athletes and by this possibly contribute to dynamic valgus.

Fatigue

Fatigue is often mentioned as a risk factor for ACL injury, however in a review from 2006 Hewett and colleagues (2006) concluded that few studies investigate this and no clear conclusion can be drawn on this issue. A latter study of ten male and ten female college athletes by McLean and colleagues (2007) found that fatigue may cause increased risk of ACL injury. This was supported by a study investigating one-leg drop landings that demonstrated that fatigue alters the movement pattern causing more extension and abduction moments (Kernozek, Torry, & Iwasaki, 2007).

3.5.2 Extrinsic factors

Risk factors that are extrinsic to the athlete have to some extent been investigated independently, but little is known about multivariate risk factors or sport specific conditions (rules, referee, coaching) (Renstrom et al., 2008)

Injuries occur during competition
Myklebust and colleagues (1997) showed that athletes are more often injured during competition compared to training. Later a study interviewing 1,700 Japanese ACL injured support this finding (Kobayashi et al., 2010). This indicates that there is a difference between training and competition, which is a significant factor for obtaining an ACL injury. The most obvious differences are level of intensity and performance; both factors may play a part in putting athletes at risk for ACL injury.

Playing surface
Friction between shoe and floor/turf may have the potential to increase the risk of rupturing ACL. Olsen and colleagues (2003) found more ACL injuries among female team handball players playing on artificial floor, than amongst those playing on wooden floor. However, this difference was not discovered amongst male players.
Australian rugby players play on similar turf as soccer players, and use similar footwear. A study from 1999 found that dry turf causes more ACL injuries that wet turf (Orchard, Seward, McGivern, & Hood, 1999), and a later study by the same research group suggests that the grass species may play a role in putting athletes at risk (Orchard, Chivers, Aldous, Bennell, & Seward, 2005). There is some evidence to support that the playing surface must be considered a risk factor for ACL injury.

3.6 Postural sway and neuromuscular control

Posture may be described as “the orientation of any body segment relative to the gravitational vector”, while balance “is a generic term describing the dynamics of body posture to prevent falling” (Winter, 1995). To maintain this equilibrium a human continuously makes small muscular contractions. This produces the center of mass

(COM) to sway within the limits of the base of support. The vertical projection of COM movement is COP (Shumway-Cook & Woollacott, 2001). This movement of COM is postural sway (PS). The magnitude of PS is affected by an athlete's height because of the inverted pendulum model, which states that the difference of COM and COP is proportional with the horizontal acceleration of COM (Winter, 1995). This is supported by a study of 25 male and 25 female young adults who found that both the sway path and the mean sway velocity were strongly dependent on height (Chiari, Rocchi, & Cappello, 2002).

In the dynamic process of performing a movement, there are two main mechanisms for controlling and adapting the movement: the feed forward control, which is an anticipatory action to a perceived change in PS that may challenge the balance, and the feedback control, which is a corrective response to sensory detection of a change in postural sway that may challenge the balance (Riemann & Lephart, 2002).

In postural control both control mechanisms are applied in combination. The action hierarchies in these mechanisms are similar: the sensory information (e.g. joint perturbation) provokes an afferent signal (e.g. mechanoreceptor), the afferent signal is transmitted by the peripheral nerves to the central nervous system where it is integrated at different levels, and an efferent stimuli is transmitted to the muscle which produces an adequate force (Riemann & Lephart, 2002). The control of even simple movements is a plastic process, constantly modified, based on feedback from three sensory sources (visual, vestibular and somatosensory). Somatosensory information is first and foremost based on proprioceptive information from joint and muscle receptors. Proprioception, that can be explained as afferent information from internal peripheral areas of the body, contributes to PS and joint stability (Riemann & Lephart, 2002).

These are primarily the same mechanisms providing neuromuscular control. Neuromuscular control is defined as "the unconscious activation of dynamic restraint occurring in preparation for and in response to joint motion and loading for the purpose of maintaining and restoring functional joint stability" by Riemann and colleagues (2002).

3.7 Injury prevention

It is well established that ACL injuries can be prevented to some extent by prevention training (Alentorn-Geli et al., 2009; Grindstaff et al., 2006; Hewett et al., 2005; Silvers & Mandelbaum, 2007). Several prospective studies have investigated different modalities of preventive exercises (Hewett et al., 1999; Mandelbaum et al., 2005; Myklebust et al., 2003). One study investigated neuromuscular training and found that female athletes who participated in a training program containing plyometric jump training and landing technique had 3.6 times decreased risk of sustaining and ACL injury compared to females not participating in the program (Hewett et al., 1999). A prospective study by Myklebust and colleagues (2003) following first to third level Norwegian female team handball players for three seasons concluded that it is possible to prevent ACL injuries by specific neuromuscular training. The study focused on knee alignment through balance tasks, cutting, jumping and landing.

In an attempt to indroduce preventive training into existing training habits and thereby improve compliance, Gilchrist and colleagues (2008) hypothesised that a simple on field warm-up program could reduce ACL injures, and replace comprehensive preventive programs which is time and effort consuming. They found, that their prevention warm-up program was effective with collegiate female soccer players. The "11+", a multifaceted warm-up program, which was constructed to replace regular warm-up and reduce risk of lower extremity injuries, showed a reduction of overall injuries by 32%, overuse injuries by 53% and severe injuries by 45% after eight months of training for approximately 1,900 female soccer players (Soligard et al., 2008). Myklebust and Steffen (Myklebust & Steffen, 2009) pointed out in an editorial that it is important that coaches and athletes in high-risk sports regard a preventive program as worthwhile to achieve compliance for such programs. It is therefore important to integrate the program in the normal training regime and to find the minimal effective "dosage".

Several reviews points out balance, plyometric strength, core stability as common factors in ACL injury prevention programs (Alentorn-Geli et al., 2009; Grindstaff et al., 2006; Hewett et al., 2005; Silvers & Mandelbaum, 2007). All though some common factors of successful prevention programs have been identified, the effect of each isolated factor, or the effect of combinations of different factors are unknown (Myklebust & Steffen, 2009). They further emphasize that to be able to develop a more specific and effective prevention program it is a prerequisite to achieve a more complete understanding of the injury mechanism. These conclusions are similar to those of Silvers and Mandelbaum (2007).

Balance and injury prevention

The practical application of neuromuscular training with balance components incorporated in preventive training has proven to reduce injury incidence (Caraffa et al., 1996; Hewett et al., 1999; Myklebust et al., 2003). Caraffa and colleagues (1996) investigated specifically balance training with 600 soccer players, where 300 players were asigned to 20 minutes a day ”wobble” board training , with increasing difficulty. The interventiongroup had significantly less ACL injuries than the control group. While Myer and colleagues (Myer et al., 2006) found that both plyometric- and balance training contributed to reduction of lower extremity valgus measures. A study of Holm and colleagues (2004) found that training designed to reduce the ACL injuries also improved the dynamic balance measured on a digital balance board with professional female team handball players, and this support the importance of balance in ACL injury prevention. Further, a more recent study found that balance training reduced peak valgus moment (Cochrane et al., 2010). The increased balance skills found by Holm and colleagues (2004) and the reduced valgus moment found by Cochrane and colleagues (2010) supports a Croatian study that found balance index that can be used to predict risk of ACL injury (Vrbanic et al., 2007). These studies may be part of the explanation to the results Zazulak and colleagues (2007) found, that impaired core proprioception could predict knee injuries in female athletes. On the contrary a Swedish study which included 221 female soccer players could show no effect of training on a balance board for eight months on injury rate (Soderman, Werner, Pietila, Engstrom, & Alfredson, 2000). In a 12 month follow-up of 56 athletes who had an ACL reconstruction, Paterno and colleagues (2010) found that postural stability deficit was one major predictor for a second ACL injury, while Herrington and colleagues (2009) found that ACL injured athletes reduced postural control in their uninjured leg compared to uninjured controls. This emphasizes the role of balance and postural control in injury risk. However, Søderman and colleagues (2001) investigated risk factors for leg injury in 146 female Swedish soccer players and found that the players that displayed low PS were more prone to traumatic injury than the players with high PS. They suggest three possible reasons for this result. First, a player with better postural stability should be able to avoid falling and by not falling ends up in more high risk situations. Their second theory is that players with better postural control are better soccer players, and therefore face more aggressive opponents. Their third theory is that the Kinesthetic Ability Trainer 2000 does not produce reliable results.

Except the two studies of Søderman, that originates from the same Swedish cohort study, the research points towards balance as an important part of preventive training programs. Further it seems that balance skill probably is a risk factor for ACL injury.

3.8 Osteoarthritis

There are several studies showing that there probably is a significantly greater risk for osteoarthritis (OA) (figure 3.8) for ACL injured patients regardless whether the treatment is conservative or surgical (Lohmander, Ostenberg, Englund, & Roos, 2004; Meuffels et al., 2009; Roos, Adalberth, Dahlberg, & Lohmander, 1995). Patients that suffers a knee injury, have a high prevalence of OA, was the conclusion of a review that found that patients who sustained an ACL injury or meniscus, had an average onset of OA at 40 years of age, compared to a control group with onset at 50 years of age (Roos et al., 1995). Lohmander and colleagues (2004) investigated female soccer players that sustained a ACL injury 12 years ago. They found that 82% had radiographic changes and 51% fulfilled the criterion for OA, however there was no control group indicating a normal prevalence for same-age soccer players. A later study by the same research group found that 50% of those diagnosed with ACL rupture or meniscus injury had OA, pain and functional impairment 10-20 years after the injury (Lohmander, Englund, Dahl, & Roos, 2007). A review by Øiestad and colleagues (2009) concluded that previously published reviews estimated the prevalence of OA to be too high, and that the prevalence of OA ten years after an isolated ACL injury was up to 13 %, while ACL injury combined with meniscus injury gave a prevalence of 21%- 48%. In a study of 210 subjects that underwent ACL reconstructive surgery, 71% showed radiographic OA, and 24% displayed moderate or severe OA (Oiestad, Holm, Engebretsen, & Risberg, 2010). However, the subjects with only low grades of OA showed associations between radiography and pain function or quality of life, while those with moderate or severe grades of OA experienced more pain and symptoms, was impaired from sport and recreational activities and had reduced quality of life. There was no difference whether the reconstruction was based on a hamstring graft or a patella tendon graft in matters of OA (Holm, Oiestad, Risberg, & Aune, 2010).

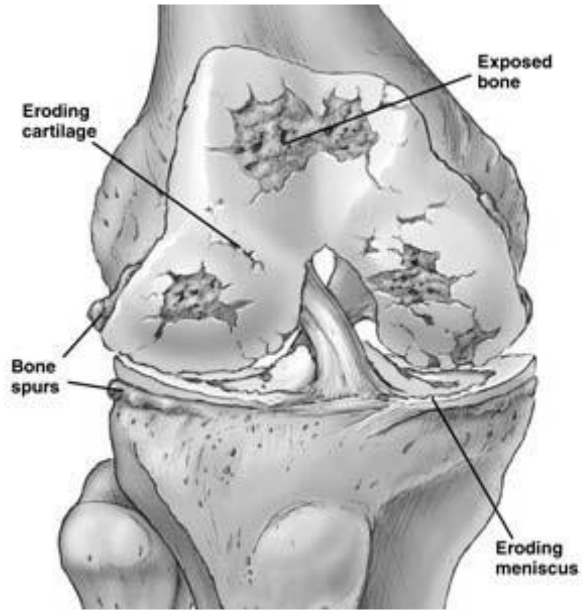

Figure 3.8: *Osteoarthritis of the knee (www.medialkneepain.com).*

In a recent literature review of prevention and management of OA, Takeda and colleagues (2011) concluded that OA is common in sports and that there is no surgical or rehabilitation treatment at present that can avoid development of cartilage damage. Therefore, more emphasis on preventive measures of sport injuries and especially ACL injuries is advisable.

4. Materials and Methods

4.1 Study design

This study is based upon the data from a larger ongoing prospective cohort study at the Oslo Sport Trauma Research Center, aimed at investigating risk factors for non contact ACL injuries. The study was started in 2007. All the players from the female elite league in team handball (Postenligaen), and the respective national team were invited to participate. In 2009 all players from female elite soccer (Toppserien) were invited as well. Additionally all new players at these levels have been invited to participate in the study annually in 2008, 2009 and 2010.

4.2 Participants

For this study the data from 2007 and 2009 were analyzed. The 2007 data consist of 184 team handball players from 14, teams and the 2009 data consist of 187 soccer players from 12 teams.

Players that were fit to participate in regular training without orthosis on testing day were allowed to carry out the tests. Players showing up with an injury, or not capable to carry out the test without the support of an orthosis, were excluded form the analysis. All players with complete data from the tests were included in the analysis. Of the excluded, five players did not complete anthropometric measures due to delay on the test station combined with a preordered flight ticket. In two cases, the tester did not recognize that the test was not successful. This was discovered during processing of the data. In five cases, the balance platform software did not record sufficient data on each trial. This was not discovered before the processing of the data. The complete list of players, both the ones available analysis and ones excluded are displayed in table 4.1.

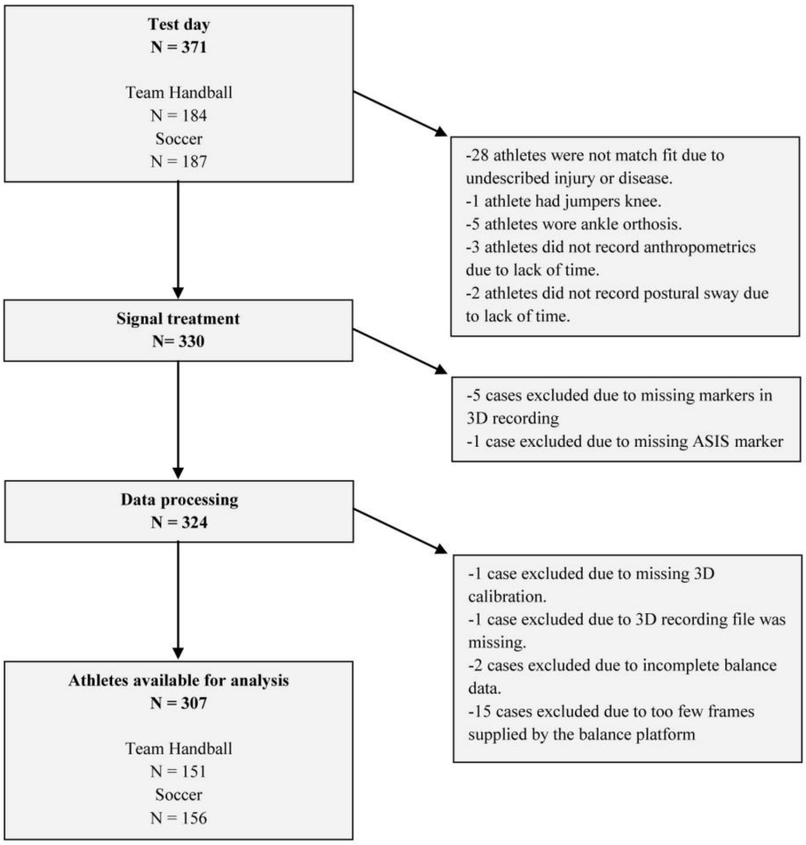

Figure 4.1: *Flowchart of participants.*

4.3 Test procedures

The testing consisted of eight different test stations including anthropometrical measures, balance-, flexibility- and strength tests, 3D recording of drop jump and a cutting task, 2D video recording of drop jump and a questionnaire. All tests were conducted at the same day, with a total duration of approximately seven hours, including a 30 minute lunch. The players were assigned to the test stations in pairs testing approximately 30 min per player on each station. All the tests were conducted at the Norwegian School of Sport Sciences.

All the teams in top level team handball (2007) and soccer (2009) received information about the study (appendix 1) and an invitation for all their players and by e-mail (appendix 2), and thereafter all coaches were contacted to make appointments for testing.

This study is based on the results from two of the test stations. In the following, the PS and VDJ test stations will be presented thoroughly.

4.4 One-leg static balance test

Two balance tests were conducted, but only the results from the first test described are used for analysis in this study. For the tests, the players wore their regular training shorts, shirts and socks. Shoes were not allowed.

For this test, a blue balance-pad (AIREX®, Magister Corporation, Chattanooga, USA) was placed on a balance platform and a 24” BenQ monitor was placed 100 cm in front of the platform (figure 4.2). The player stood on one leg for 20 sec, measuring PS as displayed in figure 4.3. The player was instructed to stand on one leg as still as possible, resting the arms, hands together, in front of the body while looking at a 20 sec countdown on the monitor. The trial was successful when the player succeeded to not move the arms, let the thighs touch or support the leg that was not tested on the balance pad. If the player fell off the platform, did a correctional, second jump or in other way not succeed, the trial was done over again. Leg dominance was determined for each player by asking which leg she would use to kick a ball as far as possible.

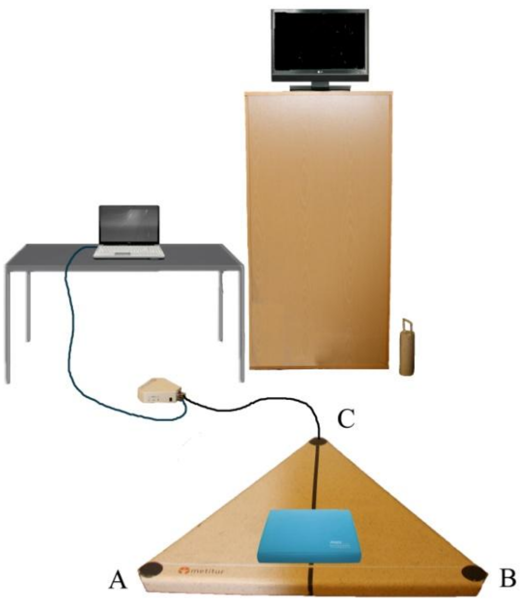

Figure 4.2:*Balance station setup.*

Figure 4.3: *Single leg static test.*

4.4.1 Test station setup

The testing was conducted in a big, well lit room. The test station was sat up using the Good Balance Metitur system which comprises a portable triangular force plate with strain gauge transducers (the balance platform) connected to a three channel A/D converter and a direct-current amplifier which was connected to a computer. All the players were tested on the GB 300, 1200mm, golf edition platform, except 12 team handball players, who were tested on GB 300, 800mm regular platform in 2007 (Good Balance; Metitur, Jyväskylä, Finland). The platform registered the movement of the COM. The balance platform was placed 1m in front of a BenQ G2400W, 24 inch monitor (figure 4.2). The computer ran the Good Balance software (Good Balance 300; Metitur, Jyväskylä, Finland) supplied by the manufacturer of the balance platform. A calibration routine defining the three corners of the balance platform with a 10 kg weight according to manufacturer's instructions was performed at the beginning of each test day, and after every restart.

4.4.2 Signal treatment

The balance platform recorded COP and provides X and Y coordinates, where X-axis is defined between A and B corner of the platform, and the Y-axis perpendicular to the X-axis (figure 4.2). The results from the tests were digitalized at 50Hz (GB 300, 1200mm, golf edition platform) and 200Hz (GB 300, 800mm regular platform), which results in 50 and 200 X and Y coordinates per second. The X and Y coordinates were exported to one text file for each test, and the name manually changed to be recognized by a MATLAB® 2007a script developed by Krosshaug and Killingmo in collaboration (appendix 3). The Matlab script filtered the 200Hz frame rate from the GB 300, 800mm regular platform down to 50Hz, and cut off first and last second of the test fot all the players. The same script calculated the outcome measures were calculated, and exported to one SPSS sheet (Version 18, SPSS Inc., Chicago, IL, USA).

4.5 Vertical drop-jump

Each player was instructed to drop off a 30 cm high box, land with one leg on each force plate (AMTI LG6-4-1, Watertown, MA 02472, USA), and immediately perform a maximum vertical jump with no requirements in matter of technique (figure 4.4). The VDJ was repeated until 5 accepted jumps were recorded. It was required by the player to perform a maximum jump and the jump was considered accepted when the player hit one foot on each platform, and no markers fell off during the recording. For these tests, the players wore only sports underwear, socks and their regular court-training shoes.

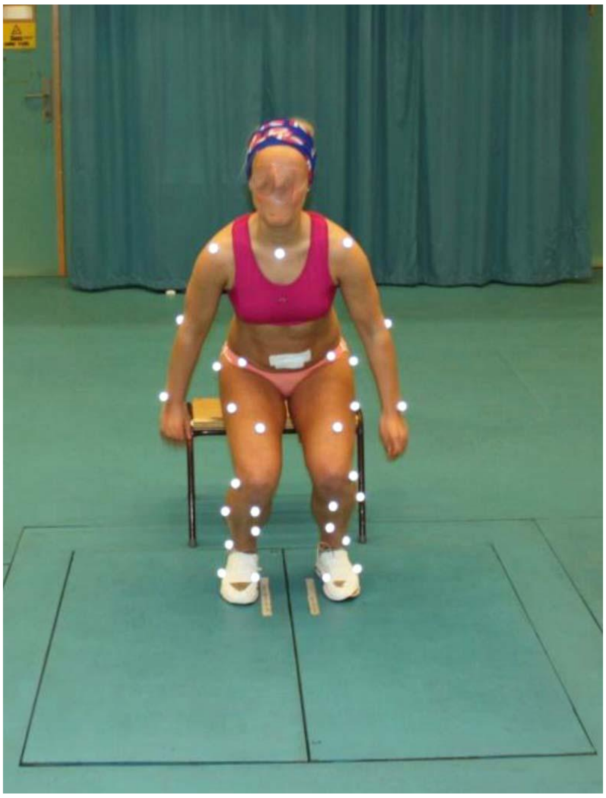

Figure 4.4: *Player executing a vertical drop-jump.*

Anthropometric measures

In order to calculate mass, inertia tensors and COM in the analysis, several anthropometrical measures were needed (appendix 4). For this study, the body was considered to consist of 13 body segments based on the multi-segment model of Yeadon (1990) (figure 4.5, table 4.1) except for the model of the foot, where Zatsiorsky's model (1983) was applied. The body segments marked S where considered as one segment in our model (figure 4.5).

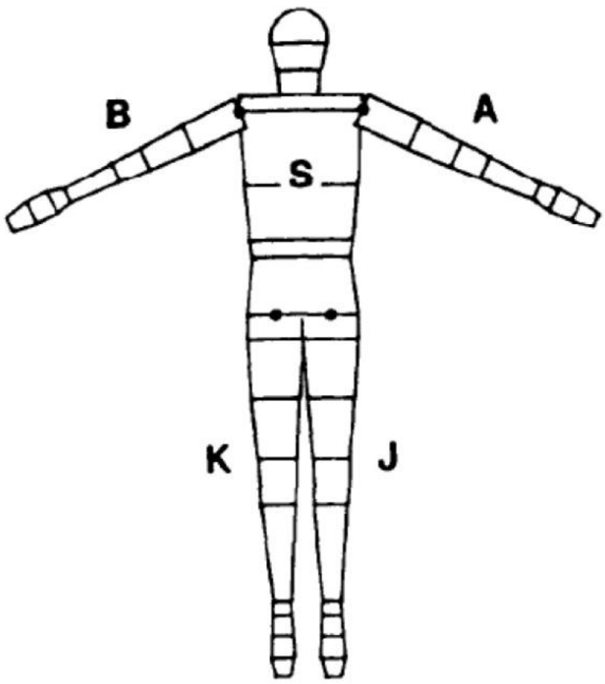

Figure 4.5: *The multisegmental model of Yeadon (1990).*

The different segments perimeters and volume were estimated from the 95 anthropometric measures of length and circumference of each segment by the model of Yeadon (1990) except for the foot, where and Zatiorsky's model was used (Zatsiorsky & Seluyanov, 1983). The measuring was conducted with a tape measure and caliper by physiotherapists to ensure accuracy. For the calculations the shape of the segments were considered to be cone shaped, except the trunkus, which was considered to be a stadium solid (figure 4.6).

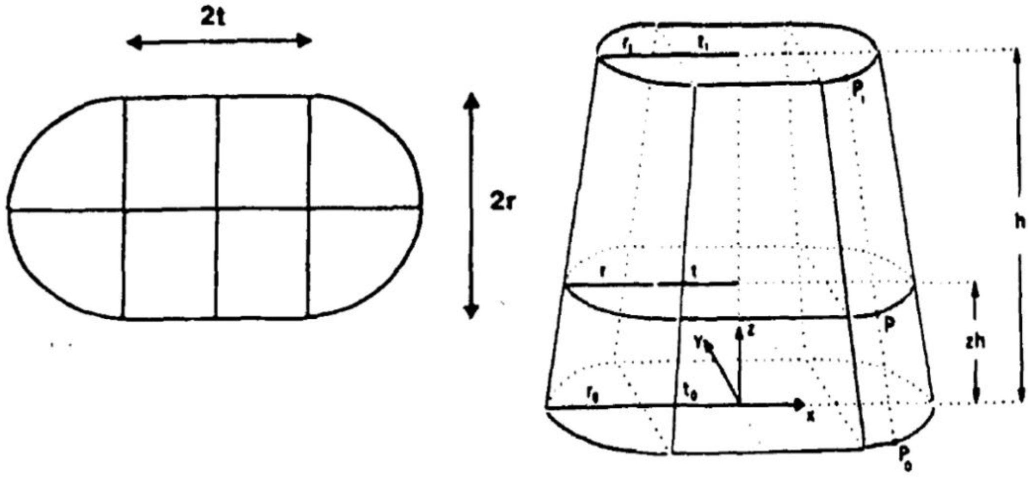

Figure 4.6: *Stadium solid (Yeadon, 1990).*

This provides an estimation of the volume of each segment, and perimeters to calculate primal axes and COM.

Table 4.1*: Segmental sections according to Yeadon (1995), and density according to Dempster(1955).*

Segments	Solids	Segment density (Dempster, 1955)
Torso S	Head neck	1.11
	Shoulders	1.04
	Thorax	0.92
	Abdomen pelvis	1.01
Left arm A	Left upper arm	1.07
	Left fore arm	1.13
	Left hand	1.16
Right arm B	Right upper arm	1.07
	Right fore arm	1.13
	Right hand	1.16
Left leg J	Left thigh	1.05
	Left shank	1.09
	Left foot	1.10
Right leg K	Right thigh	1.05
	Right shank	1.09
	Right foot	1.10

To calculate inertia and COM for each segment, an estimation of the density of each segment is required in addition to a volume estimate. These density values, obtained from the model of Dempster, are displayed in table 4.1 (Dempster, 1955). The mass, COM and inertia parameters are needed for calculating joint angles and moments.

Marker placement

There were thirty-five nine mm reflective markers to be attached to specific palpable anatomical landmarks on each player (table 4.2, figure 4.7). First, in the procedure, the palpable anatomical landmarks were identified by the directions in table 4.2. Then the area was cleaned with isopropanol and marked with a waterproof pen. Then the markers were attached to the player's body with adhesives. To ensure precise positioning the procedure was performed by physiotherapists that followed written procedures. At least four reflective markers were attached to each body segment to be able to estimate the movement of the segment (Soderkvist & Wedin, 1993). Fabric on clothing and shoes, that could produce interference on the camera recording, was covered by Strappal® adhesive sport tape.

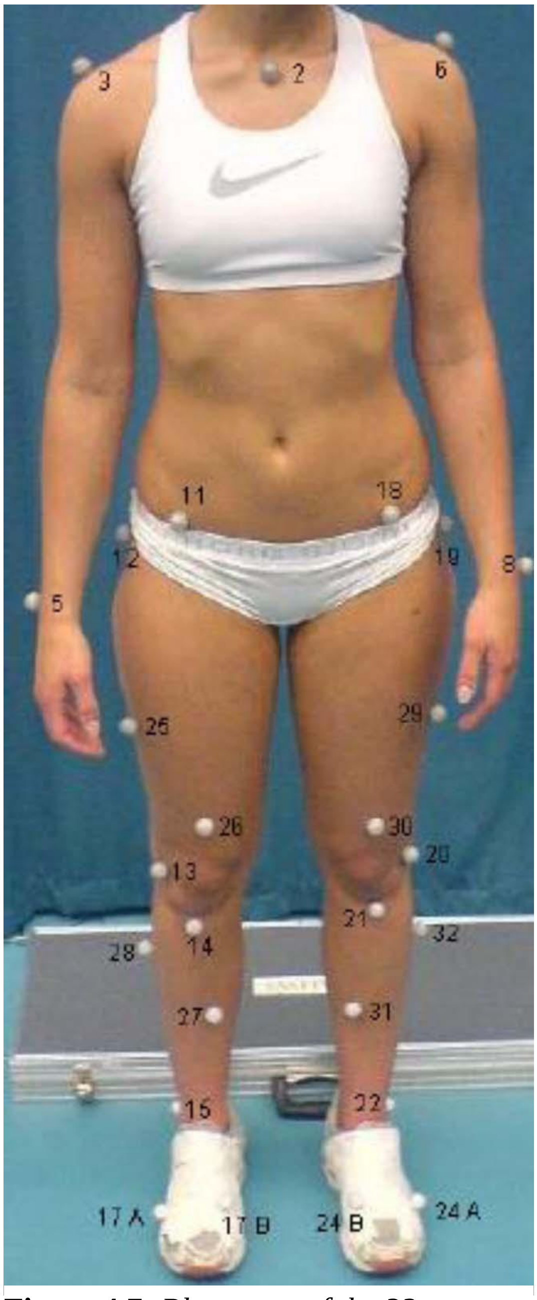

Figure 4.7: *Placement of the 32 markers. Some are not visible from this angle.*

Table 4.2: *Marker placement and directions.*

Number	Marker	Palpable landmark
1	C7 (hidden)	C7 spine.
2	Insicura jugularis	Insicura jugularis.
3	Right shoulder	The most lateral point of the lateral margin of the acromion.
4	Right elbow	Lateral epicondyle , the most distale point.
5	Right wrist	Caput ulnae, the most distale point.
6	Left shoulder	The most lateral point of the lateral margin of the acromion.
7	Left elbow	Lateral epicondyle, the most distale point.
8	Left wrist	Caput ulnae , the most distale point.
9a	PSIS (hidden)	Posterior superior iliac spine.
9b	PSIS (hidden)	Posterior superior iliac spine.
10	Middle back (hidden)	Two thirds of the distance measured from C7 and midpoint between PSIS.
11	Right ASIS	Anterior superior iliac spine , the inferior point.
12	Right trochanter	The most superior point of trochanter major.
13	Right knee	Lateral epicondyle on femur.
14	Right tuber	Tuberositas tibiae, the most prominent point.
15	Right ankle	Lateral malleolus, the most lateral point.
16	Right heel	The most posterior point of calcaneous (In the line of achilles tendon on shoe).
17a	Right 5th toe	The caput of 5th metatarsal (on shoe).
17b	Right 1st toe	The caput of 1st metatars (Adjusted in lateral direction on shoe to avoid kicking the marker off).
18	Left ASIS	Anterior superior iliac spine , the inferior point.
19	Left trochanter	The most superior point of trochanter major.
20	Left knee	Lateral epicondyle on femur.
21	Left tuber	Tuberositas tibiae, the most prominent point.
22	Left ankle	Lateral malleolus, the most distale point.
23	Left heel	The most posterior point of calcaneous (In the line of achilles tendon on shoe).
24a	Left 5th toe	The caput of 5th metatarsal (on shoe).
24b	Left 1st toe	The caput of 1st metatars (adjusted in lateral direction on shoe to avoid kicking the marker off).
25	Right thigh lateral	The mid distance between "Right trochanter" and "Right knee" markers and about 2 cm anterior direction.
26	Right thigh front	The mid distance between "Right thigh lateral" and "Right knee" markers and in anterior direction to the center of front thigh.
27	Right shank tibia	The mid distance from "Right tuber" and "Right ankle" markers on tibia (medial side).
28	Right shank lateral	The mid distance from "Right tuber" and "Right shank tibia" and most lateral point of the shank.
29	Left thigh lateral	The mid distance between "Left trochanter" and "Left knee" markers and about 2 cm anterior direction.
30	Left thigh front	The mid distance between "Left thigh lateral" and "Left knee" markers and in anterior direction to the center of front thigh.
31	Left shank tibial	The mid distance from "Left tuber" and "Left ankle" markers, on tibia (medial side).
32	Left shank lateral	The mid distance from "Left tuber" and "Left shank tibia" and most lateral point of the shank.

Static recording

Prior to the VDJ a static recording of the player, standing in an anatomical position, facing X direction on the Global Coordinate System (GCS), was performed (figure 4.8). The static recording was conducted to derive the anatomical axis of the segments in order to establish the three dimensional relationship between the reflective markers and the anatomical axis of each segment.

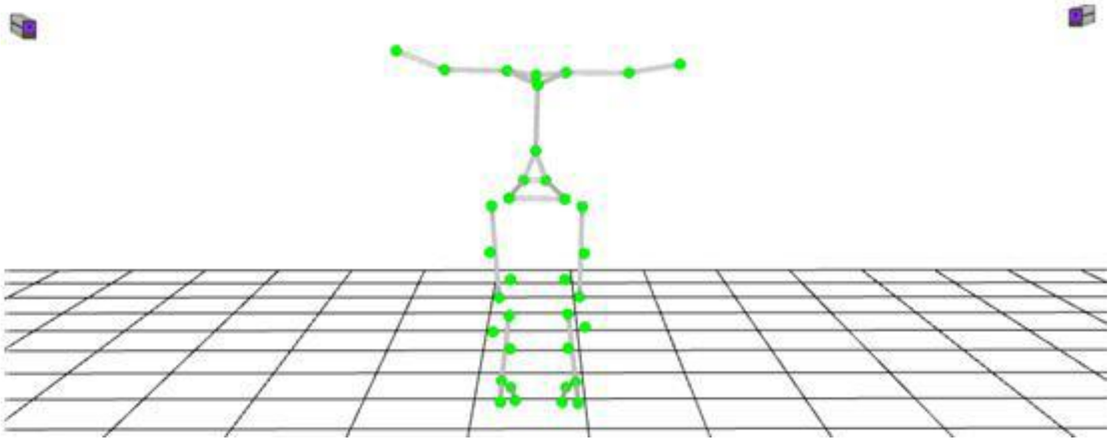

Figure 4.8: *Image of a static recording of an player from Qualisys Track Manager.*

4.5.1 Test station setup

Cinematographical system

The laboratory setup consisted of a 30cm high box on an indoor court floor surrounded by eight cameras organized in an optical tracking system (ProReflex, Qualisys INC Gothenburg, Sweden).

The cameras were emitting infrared lights at a recording frequency of 240Hz, and were recording the reflections from the reflective markers. Optimal positioning of the cameras was sought to achieve as high precision of the marker placement as possible. Marker placement precision is dependent of the number of cameras the marker is visible in (the more the better), and the intersection angle of the recording cameras (90° optimal). Figure 4.9 display the laboratory camera setup.

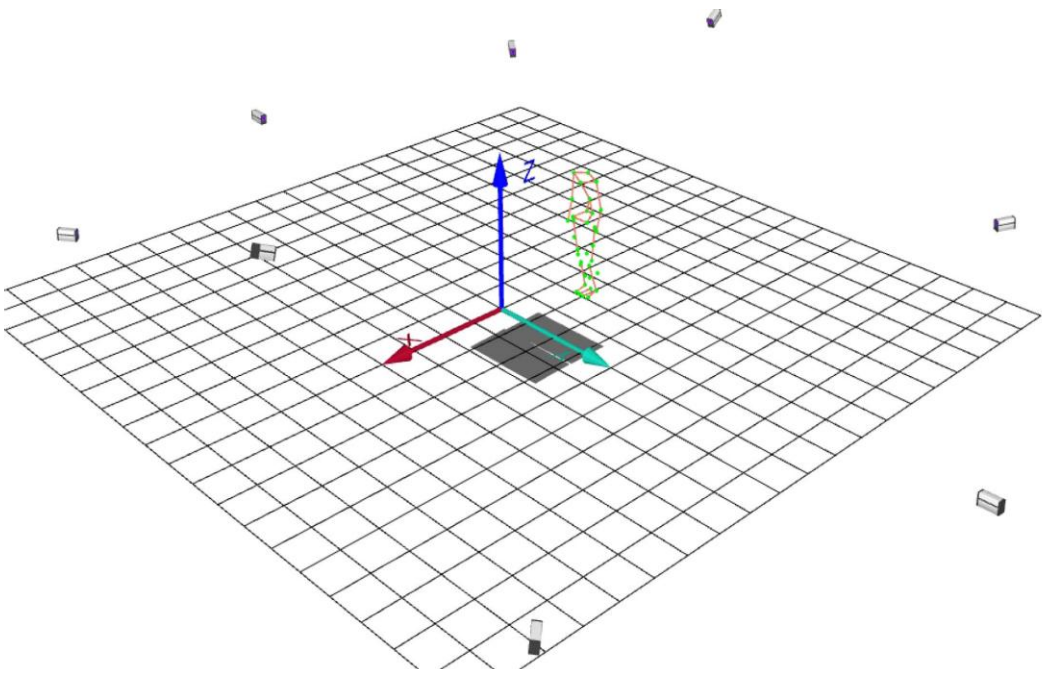

Figure 4.9: *Camera setup for the optical tracking system.*

The calibration procedure to correct for lens distortion, and to define the recording area/grid, was provided by the manufacturer of the system (Qualisys INC Gothenburg, Sweden). The procedure was a two-stage calibration. First a physical frame with four reflective markers attached was placed onto/around the force platform in center of the recording area/grid. Secondly a calibration rod, with one reflective marker in each end, was moved around in the area/grid by one tester.

This calibrated the recording area to a GCS with the X-axis defined as positive forwards, Z-axis vertically and Y-axis perpendicular to the X-axis in a right side orientation. The calibration algorithm was not provided by the manufacturer and therefore unknown. This calibration procedure was performed previous to the testing of each player.

Force platform

In the center of the recording area/grid there were two 60x122 cm^2 strain gauge force platforms (AMTI LG6-4-1, Watertown, MA 02472, USA). These force platforms measure the GRF as six components. The coordinate system was oriented according to the GCS, with origin 5cm down from surface of the floor, at the center of the force platform. The readings were amplified times 1000 and digitalized at a sample rate of 960Hz by an A/D converter. Then the recording was transmitted to the computer. A

reading without any loading on the force platforms was used to calibrate offset according to manufacturer's calibration values. This was preformed previous to the testing of each player.
The force platform data and the marker recordings were automatically synchronized by the Qtrac ver.2.02 software.

4.5.2 Signal treatment

Marker recordings

When the data recording was complete, the Qtrac software (Version 2.02, Qualisys AB, Gothenburg, Sweden) provided calculation of the 3D trajectories of each marker. To complete the 3D trajectory calculation each marker in the Qtrac software (Version 2.02, Qualisys AB, Gothenburg, Sweden) had to be manually identified according to previously described marker placement. The next step was to filter the date to remove noise or errors associated with the marker signals due to incorrect digitalization, skin movement or similar errors. The data was filtered by fitting the marker trajectories to a smoothing spline, by the fortran package developed by Woltring (1986). The same procedure interpolated missing trajectories of up to 12 frames.

Force plate
Force plate signals were converted form Volt to Newton according to the manufacturer's values and filtrated at 15Hz by a low-pass-Butterworth filter to filtrate high frequency noise.

4.5.3 3D motion analysis

Coordinate systems

Global Coordinate System
The GCS is the frame used to describe orientation and movement of the each segment in relation to each other. The coordinate system was sat up as previously described in the Laboratory setup.

Segment embedded frames

The segment embedded frames were defined for each segment by the method of Söderkvist (1993) by three body markers of each segment, and the estimated proximal joint center. These coordinate systems were needed to calculate the movement of each segment in between frames. In the following the joint center definitions for each joint is presented.

Joint center definitions

Ankle

The joint center of the ankle was positioned one centimeter distal to the midpoint between lateral and medial malleoli according to Eng and Winter (1995).

Knee

The joint center of the knee was positioned on the midpoint between the epicondyles of the femur as described by Davis and colleagues (1991). The knee joint center was assumed to lay parallel to the Y-axis in the GCS.

Hip

The joint center of the hip was defined as suggested by Bell (1990) which combines the two most accurate measures from Andriachhi and Tyklowski. The anterior-posterior location is set by Andriachhi's method predicting the hip center location within 0.73 cm of true location, and the frontal-plane location by Tyklovskis method estimating the hip center within 0.79 cm of true location. By combining these measures Bell found that the three dimensional location of the hip center can be predicted within 1.07cm of true location.

Dynamic calculations

The calculation of the anatomical axes and segment position in each frame was conducted by multiplying the motion with the previous frame by the decomposition method of Söderkvist and Wedin (1993). The segmental axes were determined by four markers on each segment (foot, shank, thigh and pelvis). These axes and the motion were required to calculate joint angles.

Joint angles

A Joint Coordinate System (JCS) was established for each segment as developed by Grood and Suntay (1983). It is a right hand oriented coordinate system where each joint is considered to have three rotations.

The JCS is made up by one axis in the frontal-plane derived from the proximal segment, the Y-axis, one axis derived from the longitudinal oriented axis in the distal segment, the Z-axis and lastly a floating axis perpendicular to the two previous axes, the X-axis. The coordinate system is not orthogonal and the Y- and Z-axes are not necessarily perpendicular. The relative rotations between two of the segments are thought of as rotations of one segment around its own axis, while the other segment is stationary. The magnitudes of these rotations are measured by the angles between the floating axis and a reference line embedded in the segment. The 3rd relative rotation around the floating axis is measured as the angle between the two fixed axes of the two segments.

The first rotation in the order of rotations set by International Society of Biomecanics is flexion/extension which is rotation around the Y-axis, the next rotation is internal/external rotation around the Y-axis , and lastly rotation is abduction/ adduction around the X-axis, (Wu et al., 2002).

Mathematical model and calculations

All the following estimations and calculations were executed by a MATLAB® 2007a script written by Oslo Sport Trauma Research Center for use in this project.

Frontal-plane projection angles, peak valgus angles and peak valgus moments were calculated during the contact phase. The contact phase was defined by a cutoff of 10N.

Inverse dynamics

The mathematical model for estimating joint forces and moments used in this study is called inverse dynamics. This model estimate joint force and moments based on mathematical calculations called iterative Newton/ Euler method (Bresler & Frankel, 1950). The joint forces and moments are calculated for each joint separately, iteratively calculating from most distal to most proximal joints. In this study the joints calculated are ankle, knee and hip. The estimations are based on body segment kinematics (position, orientation and acceleration), the kinetic variables form the anthropometric (COM, inertia) and measurements (GRF) or estimations of forces and moments. To

calculate these figures some assumptions had to be made. Each body segment is considered as a rigid body, and each joint is considered to have three degrees of freedom and frictionless motion.
The Newton/ Euler method is based on expanded equations of Newton's second law of motion.

$$F = ma$$

F = Force, m = Mass, a = Acceleration

For each body segment in the model, the Newton Euler method would be formulated:

Newton (linear) = $\Sigma\bar{F} = m\bar{a}_{com}$

$\Sigma\bar{F}$ = Sum of forces action of the segment, m = Mass of the segment, $\bar{a}_{com}$ = Acceleration vector of COM.

To be able to calculate the forces acting on the proximal joint, the equation is expanded:

$$\bar{F}_p + \bar{F}_d + \bar{G} = m\bar{a}_{com}$$

$\bar{F}_p$ = Forces acting on the proximal joint (the unknown), $\bar{F}_d$ = Forces acting on the distal joint (known form GRF or previous equation), $\bar{G}$ = Constant gravity vector acting on COM.

This makes the equation more comprehensible. The m is known from the estimation model described previously and the $\bar{a}_{com}$ is known from the optical tracking.

The $\bar{F}_d$ is retrieved from GRF when calculation the forces of the ankle joint. The GRF can be used because the mutual forces of action and reaction between two bodies are equal, opposite and collinear, by Newton's third law of motion. When calculating forces for knee and hip, the $\bar{F}_d$ is known from the results of the same equation on the more proximal joint. The unknown, $\bar{F}_p$ gives the force acting on the joint which is currently calculated.

Thereafter, the angular velocity and angular moments can be calculated. This is done by using the Euler equation for angular velocity and moments:

$$\Sigma\bar{M} = \bar{\bar{I}} \times \bar{\varpi} + \varpi \times (\bar{\bar{I}} \times \varpi), \bar{\bar{I}} = \begin{bmatrix} I_x & -D_{xy} & -D_{xz} \\ -D_{xy} & I_y & -D_{yz} \\ -D_{xz} & -D_{yz} & I_Z \end{bmatrix}$$

$\Sigma\bar{M}$= Sum of moments acting on the segment, $\bar{\bar{I}}$ = The inertia tensor, I_x, I_y, I_Z= Moments of inertia, $-D_{xy}$, $-D_{xz}$, $-D_{yz}$= Products of inertia, $\bar{\varpi}$ =Angular acceleration vector, ϖ = Angular velocity

This equation is expanded to suite the purpose in this study:

$$\bar{M}_d + \bar{M}_p + \bar{r}_d \times \bar{F}_d + \bar{r}_p \times \bar{F}_p = \bar{\bar{I}} \times \bar{\varpi} + \varpi \times (\bar{\bar{I}} \times \varpi)$$

$\bar{M}_d$=joint muscle moment of the distal joint, $\bar{M}_p$=joint muscle moment of the proximal joint (the unknown), $\bar{r}_d$= displacement vector to the distal joint center, $\bar{r}_p$= displacement vector to the proximal joint center, $\bar{F}_p$= Force vector on the proximal joint, $\bar{F}_d$= Forces vector on the distal joint (figure 4.10).

$\bar{M}_p$, the unknown, is the moment acting on the joint which is currently calculated. $\bar{M}_d$ is known because of Newton's third law of motion, that states that mutual forces of action and reaction between two bodies are equal and opposite. Following that, $\bar{M}_d$ was known form GRF for the ankle joint, and for the previous equation when calculating the knee and hip. The $\bar{r}_p$ and $\bar{r}_d$ are the displacement vectors from COM to the proximal and distal joint center as displayed in the figure x, the inertia tensor is calculated from the segmental model of Yeadon (1990) and the angular acceleration vector is found by differentiating the three orthogonal components of angular velocity. The $\bar{F}_d$ and $\bar{F}_p$ are derived from the calculations of the Newton (linear) equation.

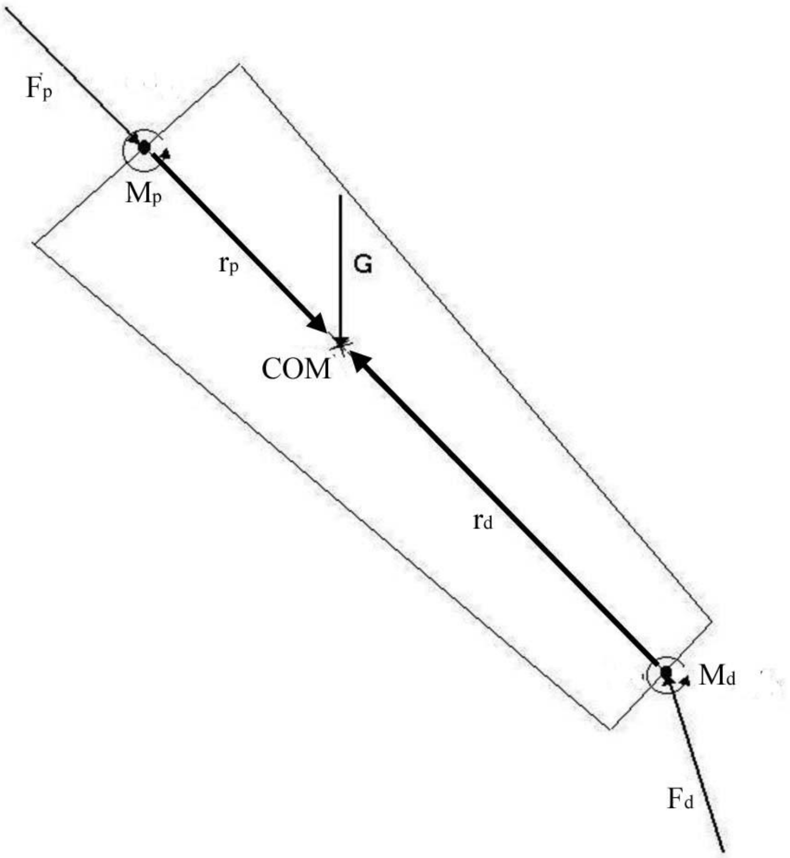

Figure 4.10: *Drawing of a segment. M_d = distal joint muscle moment, M_p = proximal joint muscle moment, r_d= distal displacement vector, r_p= proximal displacement vector, F_p = proximal force vector, F_d= distal force vector, COM= center of mass.*

From these equations, calculated joint by joint the results are derived and then exported to SPSS (Version 18, SPSS Inc., Chicago, IL, USA).

4.6 Data reporting

PS measures were quantified to mediolateral (ML) velocity (m/sec), anterioposterior velocity (AP) (m/sec), total distance of measured sway (mm) and the area of the smallest ellipse that contains 95% of the measured PS (mm^2).

From VDJ the mean of the three last approved jumps were used. The knee join kinetic and kinematic values were frontal-plane projection angle (°), maximum valgus angle (°) and maximum valgus moment (Nm/kg). Frontal-plane projection angle is the difference from IC to peak valgus angle in a 2d frontal-plane view. Valgus angle and valgus moments were calculated from the global coordinate system. We chose valgus angle and compression forces as positive values.

Valgus moments were calculated by combining the recording of anthropometrics, GRF and limb alignment during the drop jumps (from the optical tracing system) through use of inverse dynamics. The period of contact was defined as the time GRF exceeded 10N.

Table 4.3: *List of abbreviation.*

Abbreviations	Description (units)
ML velocity	Velocity in mediolateral direction (m/sec). Mean of three trials.
AP velocity	Velocity in anterioposterior direction (m/sec). Mean of three trials.
Distance	Total distance of measured sway (mm). Mean of three trials.
Area	The area of the smallest ellipse that contains 95% of the measured PS (mm^2).
FPPA	Frontal-plane projection angle from IC to max angle in 2d frontal-plane (°). Mean of three trials.
Valgus angle	Peak valgus angle (°).Mean of three trials.
Valgus moment	Peak valgus moment (Nm/kg) Mean of three trials.

4.7 Statistics

The analysis was calculated using SPSS (Version 18, SPSS Inc., Chicago, IL, USA).

A t-test was carried out to control normality distribution. To investigate the correlation between the independent variables a simple bivariate correlation analysis was conducted. This discovered an $R \geq \pm.7$ between ML velocity, AP velocity and distance, whereas area had an $R \leq \pm .2$. This indicates that it is not advisable to combine these independent variables in a multiple regression (Pallant, 2007). Height did not correlate with the PS and is not controlled for in the regression analysis.

To investigate the relationship between the independent variables and the dependent variables simple univariate linear regression calculations were conducted. We calculated one regression equation for each combination of knee joint kinetic and kinematics measure and PS. Confidence interval of 95% was used to judge statistical significance in all models. The proportions of variance attributed to the frontal plane variables by one SD change in the PS measures are presented in a bar chart, where only the statistically significant associations are displayed.

To investigate the intra-rater reliability of the Good Balance platform 13 players were tested twice; once in the beginning of the test day, and once at the end of the test day. These results were first assessed by Bland-Altman plots (Bland & Altman, 1986; Yeadon, 1990) where the reliability is considered valid when 95% of the plots are within 2SD the reliability at test is considered valid (Bland & Altman, 1986). Secondly, the ICC (Intraclass Correlation Coefficient) was calculated. In general, the reliability is considered high when the ICC coefficients above 0.90, while coefficients between 0.80 to 0.89 are considered moderate and below 0.80 are considered questionable when assessing physiological data (Vincent, 2005).

4.8 Ethics

The project was approved by the *Regional Committee for Medical Research Ethics* (appendix 5) and reported to *Norwegian Social Science Data Services* (appendix 6). The players were covered by a specific insurance policy for any injuries during testing (0398160/DnB NOR).

All the players had signed an informed consent, stating that they are volunteers, and that they could withdraw from the project at any given moment (appendix 7). Players under age of 18 had guardian consent.

The risk of injury during testing was considered equal or lower than the risk of injury in regular team training, and less than competition.

5. Results

The t-test display that team handball players were significantly higher and heavier than the soccer players.

Table 5.1: *Player characteristics.*

			Mean	SD	Range
Team handball	N = 151	Age (yrs)	22.2	3.9	16.2 - 36.8
		Height (m)	1.72	0.07	1.56 - 1.89
		Body mass (kg)	69.1	7.2	51.8 - 85.4
		BMI	23.2	1.7	19.6 - 27.9
Soccer	N = 156	Age (yrs)	22.0	4.3	16.0 - 37.0
		Height (m)	1.67	0.05	1.52 - 1.79
		Body mass (kg)	62.2	6.5	48.0 - 80.0
		BMI	22.4	1.9	18.2 - 28.8

Team handball players display 40% higher valgus angle in dominant leg and 32% higher valgus angle in non-dominant leg during VDJ than the soccer players. In PS measures, the team handball players have higher values in all measurements, especially ML velocity where the mean values from the single leg static test are 38% and 39% higher for team handball players, dominant leg and non-dominant leg respectively. Soccer players have higher maximum values and wider range of FPPA than team handball players and this is reflected in the mean (table 5.2). The mean valgus angles are lower for soccer players than for team handball players, and so is the SD, while the ranges are similar (table 5.2). The mean valgus moments show minimal variance across dominant and no dominant leg and across sports (table 5.2).

Table 5.2: *Mean SD, range (min-max), difference between groups and level of significance for joint kinetics and kinematics,and PS. Joint kinetic and kinematic variables are frontal-plane projection angle (FPPA) (difference in angle in the frontal plane from IC to maximum angle), valgus angle (peak valgus angle) and valgus moment (peak valgus moment). PS measures are area (95% ellipse area), ML velocity (mean X direction velocity), AP velocity (mean Y direction velocity) and distance (mean distance of PS).*

		Team handball N = 151			Soccer N = 156			Independent T-test	
		Mean	**SD**	**Range**	**Mean**	**SD**	**Range**	**Mean difference**	**Sig. (2-tailed)**
Dominant leg	FPPA (°)	7.8	4.6	0.0 - 26.4	8.9	5.9	0.0 - 31.1	-1.1	0.07
	Valgus angle (°)	11.1	4.8	-0.8 - 26.2	6.7	6.1	-7.5 - 20.9	4.5	0.00
	Valgus moment (Nm/kg)	0.4	0.2	0.1 - 1.5	0.3	0.1	0.0 - 0.9	0.0	0.19
Non-dominant leg	FPPA (°)	8.6	4.3	0.0- 19.3	9.3	5.6	0.0 - 31.5	-0.7	0.23
	Valgus angle (°)	11.1	5.5	-1.7 - 27.1	7.9	6.0	-6.2 - 22.8	3.3	0.00
	Valgus moment (Nm/kg)	0.4	0.2	0.0 - 1.0	0.3	0.1	-0.1 - 0.7	0.1	0.00
Dominant leg	ML velocity (m/sec)	30.4	8.4	15.7 - 58.2	18.8	4.4	10.8 - 39.0	11.6	0.00
	AP velocity (m/sec)	28.3	6.4	16.2 - 47.3	21.0	4.7	11.1 - 37.6	7.3	0.00
	Distance (mm)	828	201	481 - 1406	560	121	309 - 1080	268	0.00
	Area (mm^2)	982	312	427 - 1852	781	254	310 - 1755	201	0.00
Non-dominant leg	ML velocity (m/sec)	30.1	8.1	16.5 - 56.5	18.5	4.1	10.9 - 33.4	12.0	0.00
	AP velocity (m/sec)	28.0	6.8	15.4 - 52.3	21.0	4.5	12.2 - 38.3	7.0	0.00
	Distance (mm)	820	204	468 - 1556	555	113	352 - 995	265	0.00
	Area (mm^2)	957	318	410 - 2249	772	245	334 - 1601	186	0.00

Frontal-plane projection angle and PS

The only statistically significant associations between FPPA and PS measures were ML velocity, distance and area for non-dominant leg, in team handball players. The R^2 coefficients for the statistically significant associations were 0.054 for ML velocity, 0.036 for distance and 0.032 for area.

Table 5.3: *Univariate regression summary when predicting FPPA based on PS variables.*

		Team handball			Soccer		
	PS	**B**	**(SE)**	**p**	**B**	**(SE)**	**p**
Dominant leg	ML velocity (m/sec)	0.002	0.045	0.966	-0.054	0.034	0.118
	AP velocity (m/sec)	0.005	0.059	0.936	0.077	0.101	0.448
	Distance (mm)	0.000	0.002	0.932	0.000	0.004	0.922
	Area (mm^2)	0.000	0.001	0.870	0.001	0.002	0.627
Non-dominant leg	ML velocity (m/sec)	0.122	0.042	0.004	0.029	0.109	0.789
	AP velocity (m/sec)	0.079	0.051	0.122	0.024	0.100	0.814
	Distance (mm)	0.004	0.002	0.019	0.001	0.004	0.786
	Area (mm^2)	0.002	0.001	0.028	0.003	0.002	0.124

Valgus angle and postural sway

There was only one statistically significant association between valgus angle and PS measures (table 5.4). The association of area, non-dominant leg in team handball players is statically significant by p-value of 0.05. The R^2 coefficient of area in non-dominant leg for team handball players was 0.26.

Table 5.4: *Univariate regression summary when predicting valgus angle based on PS variables.*

		Team handball			Soccer		
	PS	**B**	**(SE)**	**p**	**B**	**(SE)**	**p**
Dominant leg	ML velocity (m/sec)	0.000	0.047	0.999	-0.123	0.110	0.266
	AP velocity (m/sec)	-0.023	0.062	0.707	-0.038	0.103	0.716
	Distance (mm)	0.000	0.002	0.874	-0.003	0.004	0.462
	Area (mm^2)	0.000	0.001	0.889	0.000	0.002	0.914
Non-dominant leg	ML velocity (m/sec)	0.070	0.055	0.209	0.094	0.117	0.421
	AP velocity (m/sec)	0.094	0.065	0.153	0.029	0.107	0.787
	Distance (mm)	0.003	0.002	0.158	0.002	0.004	0.578
	Area (mm^2)	0.003	0.001	0.050	0.787	0.002	0.277

Valgus moment and postural sway

The univariate regression equations show no significant associations between valgus moments and PS (table 5.5).

Table 5.5: *Univariate regression summary when predicting valgus moment based on PS variables.*

		Team handball			Soccer		
	PS	B	(SE)	p	B	(SE)	p
Dominant leg	ML velocity (m/sec)	-0.001	0.002	0.543	-0.001	0.003	0.739
	AP velocity (m/sec)	-0.004	0.003	0.084	0.000	0.002	0.993
	Distance (mm)	0.000	0.000	0.270	0.000	0.000	0.911
	Area (mm^2)	0.000	0.000	0.364	0.000	0.000	0.526
Non-dominant leg	ML velocity (m/sec)	0.001	0.002	0.667	-0.001	0.002	0.647
	AP velocity (m/sec)	0.000	0.002	0.838	-0.002	0.002	0.281
	Distance (mm)	0.000	0.000	0.735	0.000	0.000	0.388
	Area (mm^2)	0.000	0.000	0.594	0.000	0.000	0.599

Proportions of variance

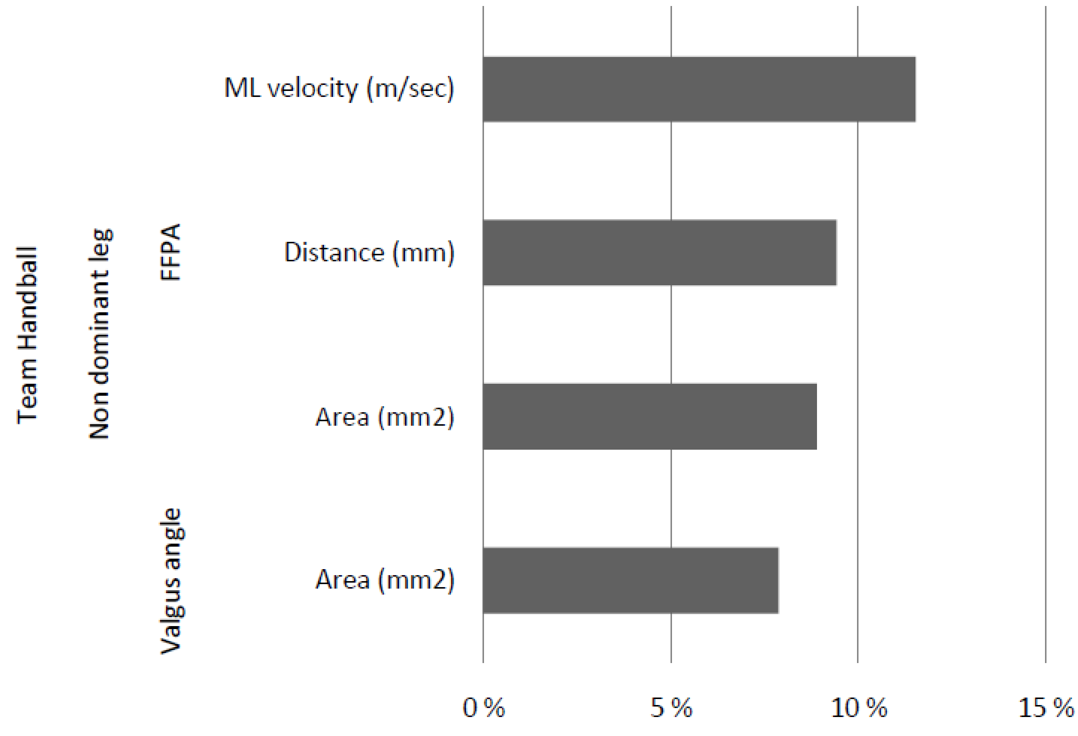

Figure 5.1: *Percent change in FPPA and valgus angle for team handball players in non-dominant leg, when the PS measures were increased 1SD. Only the statistically significant associated PS measure (ML velocity, distance and area) were displayed. All PS measures results in increase of the FPPA and valgus angle.*

Knee joint kinetics and kinematics

The FPPA through the VDJ is displayed in figure 5.2. The peak abduction angle for team handball players occurs after approximately 40-50 % of the contact phase and displays approximately 2.5° abduction. The peak mean value abduction angle for soccer players occurs after approximately 50-70 % of the contact phase and displays approximately -2.5° adduction.

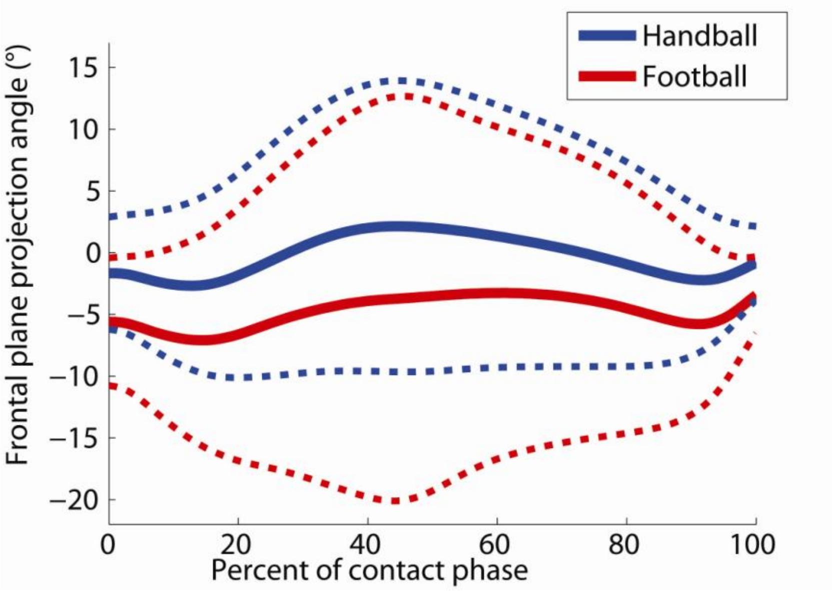

Figure 5.2: *The FPPA through the complete contact phase. Team handball and soccer mean displayed as an unbroken line ±1SD displayed as scattered line. The curves display dominant and non-dominant leg together. Positive values are abduction angle.*

Both team handball and soccer display two valgus angle peaks during the contact phase (figure 5.3). The first and lowest peak is approximately after 25-35 % of the contact phase for the team handball and 20-30 % for the soccer players. The value for the first peak is approximately 9.5° for the team handball players and 2.5° for the soccer players. These peaks probably represent the eccentric phase of the VDJ, while the second peak, which occurs at approximately 60-75 % of the total contact phase for the team handball players and at approximately 70-80 % for the soccer players, which is probably represent the concentric phase of the VDJ. The peak values at this peak are approximately 10.5° for team handball players and 4.5° for soccer players.

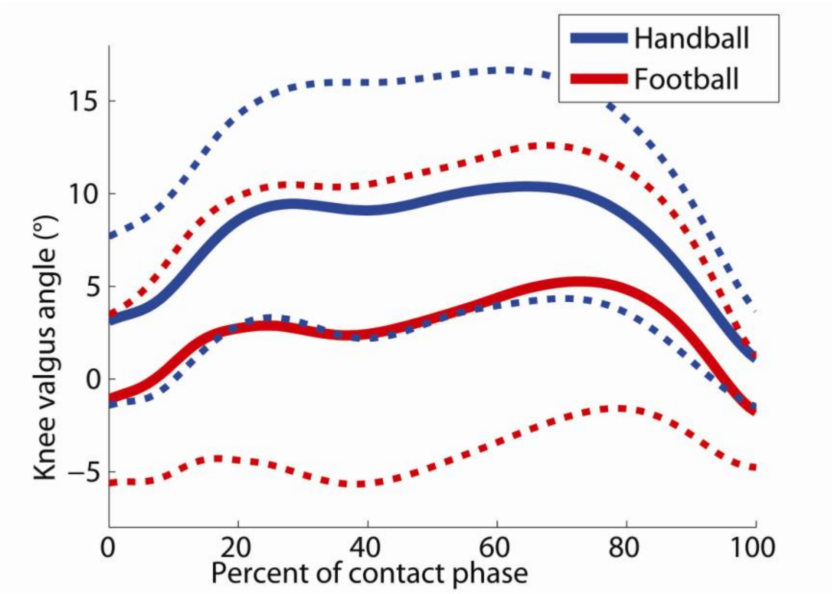

Figure 5.3: *The valgus angle through the complete contact phase. Team handball and soccer mean displayed as an unbroken line ±1SD displayed as scattered line. The curves display dominant and non-dominant leg together. Positive values are abduction angle.*

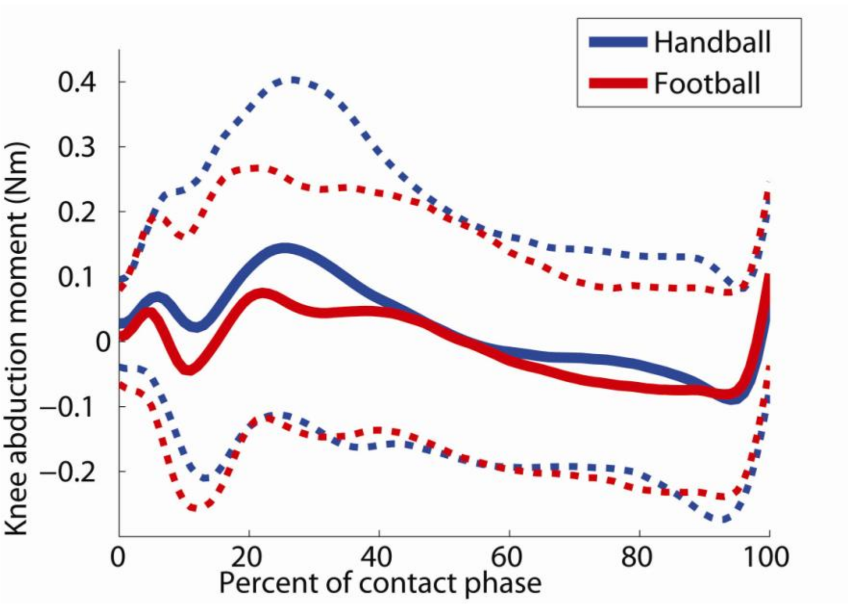

Figure 5.4: *The valgus moment (Nm/kg) through the complete contact phase. Team handball and soccer mean displayed as an unbroken line ±1SD displayed as scattered line. The curves display dominant and non-dominant leg together. Positive values are abduction angle.*

The valgus moment peak, approximately 0.15Nm/kg, was at 30 % of the total contact phase for team handball players (figure 5.4). For soccer players the valgus moment peak, approximately 0.10Nm/kg, was at 25 % of the total contact phase.

5.1 Intra-rater reliability of Good balance static test

All the 13 players perform within ± 2SD. Table 5.6 show the percent of players performing within 2SD.

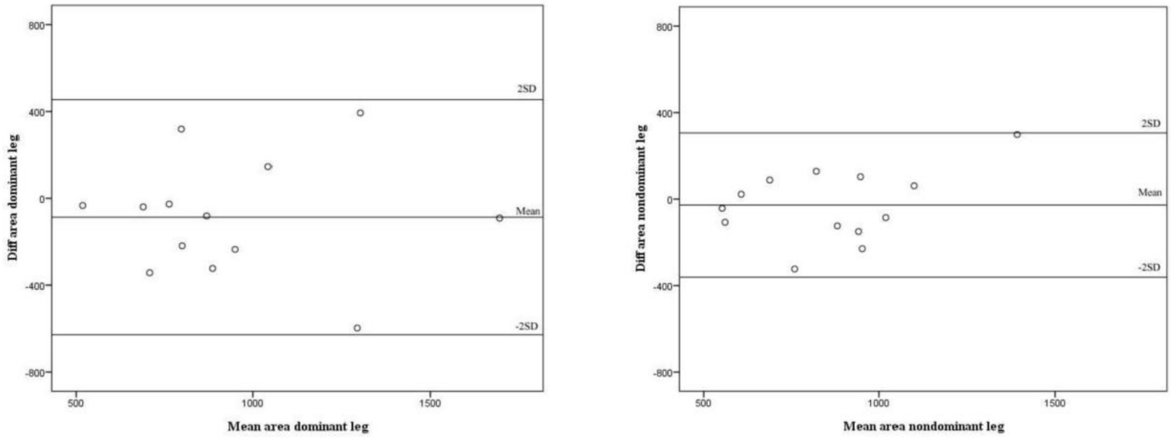

Figure 5.5: *Bland-Altman plot (1986). Dominant leg and non-dominant leg. X-axis is mean area (mm2) of test 1 and 2, and Y-axis is the difference in area (mm2) between test 1 and 2. Reference line at mean difference and ±2SD from mean diff area.*

Table 5.6: *Bland Altman results, mean difference, 2SD and % of players within 2SD.*

		Mean	2SD	%
Dominant leg	ML velocity (m/sec)	0.630	8.913	100.00
	AP velocity (m/sec)	0.951	7.098	92.31
	Distance (mm)	39.084	218.825	92.31
	Area (mm^2)	-86.936	541.445	100.00
Non-dominant leg	ML velocity (m/sec)	0.405	7.297	92.31
	AP velocity (m/sec)	2.057	5.708	92.31
	Distance (mm)	18.475	172.101	92.31
	Area (mm^2)	-27.808	333.735	100.00

To further investigate the intra-rater reliability an intra-class correlation coefficient (ICC) analysis was carried out. Table 5.72 shows that all ICC values are above 0.80 except area, which is at 0.69 in dominant leg and 0.78 in non-dominant leg. The results vary little depending on leg dominance.

Table 5.7: *The ICC and 95% confidence interval (CI).*

		ICC	**95%CI**
Dominant leg	ML velocity (m/sec)	0.82	0.51 - 0.94
	AP velocity (m/sec)	0.83	0.52 - 0.94
	Distance (mm)	0.83	0.52 - 0.94
	Area (mm^2)	0.69	0.25 - 0.90
Non-dominant	ML velocity (m/sec)	0.86	0.61 - 0.96
	AP velocity (m/sec)	0.88	0.66 - 0.96
	Distance (mm)	0.88	0.66 - 0.96
	Area (mm^2)	0.78	0.42 - 0.93

The ICC ranged from 0.69 to 0.88. The ICC ranged from 0.82 to 0.94.

6. Discussion

The objective of this study was to explore the possible association between one-leg static balance and frontal-plane projection knee angle, knee valgus angle and knee valgus moment during a VDJ in female players from the highest level of team handball and soccer in Norway.

We found no consistent association between one-leg static balance, measured as PS and frontal-plane projection knee angle, knee valgus angle and knee valgus moment during a VDJ.

6.1 Vertical drop-jump and postural sway

We discovered significant associations between FPPA and ML velocity, distance and area, in non-dominant leg for team handball players and between valgus angle and area for the non-dominant leg in team handball (tables 5.3, 5.4 and 5.5).

In the statistically significant associations, the knee joint frontal-plane kinematics were not sensitive to changes in the PS measures. The R^2 coefficients were very small and figure 5.1 display the impact in the knee joint frontal-plane kinematics caused by one SD change of the PS measures had on the frontal plane knee joint kinetics and kinematics. One SD change of ML velocity caused an 11.5% change of the independent variable (FPPA). The mean FPPA for the non-dominant leg in team handball players was 8.6° ±4.3°.One SD change in ML velocity caused less than one % change in FPPA. The correlation coefficients of the four statistically significant PS measures were small and have probably no clinically relevance.

There were in total only four statistically significant associations of all together 48 investigated associations between PS and frontal plane knee joint kinetics and kinematics. There are, to our knowledge, no other study investigating the relationship of PS and frontal plane knee joint kinetics and kinematics. However, accepting the frontal plane variables as a surrogate to ACL injury, these results may be evaluated in light of studies investigating the association of balance and injury. The lack of consistent associations between the one-leg static balance and in our study, stand in contrast to other studies that have shown an association in balance training and reduced injury incidence (Caraffa et al., 1996; Hewett et al., 1999; Holm et al., 2004; Myklebust et al., 2003). Although none of these studies have investigated the association of PS and frontal-plane knee joint kinetics and kinematics, we anticipated that the players who had good balance skills (little PS on one-leg static balance test) would display lower frontal-plane knee joint excursions- and forces. This was based on the rationale that one-leg static balance challenge the body orientation with respect to the gravity to prevent the player from falling, and to maintain body orientation/gravity equilibrium the player makes continuously small muscular contractions, measured as PS (Shumway-Cook & Woollacott, 2001). These muscle contractions are induced by feed forward and feedback mechanisms which combines the anticipation of the task at hand, and the proprioceptive afferent information (Riemann & Lephart, 2002). The anticipation of the task and the proprioceptive information also control the joint alignment, by muscle contractions that would increase the ability of the player to resist sudden joint displacements more effectively and thereby reduce destabilizing forces (Riemann & Lephart, 2002). The expected association of PS and frontal plane variablesare in line with the results from several studies identifying associations of PS, proprioception, muscle stiffness, and injury occurrence (Lloyd & Buchanan, 2001; Paterno et al., 2010; Soderman, Alfredson, Pietila, & Werner, 2001; Zazulak et al., 2007). Even the study of Søderman and colleagues (2001) that suggest that low PS is associated with increased risk of traumatic injury are not in any matter supported by our results. Our results suggest no association between PS and frontal plane knee joint kinetics and kinematics, which means that neither high nor low PS values are associated with increased frontal plane knee joint kinetics and kinematics.

In contrast to this study, most of the studies that have investigated an association between balance and PS have found such an association. This may be explained by that the data of knee joint kinetics and kinematics in this study are restricted to the frontal-plane. There could be that assessing the knee joint kinetics and kinematics in other planes than solely frontal would reveal other results as suggested by some authors (DeMorat et al., 2004; Sell et al., 2007; Speer et al., 1992).

However, there is more likely that the one-leg static balance test or VDJ did not challenge the properties that we wanted to investigate sufficiently and failed to identify the players at risk. The level of difficulty of the one-leg balance test as low compared to the challenges to PS team handball and soccer players faces in practice and competition daily. This could possibly make the players' results cluster at a very high level and unable us to identify players with poor balance skills. Such clustering are in some degree displayed in the scatter plots of the PS measures. This could represent a ceiling effect, similar to what Era and colleagues (2006) found in their comprehensive study where there was a significant ceiling effect for populations <60 years of age when testing normal stand, semi tandem stand and tandem stand.

ACL has a ultimate load at about ACL to 2160N ± 157N (Woo, Hollis, Adams, Lyon, & Takai, 1991), and that would normally require high impact situations for a ACL injury to occur. Olsen and colleagues (2004) found that most ACL injuries followed forceful jumping- and cutting situations. It could be that the one-leg static balance test only measures the player's skills in one-leg balance, while the players balance skills in a dynamic, high impact situation are different. Following this the players who get a high one-leg static balance score, are possibly not the same players that would get a high score in a dynamic, high-impact balance test.

A VDJ does not produce high frontal-plane joint excursions and moments compared to cutting (Kristianslund & Krosshaug, 2011), and was possibly unsuitable to identify the players at risk by frontal plane knee joint kinetics and kinematics. The assumption that frontal plane knee joint kinetics and kinematics in a VDJ would be able to identify players at risk was primarily based on the results of Hewett and colleagues (2005). They found that out of 205 athletes, the nine athletes that ruptured their ACL the following season displayed greater maximum knee abduction angle and greater peak knee abduction moment than the uninjured group in a VDJ, but to our knowledge, these results have not been reproduced in other studies.

Further, the frontal plane variables produced by the VDJ were low values with relative high SD (table 5.2). This could indicate that the VDJ do not challenge the players enough to discriminate the players at risk from the rest. The article in press from the same cohort as this study, by Kristianslund and Krosshaug (2011) found an average of 1.54Nm/kg ± 0.64Nm/kg, in valgus moments during cutting tasks, while the valgus moments from VDJ in this study ranged from 0.3Nm/kg to 0.4Nm/kg and the SD from ±0.1Nm/kg to ±0.2Nm/kg (table 5.2). This shows that the valgus moments in VDJ are significantly lower than during a cutting task. The low valgus moment values may expose the test to clustering of the players, preventing us to identify the players at risk.

There may have been non-linear associations that were not identified by a linear regression analysis. The scatter plots of the data reveal do not suggest such an association, but the FPPA curve (figure 5.3) display a substantial increase in SD from approximately 30% through to approximately 70% of the contact phase. It is evident that the variations of the players' angles were much higher in this phase. The same trend is evident in the valgus curve, however, not in the same magnitude (figure 5.4). The increase in SD followed just after the peak valgus moments (at approximately 25%-30% of the contact phase). This may suggest that some of the absorption of the GRF results in joint excursions in either valgus or varus direction when the players fail to absorb the GRF in by other mechanisms (Lloyd & Buchanan, 2001). A non-linear regression could have discovered such associations.

Further, the mean FPPA values in our study were, as opposed to what was expected, substantially lower than mean valgus angle values (table 5.2). FPPAs and valgus angles were both measured in degrees. We expected that FPPA would display the highest figures because this measure is a combination of multiple motions. While valgus angle was measured as pure knee joint valgus or varus, FPPA was a combined motion of valgus/varus, and the frontal projection of hip rotation and abduction combined with knee flexion. The maximum valgus directed FPPA in a two legged VDJ is restricted by the knees colliding in what is referred to as "knocking-knees", while the maximum varus directed FPPA is the resultant of knee adduction, hip external rotation and knee flexion as no such external mechanical restriction. The players with high varus directed FPPAs may "cancel out" the players with high valgus directed FPPAs and result in a low FPPA mean. The cancelling out may be the reason for the evident increase of SD around peak FPPA (approximately at 40% of the contact phase) (figure 5.2). Dividing the players into valgus and varus groups may display different results.

The lack of a consistent pattern in the associations of PS measures and frontal plane knee joint kinetics and kinematics, makes it is important to consider that the four significant associations were due to chance. The 95% confidence interval entails a five % risk of discovering associations that are false, usually referred to as type one error

(Vincent, 2005). This risk increase when multiple regression equations are conducted on the same dependent variable, as in this study. A common method modification to reduce the risk of this error is to do a Bonferroni adjustment (Vincent, 2005), however the Bonferroni adjustment lowers the confidence interval and thereby increase the risk of miss out actual significant associations (type two error). A Bonferroni adjustment increases the confidence interval to 98.75% and leaves only one statistically significant association, FPPA and ML velocity.

6.2 Limitations

This was a cross sectional study, and limited to investigate whether there were any association of one-leg static balance measured as PS and the frontal plane knee joint kinetics and kinematics in a VDJ.

Both one-leg balance test and the VDJ were to some degree prone to investigator errors. All the players were instructed individually preparing for each test, and these instructions may have varied. The tester was responsible for accepting the test, or to require a new trial. This judgment was to some extent a subjective assessment. The signal treatment and the tracking of the 3D marker trajectories were in some part carried out manually and mistakes may have occurred. However, the risk of investigator error was minimized in this study by several measures. The testers underwent pilot testing days where they were trained in their tasks. Further, the test station setup and the test procedures were identical for all the players and the testers were all the same on the test stations for this study. In marker placement, where knowledge of human anatomy and palpation skills were required, physiotherapists conducted the tasks.

In the previous chapter, we addressed the question of whether the tests sufficiently challenged the balance skills and frontal plane knee joint kinetics and kinematics. We were aware that the challenge of the one-leg static balance test was quite low. In spite this we expected that the test would be able to identify the players with poor balance because there were several players who needed two and three extra attempts to succeed at the task. Further, the tests are easily reproducible matter, with little instructions and restrictions for the player to relate to and the instructions were in a written protocol, providing the same information for each player. Additionally the one-leg static balance is quite similar to exercises well known to the players through injury prevention and rehabilitation programs and was a familiar task. This reduced the risk that players would not be able to perform at their best level. The dynamic balance task that were conducted in addition to the one-leg static balance test proved to be too dependent on the players perception of the task, and the technique used to solve the task, because the players were not familiar with the task in advance.

The VDJ is easily reproducible and familiar to most players as it is commonly used in jump training. This is important to reduce the influence of the test situation to the players' performance. The test was primarily chosen because it was used successfully to identify players at risk in Hewett and colleagues (2005) study. We were not able to reproduce the differences that that study did in terms of frontal plane knee joint kinetics and kinematics. The FPPAs and the valgus moments in our study were low, and this may reduce the tests ability to identify players at risk. The valgus angles were substantially higher, but it is still uncertain whether it would identify the players at risk. The frontal plane variables were low compared to the cutting task (Kristianslund & Krosshaug, 2011) and the association of knee joint kinetics and kinematics and risk of sustaining an ACL injury found by Hewett and colleagues (2005), were based on only 9 injuries, and to our knowledge these results have not been reproduced. Considering this, the frontal plane variables ability to identify players at risk is questionable.

6.2.1 Intra-rater reliability of one-leg static balance test

The intra-rater reliability investigations displayed no inclining or declining trend in any of the Bland-Altman plots (appendix 8). This indicates that the reliability is the same whether the players get a high or a low score. The lack of an inclining or declining trend also indicates that the order of the test had little impact on the players. The team handball and soccer players attended eight different tests stations during the test day. To manage the number of players that had to be tested each test-day, the order of tests had to vary for the players. Some players conducted the one-leg static balance test as their first test, and were rested at start, and some players were tested later in the day, after strength testing and VDJ, and may have been tired. This could presumably cause some differences in the results of the one-leg balance test. When testing the reliability, the players performed the one-leg static balance as their first and last test. If their performance were dependent on the order of tests, the performance of the first and second balance test should have revealed a declining or inclining trend. No such trend was evident in the Bland-Altman plots (appendix 8).

The assessment of the Bland-Altman plots display satisfactory results of ML velocity dominant leg and area for both non-dominant and dominant leg, in terms of percent within ± two SD of the mean between-test differences. All other PS measures have 92.3 % of the players within the range of two SD (table 5.7). The Bland-Altman plot requires 95 % of the results to fall within two SD of mean between-test difference. This should be interpreted with caution, considering that the study included only 13 players and may be vulnerable to one high or low value due to extreme players' skills, instrumental errors and other inaccuracies.

We further investigated the intra-rater reliability by ICC which values the variations between the players relative to the general variation, and gives nuanced and detailed information that Bland-Altman plots. All PS measures were to be considered moderate reliable (ICC > 0.8) except area in dominant and non-dominant leg, that were to be considered questionable (ICC < 0.8) (table 5.8) (Vincent, 2005).

The reliability of PS measures has also been investigated in several studies with different conclusions (Bauer et al., 2010; Bauer, Groger, Rupprecht, & Gassmann, 2008; Chockalingam, Giakas, & Iossifidou, 2002; Doyle, Hsiao-Wecksler, Ragan, & Rosengren, 2007; Santos, Delisle, Lariviere, Plamondon, & Imbeau, 2008) but our knowledge, there has been no study investigating the reliability of one-leg static balance, on a balance pad in athletes. However, several studies have assessed the reliability of different PS measures on similar balance platforms.

Chockalingam and colleagues (2002) investigated the dynamic accuracy when measuring COP using strain gauge force platform. They found that there was a minimum vertical force threshold at 113N to estimate COP within a SD of 0.3cm. They also found that the accuracy decreased from medial to lateral on the force platform, because of the platform stiffness, and shorter length in the lateral direction. This does probably not apply for the Good Balance platform used for this study, because it is triangular, and therefore the anterio-posterior and medio-lateral length are approximately equal.

Bauer and colleagues (2008) tested the intra-session reliability of PS in 63 healthy elderly standing measured by total length of displacement, area of sway, length of COP displacement in saggital and frontal-plane were all reliable. The ICC values varied form 0.71 for wide stance and eyes open, to 0.95 for narrow stance eyes closed. In a later study, intra-tester reliability proved to be high (>0.90) for mean speed of COP movement during narrow stand, area of sway during narrow stand, length of sway during normal and narrow stand (Bauer et al., 2010). A Norwegian study that investigated the intra-tester and inter-tester reliability on a newer model of the Good Balance Metitur system than the one we used in our study, found that the reliability for one-leg stance ranged from ICC 0.82-0.93 (Engebretsen, Mork, & Risberg, 2007).

None of the balance tasks in these studies are as challenging as ours, and all these studies reveal a trend that the more challenging the test are, the more reliable the results seems to be (Bauer et al., 2010; Bauer et al., 2008; Harringe, Halvorsen, Renstrom, & Werner, 2008). This is further supported by data revealing that two legged stance on foam, with eyes closed provides higher reliability than standing on firm surface eyes open or closed (Salavati et al., 2009). These studies indicate that the intra-rater reliability of the PS on balance platform range from moderate to strong, but none of these studies investigated the same testing procedure, or the same equipment as in our study. Consequently, the results from these studies should be interpreted with caution. However, the result from these studies coincides well with our reliability results and, to some extent, strengthens our reliability results.

The reliability could perhaps be further strengthened by increasing the duration of the tests and the number of trials. In our study, we required three valid tests of 20sec each, from where we used the mean score. To improve reliability Doyle and colleagues (2007) suggest, in a review, that at least five tests of at least 60 sec should be used to achieve a ICC coefficient >0.70, whereas Le Clair and Riach (1996) found that measures of 10-30 sec were reliable inter day. Further, a study with 12 participants measured wide stance that investigated number of trials needed to provide reliable figures, state that seven or more trials, or average results between days would improve reliability (Santos et al., 2008). This was not possible in this study because of the geographical disperse and time schedule of the whole test day

In light of other reliability studies on PS, the Bland-Altman plots, and the ICC coefficients, the intra-rater reliability using the Good Balance system, should be considered moderate, except in the case of area measures, which should be considered questionable.

6.2.2 3D motion analysis

3D motion analysis has been used in this study have been validated and used in several studies (Ford, Myer, & Hewett, 2007; Krosshaug et al., 2007; Krosshaug & Bahr, 2005; Krosshaug, Slauterbeck, Engebretsen, & Bahr, 2007). The method have proven to be the most reliable method in hand to evaluate such movements as in this study (Krosshaug et al., 2007), but there are some methodological sources of inaccuracy which should be paid attention to.

Instrumental errors

The cinamatographical system used in our study was very precise, and the source of this error is likely to be very small. The laboratory setup was according to the recommendations of Qtrac Capture & view reference manual (ProReflex, Qualisys INC Gothenburg, Sweden) (figure 4.9).The aim of the eight camera setup is to capture the each reflection marker with at least two cameras that are as close to 90° on an optical axis as possible. The distance to the player was approximately five meters and the calibration output indicates that the system recorded the reflective markers within two millimeter of actual position. All the markers were probably not captured by two cameras at a 90° angle at all times, and in some instances, lower camera angles have been used to estimate the marker position. However, this source of error is negligible compared to other sources of error in 3D motion analysis.

Marker placement

Marker placement was conducted by a physiotherapist who had sufficient knowledge of the human anatomy, that was trained in palpation and who filled in a detailed written form (appendix 4). There was one physiotherapist responsible for marker placement on all the team handball players and one physiotherapist responsible for marker placement on all the soccer players. Della Croce and colleagues (1999) found that the precision in palpation of anatomical landmarks varied with up to 21mm for intra-examiner and 25mm for inter-examiner, which causes significant differences in joint angle estimation and showed that the accuracy of identifying the placement for the marker, and the precise positioning of the marker is crucial to the precision of 3D motion. It is emphasized that the experience of the examiner is very important in securing precise marker placement (Besier, Sturnieks, Alderson, & Lloyd, 2003). All though the marker precision is a significant source of error, measures was taken to minimize the magnitude of these errors.

Soft tissue artifact

During the recording of 3D motion, some of the players VDJs were recorded using high-speed camera (Casio Exilin Ex-F1), and we discovered ca 40mm displacement of the left knee marker (table 4.2) during the task. This displacement was due to soft tissue artifact (STA). The STA is movement of the skin, or soft tissue in relation to the underlying bone and cause discrepancies between actual skeletal movement, and movement recorded according to the markers attached to the skin. It has been suggested that the skin marker move up to 20mm in relation to the underlying bone (Fuller, Liu, Murphy, & Mann, 1997). This is a factor that challenges the reliability of 3D measures, and that the results from tests using skin markers should be interpreted with caution (Leardini, Chiari, Della, & Cappozzo, 2005)

The alternative to skin markers is bone pins, which are markers pinned to the bone through the skin. Reinschmidt and colleagues (1997) found that skin markers on tibia and femur provides reliable information on flexion and extension in the knee when compared to bone pins in walking. However abduction/adduction and internal/external rotation show a great difference compared to the amplitude of the movement while running and should be interpreted with caution.

Stagni and colleagues (2005), however, found that bone pins themselves can reduce skin movement in a study investigating this combining fluoroscopy and 3D motion analysis. This could lead to underestimations of STA, mostly for ab-/adduction and internal/external rotations. Bone pins are much more invasive than skin markers, and the use of such pins have more complications that skin markers. The placement of the bone pins are much more extensive and require more training than placement of the skin markers. Further, the bone pins are more expensive, somewhat painful and will possibly affect the player performance in the VDJ. With this in mind, the bone pins are of little advantage compared to the invasiveness to the player, the cost and time consume, and was not suitable to our study.

It is important to consider the potential error of STA when interpreting the results in this study. STA could represent great sources of error in the method, but when these artifacts are recognized, skin markers is the most accurate method available considering the time the participants have to spend, the magnitude of the interference that a pre-season player may allow, and the cost for the project.

Joint center estimations

When looking at the markers trajectories in some players during a VDJ from our study, we could see the markers defining the knee joint were visibly skidding at a time during knee flexion. This is probably due to inaccuracy of joint center estimations. Studies have established that in the method used in our study the knee and ankle joint center differs little from true position (Davis, Ounpuu, Tyburski, & Gage, 1991, Eng & Winter, 1995). The hip joint center estimations are a bit more rough, and range within 1.07 cm of true location (Bell, Pedersen, & Brand, 1990).

Joint center accuracy is dependent on joint center estimates and crucial to provide reliable information to the calculation of knee joint kinetics and kinematics. The knee joint estimations are more accurate estimations than the hip joint estimation, but it would still reduce the accuracy of the knee joint kinetics and kinematics to a modest extent.

In-session reliability

In our study, the players performed at least five VDJs and the three latter that were approved were used for analysis. This gives a reliable representation of the players maximum drop jump and reduces the risk of sub optimal performance. Ford and colleagues (2007) investigated the session reliability in 3D analysis of the lower extremities in adolescents performing a drop jump. They investigated discrete joint variables such as maximum joint moment and maximum and minimum joint angle. Regarding knee movement, their study found excellent reliability within session for all planes and moments, while between session results showed fair to excellent reliability varying on which parameters that were chosen.

The 3D motion analysis used for our study contains numerous potential sources of error that all should be considered when interpreting the result of this study. The instrumental errors are small and may be neglected in this context. Marker placements are of more concern, but the written procedures and the training of the investigators reduce the risk of this source of error significantly. The joint center estimation of the knee is small and should not challenge the reliability of the results significantly. The greatest source of error is the STA, and this reduces the precision of the estimation of the segments considerably.

Nevertheless, 3D motion analysis is a great instrument for analyzing joint kinetics and kinematics, and probably the best method available for collecting data in a study like ours (Krosshaug et al., 2007).

7. Conclusions

The results of this study did not reveal any association between PS in one-leg static balance and FPPA, valgus angle and valgus moments in VDJ. However, considering the limitations in this study a possible association between PS and knee joint kinetic and kinematics should not be dismissed.

7.1 Clinical implications and future research directions

Several studies support the inclusion of balance training in preventive training because of its potential to reduce sport injuries (Caraffa et al., 1996; Hewett et al., 1999; Holm et al., 2004; Hrysomallis, 2011; Myklebust et al., 2003) and we recommend that the present advice for clinical practice, of including balance training in preventive training should not be changed on behalf of this study.

We think that there is need of more research on the association of balance skills and injury risk. In the ongoing cohort the players that were tested later than spring 2009 performed a one-leg landing on the balance platform in addition to the one-leg static stance. This is a more challenging test that may avert the ceiling effect, and identify the players with poor balance skills. This test may reveal an association between PS and frontal plane knee joint kinetics and kinematics. Further, one should look into the potential association between PS tasks and frontal plane variables during cutting tasks as cutting tasks may be more adequate to identify players at risk of sustaining an ACL injury.

It is reasonable that frontal-plane knee joint excursions may be associated with increased injury risk and it is conceivable that frontal-plane knee joint excursions may associate with a balance task performance. Establishing such an association would have important benefits. It could potentially increase the efficiency of preventive training by targeting the important exercises more precisely, and it could make us able to identify athletes at risk without requiring expensive, complicated and time consuming methods like 3D motion analysis. A simple balance test would become an important instrument to coaches and medical staff to identify athletes at risk, so that preventive measures could be taken. These potential benefits justify further research on the issue.

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Abbreviations

ACL	Anterior cruciate ligament
AP	Anterioposterior
BMI	Body mass index
COM	Center of mass
COP	Center of pressure
FPPA	Frontal plane projection angle
GCS	Global coordinate system
GRF	Ground reaction force
IC	Initial contact
ICC	Intraclass correlation coefficient
JCS	Joint coordinate system
ML	Mediolateral
OA	Osteoarthritis
PS	Postural sway
SD	Standard deviation
STA	Soft tissue artifact
VDJ	Vertical drop jump

Appendices

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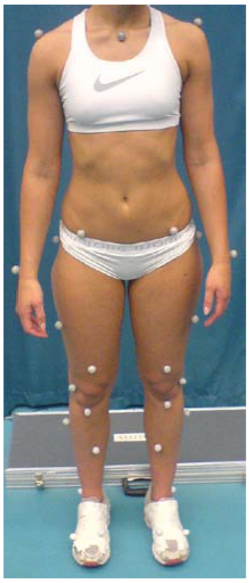
Information to the players

SENTER FOR
Idrettsskadeforskning
KLOKE AV SKADE

Forskningsprosjekt blant fotballspillere i Toppserien

Senter for idrettsskadeforskning gjennomfører nå et forskningsprosjekt der vi undersøker hvem som er utsatt for å få korsbåndskader. Vi testet hele eliteserien i håndball for kvinner i løpet av juni 2007, og tilsvarende testing skal nå gjennomføres blant kvinnelige fotballspillere i Toppserien i februar og mars 2009. Spillerne vil deretter følges i 4 år, hvor vi registrerer korsbåndskader som oppstår i disse årene.

Vi har satt av tid til testing av **Kolbotn tirsdag 24.februar og tirsdag 3.mars**. De som har sagt ja til å delta i prosjektet vil bli testet ved Norges idrettshøgskole (NIH). Spillerne møter opp ved resepsjonen på NIH, og vi vil først ha et kort informasjonsmøte hvor dere får mer informasjon om prosjektet. Etter dette ber vi dere om å skrive under på en erklæring på at dere samtykker i å delta i prosjektet.



Vi har totalt 7 teststasjoner hvor dere skal gjennomføre tester av blant annet styrke, balanse og bevegelighet, samt en bevegelsesanalyse. Dere bruker omtrent en time på hver stasjon, og hele testingen vil derfor ta omtrent 7 timer. Dere vil selvfølgelig få mat og drikke underveis.

Dere har på dere treningstøy og hallsko under testingen. For å gjøre testingen lettere bør dere bruke en shorts eller tights som ikke går nedenfor knærne. På overkroppen bruker dere sports-BH og en stram topp. Noen av testene gjennomføres i undertøyet, så ta gjerne på en boksershorts eller bikinitruse til disse testene (se bilde). Markørene vi bruker til bevegelsesanalysen festes med teip, så **unngå å bruke bodylotion** på testdagen.

For å se bilder fra testingen, kan dere finne dette på hjemmesiden til Senter for idrettskadeforskning; www.klokeavskade.no og søke på korsbåndsstudie.
Eller benytte linken;
http://www.klokeavskade.no/no/Nyhetsarkiv/Nyhetsarkiv-2007/Ny-studie-i-kvinnenes-eliteserie-i-handball/

Ta gjerne kontakt på e-post (agnethe.nilstad@nih.no) eller telefon (99 22 44 69) dersom dere har spørsmål.

Vennlig hilsen

Agnethe Nilstad
Senter for idrettsskadeforskning

Letter of invitation

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

Toppserieklubb
v/Sportslig leder

Deres ref.	Vår avd.	Vår ref.	Vår dato:
	Senter for idrettsskadeforskning	KS www.klokavskade.no	12.12.2008

Nytt prosjekt: Undersøkelse av risikofaktorer for korsbåndskader i Toppserien

Senter for idrettsskadeforskning ved Norges idrettshøgskole planlegger sesongen 2009 et stort prosjekt i Toppserien der formålet er å få detaljert kunnskap om hva som forårsaker de mange korsbåndsskadene som oppstår blant kvinnelige fotballspillere.

Til tross for god kunnskap om skadeforebygging, er det fremdeles langt igjen til at vi kan forebygge disse skadene så effektivt som ønskelig, til beste for den enkelte spiller og til beste for norsk fotball.

Det første delmålet på veien til effektiv forebygging av disse alvorlige skadene er å forstå hva som gjør at enkelte spillere lettere blir skadet enn andre. Derfor vil Senter for idrettsskadeforskning støttet av Norges fotballforbund starte et nytt prosjekt som involverer alle spillerne i Toppserien.

Prosjektet

I prosjektet vil alle spillerne gjennomgå en grundig kartlegging og testing av mulige risikofaktorer for korsbåndskade før sesongstart. Deretter vil alle nye korsbåndsskader bli registrert gjennom de fire påfølgende sesongene. Resultatene vil forhåpentligvis fortelle oss hva som karakteriserer spillere som får korsbåndskader. På bakgrunn av denne kunnskapen kan vi utvikle mer målrettede tiltak for å forebygge korsbåndskader hos spillere med størst risiko.

I praksis innebærer dette at alle lagene i Toppserien vil bli invitert til en testdag på Norges idrettshøgskole i Oslo. Testperioden er valgt fra slutten av januar og ut februar 2009. Testene vil måle styrke, spenst, bevegelighet og andre faktorer som kan påvirke risikoen for å pådra seg en korsbåndskade. Det vil også bli gjennomført en tredimensjonal videoanalyse av fotballspesifikke bevegelser til spillerne i forbindelse med en finte eller vending. Prosjektet vil gi lagene kunnskap om spillernes fysiske prestasjonsevne som vil være verdifull for trenerne i evalueringen og planlegging av treningsarbeidet. I 2009 vil vi i tillegg registrere alle skadene som oppstår gjennom sesongen.

Senter for idrettsskadeforskning vil dekke alle reise-, bo- og matutgifter for spillerne i forbindelse med testdagen.

Hvem er vi?

Senter for idrettsskadeforskning er en forskningsgruppe bestående av leger, fysioterapeuter, og idrettsforskere, og er lokalisert ved Norges idrettshøgskole i Oslo. Senter for idrettsskadeforskning har allerede gjennomført flere vellykkede prosjekter i samarbeid med Norges fotballforbund og Norges håndballforbund. Senteret og prosjektet er finansiert gjennom midler fra Helse Sør Øst, Kulturdepartementet, Norges Idrettsforbund og Olympiske Komité og Norsk Tipping AS og FIFA.

Deres klubb vil bli kontaktet etter nyåret for å avklare om dere ønsker å delta og avtale tidspunkt for gjennomføring av testene. Ring gjerne til Kathrin Steffen (99 00 43 98) hvis du allerede har spørsmål om prosjektet.

Vi ser frem til et godt samarbeid!

Vennlig hilsen

Thor Einar Andersen (Leder medisinsk komité, NFF, forsker, Senter for idrettsskadeforskning)
Kathrin Steffen (Prosjektleder og forsker, Senter for idrettsskadeforsakning)
Roald Bahr (Professor og leder, Senter for idrettsskadeforskning)

Norges idrettshøgskole
Norwegian School
of Sport Sciences

Postadresse:
P. O. BOX 4014 – Ullevål Stadion
NO-0806 OSLO

Besøksadresse:
Sognsv. 220
0863 OSLO

Tel.: (+47) 23 26 20 00
Org.nr.: 971526033
www.nih.no

Matlab script for calculating measures of postural sway

```
% Oslo Sports Trauma Research Centre
% www.ostrc.no
%
% Authors: Aleksander Killingmo and Trond Krosshaug, 2011
% contact information: alxkill@gmail.com
%
% This script converts postural sway data retrieved from
% Good Balance Metitur balance platform.
%
% This script converts .txt files with X, Y coordinates organized
% vertically for each time frame. Adjustments to number of frames
% and directory of the txt files have to be made.
% The text files should be named according to number of the team,
% number of the player on the team, type of test, and number of
% the test. e.g.: team1_player1_staticl1

% This script requires a teamnumber_sport m file which reads in
% the number of the athletes which data to be calculated.

% This script calculates means of 3 trials for medio lateral
% (mean_xvel) and anterio posterior sway (mean_yvel),total
% distance covered (mean_xydist)and 95% confidence ellipse of
% the area covered (mean_COP_area).
% All data are written to one txt file, accompanied by a syntax
% file for SPSS.

%fc = Cutoff frequency
%fs = Framesize (Hz)
fc= 6;
fs= 50;

%Function that reads which players form which teams to be loaded.
teamnumber_soccer
iPlayer = 1;
% Loads 6 text files for each player
for i = teamNumber(1,:)
    for j = 1:teamNumber(2,i)
%  % k = team code that starts with team100. The perviously used 'i'
variable
%  % starts at 1 and is used as playerID as well.
        k = i+300;
        eval(['jj=team' num2str(k) '(j)']);
        COPl1_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticl1.dat']);
        COPl2_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticl2.dat']);
        COPl3_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticl3.dat']);
        COPr1_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticr1.dat']);
        COPr2_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticr2.dat']);
```

```
        COPr3_unfiltered=load(['data_GoodBalance_soccer\team'
num2str(k) '\static\team' num2str(k) '_player' num2str(jj)
'_staticr3.dat']);

% Applies a Butterworth filter
        COPl1 = zeros(size(COPl1_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
             COPl1(:,m)= filtfilt(b,a,COPl1_unfiltered(:,m));
        end
        clear b a

        COPl2 = zeros(size(COPl2_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
            COPl2(:,m)= filtfilt(b,a,COPl2_unfiltered(:,m));
        end
        clear b a

        COPl3 = zeros(size(COPl3_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
            COPl3(:,m)= filtfilt(b,a,COPl3_unfiltered(:,m));
        end
        clear b a

        COPr1 = zeros(size(COPr1_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
            COPr1(:,m)= filtfilt(b,a,COPr1_unfiltered(:,m));
        end
        clear b a

        COPr2 = zeros(size(COPr2_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
            COPr2(:,m)= filtfilt(b,a,COPr2_unfiltered(:,m));
        end
        clear b a

        COPr3 = zeros(size(COPr3_unfiltered));
        [b,a]=butter(2,2*(fc/fs));
        for m=1:2;
            COPr3(:,m)= filtfilt(b,a,COPr3_unfiltered(:,m));
        end
        clear b a

        savefilename = ['save\soccer\team' num2str(i) '_player'...
            num2str(jj)];


% Function analysis calculates means of 3 trials for x and y velocity
(mean_xvel,
% mean_yvel), total distance covered (mean_xydist) and area 95%
confidence
% ellipse of the area covered (mean_COP_area).


[mean_xvel_r,std_xvel_r,mean_xvel_l,std_xvel_l,mean_xydist_total_r,...
```

```
        std_xydist_total_r,mean_xydist_total_l,std_xydist_total_l,...

mean_yvel_r,std_yvel_r,mean_yvel_l,std_yvel_l,mean_COP_area_r,...
        std_mean_COP_area_r,mean_COP_area_l,std_mean_COP_area_l] =...
        analysis(COPl1,COPl2,COPl3,COPr1,COPr2,COPr3, savefilename);

        playerID = 20000+i*100+jj;

        output(iPlayer,1:17) =
[playerID,mean_xvel_r,std_xvel_r,mean_xvel_l,std_xvel_l,mean_xydist_to
tal_r,...
        std_xydist_total_r,mean_xydist_total_l,std_xydist_total_l,...

mean_yvel_r,std_yvel_r,mean_yvel_l,std_yvel_l,mean_COP_area_r,...
        std_mean_COP_area_r,mean_COP_area_l,std_mean_COP_area_l];
        iPlayer = iPlayer+1;

    end
end

% Names the variables exported to SPSS syntax file
varnames = ...

char('subjectID','mean_xvel_r','std_xvel_r','mean_xvel_l','std_xvel_l'
,'mean_xydist_total_r',...

'std_xydist_total_r','mean_xydist_total_l','std_xydist_total_l',...

'mean_yvel_r','std_yvel_r','mean_yvel_l','std_yvel_l','mean_COP_area_r
',...
    'std_mean_COP_area_r','mean_COP_area_l','std_mean_COP_area_l');

varnames = cellstr(varnames);
spssfilename = ['Results' date '.txt'];

% Exports selected parameters to SPSS
save4spss(output,varnames,spssfilename);
```

Yeadon body segment parameter form

Fornavn:	Etternavn:	Personnr:	Dato:	Lag:

Torso	p	w	h		
Ls0 hip joint centre				P=perimeter	
Ls1 umbilicus				w=width	
Ls2 lowest front rib				h=height	
Under BH (Ls2b)					
Ls3 nipple				Distance shoulder joint centre	Depth shoulder
Ls4 shoulder joint centre					
Ls5 acromion/neck					
Ls6 beneath nose				Depth above ear	
Ls7 above ear					
Ls8 top of head					

Left arm	p	w	h
La0 shoulder joint centre			
La1 mid-arm			
La2 elbow joint centre			
La3 maximum forearm perimeter			
La4 wrist joint centre			
La5 length hand			

Length finger - finger
ASIS distance
Height
ASIS - umbilicus
Umbilicus - proc. xyphoid
Proc. xyphoid. - C7
C7 - top of head

Right arm	p	w	h
Lb0 shoulder joint centre			
Lb1 mid-arm			
Lb2 elbow joint centre			
Lb3 maximum forearm perimeter			
Lb4 wrist joint centre			
La5 length hand			

Left Leg	p	w	h	Anisomali (YES / NO)	Side higher
Lj0 hip joint centre					
Lj1 crotch					
Lj2 mid-thigh					
Lj3 knee joint centre					
Lj4 maximum calf perimeter					
Lj5 minimum calf perimeter					
Lj6 ankle joint centre					
Lj7 ankle-floor height					
Lj8 length foot					

Right Leg	p	w	h
Lk0 hip joint centre			
Lk1 crotch			

Lk2 mid-thigh			
Lk3 knee joint centre			
Lk4 maximum calf perimeter			
Lk5 minimum calf perimeter			
Lk6 ankel joint centre			
Lj7 ankle-floor height			
Lj8 length foot			

	Right	Left
Width femur condyles		
Width tibia condyles		
Length foot		
Width foot		
Width ankle		
Width elbow		
Width hand		
Height floor-hip joint centre		
Height floor-shoulder		

Notes

Approval from Regional Committee for Medical Research Ethics approval

UNIVERSITETET I OSLO

DET MEDISINSKE FAKULTET

Forsker dr.scient. Tron Krosshaug
Norges idrettshøgskole
Pb. 4014 Ullevål Stadion
0806 Oslo

Regional komité for medisinsk forskningsetikk
Sør- Norge (REK Sør)
Postboks 1130 Blindern
NO-0318 Oslo
Telefon: 228 44 666
Telefaks: 228 44 661
E-post: rek-2@medisin.uio.no
Nettadresse: www.etikkom.no

Dato: 10.4.07
Deres ref.:
Vår ref.: S-07078a

S-07078a Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndballspillere - en prospektiv kohortstudie [2.2007.511]

Vi viser til brev datert 19.3.07 revidert informasjonsskriv med samtykkeerklæring og kopi av brev til klubbene.

Komiteen tar svar på merknader til etterretning.

Komiteen har ingen merknader til revidert informasjonsskriv med samtykkeerklæring.

Komiteen tilrår at prosjektet gjennomføres.

Vi ønsker lykke til med prosjektet.

Med vennlig hilsen
Kristian Hagestad
Kristian Hagestad
Fylkeslege cand.med., spes. i samf.med
Leder

Jørgen Hardang
Jørgen Hardang
Sekretær

Approval from Norwegian Social Science Data Services

Norsk samfunnsvitenskapelig datatjeneste AS
NORWEGIAN SOCIAL SCIENCE DATA SERVICES

NSD

Harald Hårfagres gate 29
N-5007 Bergen
Norway
Tel: +47-55 58 21 17
Fax: +47-55 58 96 50
nsd@nsd.uib.no
www.nsd.uib.no
Org.nr. 985 321 884

Tron Krosshaug
Senter for idrettsskadeforskning
Norges Idrettshøgskole
Postboks 4014 Ullevål Stadion
0806 OSLO

Vår dato: 03.05.2007 Vår ref: 16639/KS Deres dato: Deres ref:

TILRÅDING AV BEHANDLING AV PERSONOPPLYSNINGER

Vi viser til melding om behandling av personopplysninger, mottatt 29.03.2007. Meldingen gjelder prosjektet:

16639	*Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndballspillere – en prospektiv kohortstudie*
Behandlingsansvarlig	*Norges idrettshøgskole, ved institusjonens øverste leder*
Daglig ansvarlig	*Tron Krosshaug*
Student	*Eirik Kristianslund*

Personvernombudet har vurdert prosjektet, og finner at behandlingen av personopplysninger vil være regulert av § 7-27 i personopplysningsforskriften. Personvernombudet tilrår at prosjektet gjennomføres.

Personvernombudets tilråding forutsetter at prosjektet gjennomføres i tråd med opplysningene gitt i meldeskjemaet, korrespondanse med ombudet, eventuelle kommentarer samt personopplysningsloven/-helseregisterloven med forskrifter. Behandlingen av personopplysninger kan settes i gang.

Det gjøres oppmerksom på at det skal gis ny melding dersom behandlingen endres i forhold til de opplysninger som ligger til grunn for personvernombudets vurdering. Endringsmeldinger gis via et eget skjema, http://www.nsd.uib.no/personvern/endringsskjema. Det skal også gis melding etter tre år dersom prosjektet fortsatt pågår. Meldinger skal skje skriftlig til ombudet.

Personvernombudet har lagt ut opplysninger om prosjektet i en offentlig database, http://www.nsd.uib.no/personvern/register/

Personvernombudet vil ved prosjektets avslutning, 01.06.2017, rette en henvendelse angående status for behandlingen av personopplysninger.

Vennlig hilsen

Bjørn Henrichsen Katrine Utaaker Segadal

Kontaktperson: Katrine Utaaker Segadal tlf: 55 58 35 42

Vedlegg: Prosjektvurdering

Kopi: Eirik Kristianslund, Nedre Ullevål 9 - H0407, 0850 OSLO

Avdelingskontorer / District Offices:

Informed consent

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

FORESPØRSEL OM DELTAKELSE I PROSJEKTET: *"Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndball og -fotballspillere - en prospektiv kohortstudie"*

Bakgrunn for undersøkelsen

Korsbåndsskader i fotball og håndball har i det siste vært et svært aktuelt tema, både i media og i forskningssammenheng. Dette skyldes først og fremst den relativt store hyppigheten av denne alvorlige skaden, spesielt blant kvinnelige utøvere, som ser ut til å skade seg 3-7 ganger hyppigere enn menn. Problemet så langt er imidlertid at vi vet for lite om risikofaktorene og skademekanismene for korsbåndskader. Denne informasjonen er viktig når vi forsøker å forebygge skader, både for å kunne vite hvem som vil ha størst glede av forebyggende trening og for å kunne utvikle mest mulig effektive treningsmetoder.

Senter for idrettsskadeforskning er en forskningsgruppe bestående av fysioterapeuter, kirurger og biomekanikere med kunnskap innen idrettsmedisin. Vår hovedmålsetting er å forebygge skader i norsk idrett, med spesiell satsning på fotball, håndball, ski og snowboard. Denne studien er en viktig brikke i arbeidet med å finne ut hvorfor noen får en korsbåndskade. Vi ønsker nå å undersøke ulike mulige risikofaktorer for korsbåndskader, for deretter å kartlegge hvem som får korsbåndskader de påfølgende sesongene.

Gjennomføring av undersøkelsen

Vi ønsker at du som elitespiller deltar i denne studien, og deltakelsen er frivillig. Testingen vil finne sted på Norges idrettshøgskole. I løpet av en dag vil vi gjennomføre ulike styrke-, balanse- og bevegelighetstester, anatomiske målinger, samt gjennomføre en bevegelsesanalyse av hvordan du finter, vender, hopper og lander. Undersøkelsen starter med en kort oppvarming, deretter får du festet små refleksmarkører på kroppen (33 stk totalt). Du vil så bli bedt om å gjennomføre tre finter/vendinger og tre fallhopp. Under disse øvelsene vil det være 8 infrarøde kamera som filmer markørene, samtidig som kreftene fra underlaget blir målt. Dataene fra markører, kraftplattform og anatomiske mål benyttes i en matematisk modell som gir ut leddkrefter og momenter. Disse kreftene/momentene gir oss informasjon om hvordan muskler og passive strukturer som leddbånd belastes.

Bevegelsesanalysen vil ta ca. 1.5 time, inkludert anatomiske målinger og påsetting av markører. De andre testene gjennomføres resten av tiden laget er på NIH, og totalt vil testene ta om lag åtte timer. I tillegg til disse testene vil du få utdelt et skjema, der vi spør om treningserfaring, tidligere skader, skade i familien, treningsmengde, menstruasjonsstatus og knefunksjon. Spørreskjemaet besvares i løpet av testdagen, og det vil ta ca. 30 min.

Behandling av testresultatene

Vi vil de neste tre sesongene følge opp alle lag og spillere som har deltatt på testing hos oss for å registrere alle korsbåndskader som oppstår.

Vi er også interessert i å kunne kontakte deg senere med tanke på oppfølgningsstudier. Dette kan f.eks. skje ved at du får tilsendt et spørreskjema. Av den grunn vil vi lagre resultatene fra testene og svarene på spørreskjemaet fram til 1.6.2017. Etter dette vil dataene bli anonymisert. Dataene vil bli behandlet konfidensielt, og kun i forskningsøyemed. Alle som utfører testingen og forskere som benytter dataene er underlagt taushetsplikt. Dersom du ikke ønsker å være med på etterundersøkelser, kan du reservere deg mot dette i samtykkeerklæringen. I så fall vil alle dine data bli anonymisert etter fire år.

Vi vil underveis i testingen ta videoopptak av dere som vi senere kan ønske å bruke i undervisnings- og formidlingssammenheng. Opptakene inkluderer situasjoner der dere kun har på shorts og sports-BH. Dersom dere ikke vil at deres opptak skal være aktuelle for slik bruk krysser dere av for det i samtykkeerklæringen.

Hva får du ut av det?

Vi kan ikke tilby noe honorar for oppmøtet, men vil dekke eventuelle reise- og matutgifter. I tillegg vil du få kopi av dine resultater fra styrketestene som gjennomføres i løpet av testdagen.

Angrer du?

Du kan selvfølgelig trekke deg fra forsøket når som helst uten å måtte oppgi noen grunn. Alle data som angår deg vil uansett bli anonymisert.

Spørsmål?

Ring gjerne til Tron Krosshaug, tlf.: 45 66 00 46 hvis du har spørsmål om prosjektet, eller send e-post til tron.krosshaug@nih.no.

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

"Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndball og -fotballspillere - en prospektiv kohortstudie"

SAMTYKKEERKLÆRING

Jeg har mottatt skriftlig og muntlig informasjon om studien *Risikofaktorer for fremre korsbåndskader hos kvinnelige elitefotballspillere - en prospektiv kohortstudie*. Jeg er klar over at jeg kan trekke meg fra undersøkelsen på et hvilket som helst tidspunkt.

- ☐ Jeg ønsker ikke å bli kontaktet etter endt karriere med tanke på oppfølgningsstudier
- ☐ Jeg ønsker ikke at video av meg skal brukes i undervisningssammenheng

Sted ……………………….. Dato ………………………..

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Underskrift

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Navn med blokkbokstaver

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Adresse

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Mobiltelefon

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E-postadresse

Bland- Altman plot for intra-rater reliability