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# Kinetic and kinematic analysis of the lower extremities during a vertical drop-jump with and without an overhead target

Implications for ACL injury screening

**Master thesis in Sport Sciences**

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Norwegian School of Sport Sciences, 2013

# Abstract

**Introduction:** Former studies have indicated that knee valgus angles and -moments in a Vertical Drop-Jump (VDJ) can predict future Anterior Cruciate Ligament (ACL) injuries. The use of an overhead target has proved to improve effort on subjects tested. This may in turn result in a test, which challenges the dynamic knee control to a greater extent. The aims of the study were to investigate if:

- 1) Elite female handball and football and handball athletes improve jump-height in overhead-target Vertical Drop Jumps compared to non-target Vertical Drop Jumps.
- 2) An overhead-target induces changes in lower extremity frontal plane and sagittal plane biomechanics, in elite female handball and football athletes?

**Method:** Thirty-six elite handball- ( $172.3 \pm 6.0$  cm,  $69.9 \pm 7.3$  kg) and thirty-five football players ( $164.8 \pm 5.7$  cm,  $62.2 \pm 6.7$  kg) conducted three two-legged VDJ without and with an overhead target in a 3D motion analysis lab. The athletes was equipped with 35 reflex markers, and filmed with an 8-camera movement analysis system (ProReflex, Qualisys). Two force platforms measured forces from the ground. Kinetic and kinematic data was calculated using standard inverse dynamics.

**Results:** The athletes exhibited a significant improvement in performance, i.e. jump height progressed (41.2 cm vs. 43.8 cm,  $p < 0.01$ ). A Bland-Altman plot displayed no relation between jump height and change in jump height. The athletes exhibited a significant change in maximum knee valgus angle (Right:  $9.9^\circ$  vs.  $10.4^\circ$ ,  $p < 0.01$ . Left:  $10.8^\circ$  vs.  $11.3^\circ$ ,  $p < 0.05$ ). The knee valgus moments also changed significantly (Right: 0.41 vs. 0.45 Nm/kg,  $p < 0.01$ . Left: 0.37 vs. 0.40 Nm/kg,  $p < 0.05$ ). The athletes displayed less flexion angles in overhead target situations.

**Discussion:** The overhead target will likely have a positive effect on jump performance, as the athletes improved jump-height significantly (2.6 cm). They also utilize a quicker and more erect jump-technique, due to the changes in hip ( $4.2^\circ$ ) and knee ( $3.1^\circ$ ) flexion angles. Despite exhibiting statistically significant changes in valgus angle ( $0.5^\circ$ ) and –moments (0.04 Nm/kg), the changes were likely clinical insignificant.

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## **2 Introduction**

### ***2.1 Occurrence of ACL injuries in team sports***

In female team sports one of the most common injury sites is the knee, and Anterior Cruciate Ligament (ACL) rupture is perhaps the most devastating knee injury. ACL injuries are often divided into contact and non-contact injuries. The contact ACL injuries are defined as a direct blow to the knee. The non-contact injuries are divided into situations with no contact or indirect contact i.e. defined as a push, pull or any other perturbation to another body part than the knee. Such a perturbation can e.g. cause awkward foot positioning or unfavorable lower extremity alignment i.e. a movement that the athlete cannot control (Olsen, Myklebust, Engebretsen, & Bahr, 2004). This type of ACL injury is often seen in sports that include high force jumping and quick changes of direction (Besier, Lloyd, & Ackland, 2003). Team sports like basketball, volleyball, handball and football are in this category, and subjects in ACL-studies are often selected from these sports. Up to 90% of ACL injuries have been found to occur in non-contact situations, during deceleration, landing or during sidestep cutting which likely is due to high external forces on the knee joint (Myklebust, Mæhlum, Holm, & Bahr, 1998), however, it should be noted that later studies have found the share to be about 70% (Boden, Dean, Feagin, & Garrett, 2000; Besier et al., 2003).

In the last 40 years the female participation in high-school sports in USA has increased more than a 9-fold (Hewett, et al., 2005). In Norway the female participation in organized handball and football has increased between 10-25% over the last nine years (Norges Idrettsforbund, Årsrapport 2011, 2012; Norges Idrettsforbund, Årsrapport 2002, 2003). This however is a perplex situation as studies have shown a three to five times greater ACL injury rate in female athletes in team sports, compared to their male counterparts (Myklebust, Mæhlum, Engebretsen, Strand, & Solheim, 1997; Myklebust et al., 1998). Anatomical, hormonal and neuromuscular factors are suggested to contribute to these gender differences (Hewett, et al., 2005).

## **2.2 Consequences**

An ACL injury is a devastating injury with serious consequences for the player. It is known that, as much as 5-10% of the players at elite level in handball is exposed to an ACL injury every season (Myklebust et al., 1997, 1998). If an athlete wishes to return to the sport, surgery is recommended, and initiated within 4-8 weeks to reestablish mechanical stability in the knee (Myklebust & Bahr, 2005).

ACL injuries require long-term rehabilitation to restore knee joint function, but earlier studies have reported that 58-68% of the injured athletes in handball and football return to the same level of participation at some point (Bak, Jørgensen, Ekstrand, & Scavenius, 2001; Myklebust, Holm, Mæhlum, Engebretsen, & Bahr, 2003a).

Nevertheless another study has shown that only 30% of elite football players were still active three years after their injury, and after seven years none were participating at elite level (Roos H. , Ornell, Gärdsell, Lohmander, & Lindstrand, 1995). For a high-school player, an ACL injury will not only ruin the season or career, and it can also mean withdrawal of scholarship, and thus serious educational consequences.

The greater participation in female sports has in turn had an impact on surgery and rehabilitation costs, and in the United States \$119 million is spent annually on rehabilitating female high-school basketball players (Hewett, Lindenfeld, Riccobene, & Noyes, 1999). So both on an individual and a society scale, it follows that efforts to reduce and prevent ACL injury incidence are paramount.

## **2.3 Prevention work**

Identifying and preventing risk factors for ACL injuries has therefore been a focus in research for the last ten years. Thorough analysis of athletes' movement pattern and specifically the relation between movements and ACL injury has been discussed (Myklebust et al., 1998; Boden et al., 2000; Bahr & Krosshaug, 2005). It has been shown that non-contact injuries often occur when the knee joint is loaded with high external forces such as during jump-landing, deceleration and/or sudden direction changes (Boden et al., 2000; Besier et al., 2003). It has further become clear that external forces combined with valgus motion and internal rotation of the knee is one of the main causes of ACL injuries (Koga, et al., 2010). Thus coaches, organizations

and researchers have been trying to develop screening tools to identify player exhibiting these movement patterns associated with ACL-injuries, and training schemes and exercises that can reduce the risk of sustaining ACL injury.

## **2.4 The Vertical Drop-Jump**

As athletes strive to fulfill the Olympic motto “Citius, Altius, Fortius” (Faster, Higher, Stronger), new ways of training are established. In team sports, such as handball and football, speed and jumping ability are essential attributes on top level. Plyometric training exercises such as the vertical drop jump (VDJ) has in sports been used to improve these to attributes (Newton, Kraemer, & Hakkinen, 1999; Myer G. , Ford, Palumbo, & Hewett, 2005; Rønnestad, Kvamme, Sunde, & Raastad, 2008). The study by Myer et al., (2005) displayed at the same time that plyometric training, including VDJ, increases the dynamic knee stability and therefore benefits ACL injury prevention work. Recently VDJ was successfully used to examine gender-related differences in knee valgus motion of high school athletes (Ford, Myer, & Hewett, 2003; Hewett, Myer, & Ford, 2004). The studies revealed that the female athletes exhibited greater dynamic knee instability, with both greater maximum valgus angle and greater valgus Range of Motion (ROM). These studies were some of the first to exhibit, that athletes with decreased knee joint control in a VDJ could be exposed to an increased risk of ACL injury, (Ford et al., 2003). This was confirmed in a prospective study, where ACL injured athletes previously had displayed decreased knee joint control in a VDJ compared to non-injured athletes (Hewett, et al., 2005). Thus the VDJ is accepted as a good tool to screen athletes for decreased dynamic knee control.

Newer studies introduced overhead target/hurdle in order to stimulate the extrinsic motivation and improve effort in a VDJ (Ford, Myer, Smith, Byrnes, Dopirak, & Hewett, 2005; Smith, Kernozek, Kline, & Wright, 2011b). The collegiate athletes increased the performance when using an overhead goal/hurdle. Concurrently they looked at the technical changes that caused the possible effect. An increase in Ground Reaction Force (GRF) compared to no-target/hurdle situations was mainly explained by an increased recruitment of the knee extensors (Ford et al., 2005; Smith et al., 2011b). Even though both studies investigated kinetics and kinematics of lower

extremities, they failed to investigate the dynamic knee control. Since an increase in performance and thereby forces acting on the subject will increase, the dynamic knee control may be challenged further.

## **2.5 Aim of the Study**

The importance of good dynamic knee control should not be underestimated, and most athletes are aware of this when performing a VDJ. With an overhead target, the athletes will most likely focus more on reaching the target, than controlling the lower extremities during the VDJ. As mentioned above, recent studies give us a reason to believe that a target or hurdle will inspire the athletes' motivation and increase the effort (Ford et al., 2005; Smith et al., 2011). With greater effort, greater forces are in need to be controlled.

As a part of screening female athletes for the risk of ACL injury, the present study aims to investigate if higher effort is seen among the athletes as well as to investigate if the decreased knee control will occur, when an overhead target is introduced.

### **Aims:**

The aim of the study was to answer the two following questions:

- 1) Will elite female handball and football and handball athletes improve jump-height in overhead-target Vertical Drop Jumps compared to non-target Vertical Drop Jumps?
- 2) Will an overhead-target induce changes in lower extremity frontal plane and sagittal plane biomechanics?

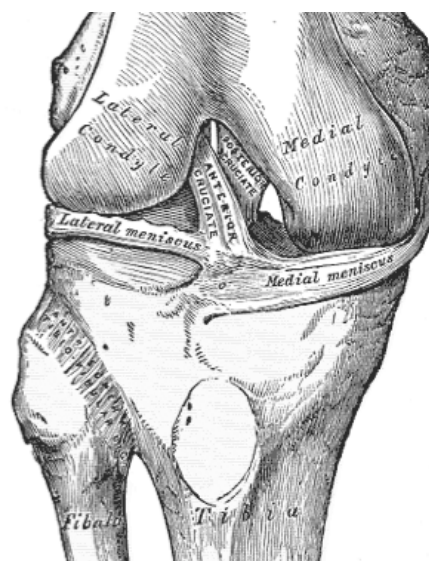


## 3 Theory

### 3.1 The Knee Joint

#### 3.1.1 Fundamentals of the Knee joint

The knee joint is the largest hinge joint in the human body, and it is a very complex joint (Dahl & Rinvik, 2007). It is also a synovial joint, which in terms means the joint has a gap between the articulating bones, and is protected by a synovial membrane that contains synovial liquid. A fibrous joint capsule that prevents the joint from dislocating covers the membrane, and is supported by the Medial Collateral Ligament (MCL) and the Lateral Collateral Ligament (LCL) that will keep the bones close together (Tortora & Derrickson, 2009). Femur, tibia and patella, articulates in the knee joint forming a tibiofemoral- and a patellofemoral joint (Zatsiorsky, 1998). Inside the synovial membrane the femoral and tibial ends are protected with thick articular cartilage. Between the cartilages, the medial and lateral meniscus is located. The synovial liquid covers and lubricates the components inside the synovial membrane to decrease friction, absorb shock and secure smooth movement of the bones in the joint (Tortora & Derrickson, 2009). As seen in figure 3.1 the two cruciate ligaments, the ACL and the posterior cruciate ligament (PCL) are connecting the tibia with the femur. The ligaments are located inside the joint capsule, but extrasynovially (Dahl & Rinvik, 2007).



*Figure 3.1: The Right Knee Joint seen from the front.*

The tibiofemoral part of the knee joint is capable of motion in all three planes. Flexion-extension motion in the sagittal plane, internal-external rotation motion in the horizontal plane and adduction-abduction motion in the frontal plane, this motion is also known as valgus-varus motion. The patellofemoral part of the joint is rather complex, where the patella glides and rolls on the femur during knee flexion, and performs a bit of translation as well (Zatsiorsky, 1998). The diversity of the knee joint is exceptional, as it functions as a solid and stabile pillar when standing, as well as being agile and flexible during motion.

### **3.1.2 Anatomy of the ACL**

The ACL is one of the four great ligaments in the knee joint, it connects tibia with femur. It has a footprint like origin anteriorly on the intercondylar area of the tibia, from where it extends posteriorly, laterally and proximally to attach on the medial part of the lateral condyle of the femur (Dahl & Rinvik, 2007). It is generally accepted that the ACL consists of an anteromedial and a posterolateral bundle, even though some recognizes two bundles as an oversimplification (Petersen & Zantop, 2006). In some cases only the anteromedial or the posterolateral bundle is torn for a partial rupture of the ACL (Engebretsen & Bahr, 2004). Gender-differences are present in the anatomy of the ACL. Female ACL has been demonstrated to have a significantly smaller minimum cross sectional area, thus the female ACL will be exposed to greater stress for an equal load (Chandrashekar, Slauterbeck, & Hashemi, 2005). The shorter female ACL will endure less elongation before failure because of increased strain. It is also suggested by Chandrashekar et al., (2005) that female ACL cannot absorb the same amount of energy as a male ACL because of less volume.

### **3.1.3 Composition of the ACL**

The ACL consists of ground substance and collagen, the connective tissue has limited content of elastic fibers and as a result not very flexible, but rigid and resistant to applied forces (Alter, 2004). A transition zone is situated between the ligament and the bone, consisting of fibrocartilage and mineralized fibrocartilage, resulting in a

smooth increase in stiffness from ligament to bone, and also alleviates stress concentrations at the points of attachment (Arnoczky, 1983).

Three various types of mechanoreceptors have been identified in the ACL as well as free nerve-endings (Schutte, Dabezies, Zimny, & Happel, 1987). The two types of Ruffini endings are slow adapting mechanoreceptors with a high sensitivity; they supply the central nervous system with specific and continuous feedback about variation of tension in the ACL according to position, angle and motion of the joint. The Pacinian corpuscles are fast adapting mechanoreceptors that are activated by any movement of the joint and as a result the speed of the movement. Thus the mechanoreceptors update the nervous system about direction, speed, acceleration and position of the joint during motion (Schutte et al., 1987). Free nerve-endings is known to function as pain-receptors, but since the ACL only has a small amount of these receptors, the ACL is relatively insensitive to pain (Schutte et al., 1987), which in turn may explain why experienced pain is limited when tearing the ACL.

#### **3.1.4 Function of the ACL**

The ACL plays an important role along with other knee ligaments, muscles and the joint capsule as a stabilizing factor of the knee joint. Even though the ACL is one of many stabilizers in and around the knee joint, the knee cannot function to its full potential during motion, without the ACL. The anteromedial part of the ACL is tightened when the knee is flexed and the posterolateral part is tightened in extension, shown in video 3.1 (Girgis, Marshall, & Monajem, 1975; Furman, Marshall, & Girgis, 1976). Thus the ACL contributes significantly to knee joint stability, because of the constant tension in some part of the ACL (Arnoczky, 1983). One of the main functions of the ACL is to prevent anterior translation of tibia in relation to femur, and that specific motion-pattern is also tested in a Lachmans test, which indicates an ACL tear (Furman et al., 1976; Engebretsen & Bahr, 2004). By cutting the ACL in fresh cadaveric knees, it was demonstrated that it also would prevent hyperextension of the knee joint as well as internal and external rotation (Furman et al., 1976). The same study failed to find differences in valgus motion between a “healthy” knee and a knee with a cut ACL.



*Video 3.1: Strain of the ACL during flexion and extension of the knee joint.*

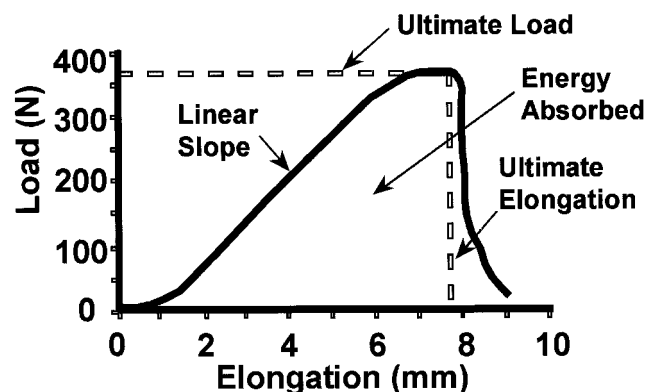
The MCL may explain the lack of result here, because it is known to be the primary restraint of valgus motion (Inoue, McGurk-Burleson, Hollis, & Woo, 1987; Woo, Debski, Withrow, & Janaushek, 1999). However it has been shown that the ACL also contributes to restrain valgus motion (Shin, Chaudhari, & Andriacchi, 2009).

It should be noted that direct in vivo assessment of ACL forces or strains is not currently feasible and the studies mentioned are based on simulation models. Shin et al., (2009) reported that ACL strain reaches a plateau with a valgus moment of about 50 Nm. As valgus moment increases the tibia performs an external rotation, which reduces the influence of the valgus moment on the ACL-strain (Shin et al., 2009). Consequently valgus motion combined with internal rotation will induce great strain on the ACL and that a valgus motion alone may not create a big enough strain to cause an ACL-tear (Shin, Chaudhari, & Andriacchi, 2011; Oh, Lipps, Athon-Millar, & Wojtys, 2012).

### **3.1.5 Structural properties**

The Load-Elongation Curve is explained by the structural properties of the ACL, as shown in figure 3.2. As the strain increases on the ACL, it will be elongated. The first part of the curve is called the toe region. The ligament has a relatively low stiffness, and the collagen fibers are elongated easily, so low force can create large elongation (Woo et al., 1999). The second part of the curve is the linear slope, the stiffness of the

ligament is higher and the elongation increases linearly with the load. At the end the ligament reaches its ultimate length, and the bone-ligament-bone complex will tear with further loading (Woo et al., 1999). The ultimate load will depend on different factors, such as age and orientation of tension. Younger (22-35 years) ACL specimens resisted greater load compared with middle aged (40-50 years) and older (60-97 years) ACL specimens (Woo, Hollis, Adams, Lyon, & Takai, 1991). The same study also found that with the tension orientated along the direction of the ACL, the ultimate load is greater compared with the tension orientated along the tibia. With load along the ACL, older specimens displayed rupture of the ligament more frequently than younger specimens, which in turn had a greater incidence of avulsions of the bone directly under the tibial insertion. With load along the tibia, older specimens still had a greater deal of ligament rupture, and the younger specimens exhibited greater deal of avulsions, but all specimens increased the part of rupture at the insertion of the ligament on the bone (Woo et al., 1991).



**Figure 3.2:** Load-Elongation Curve. The length of the ligament increases with the load<sup>1</sup>.

### 3.2 ACL injuries

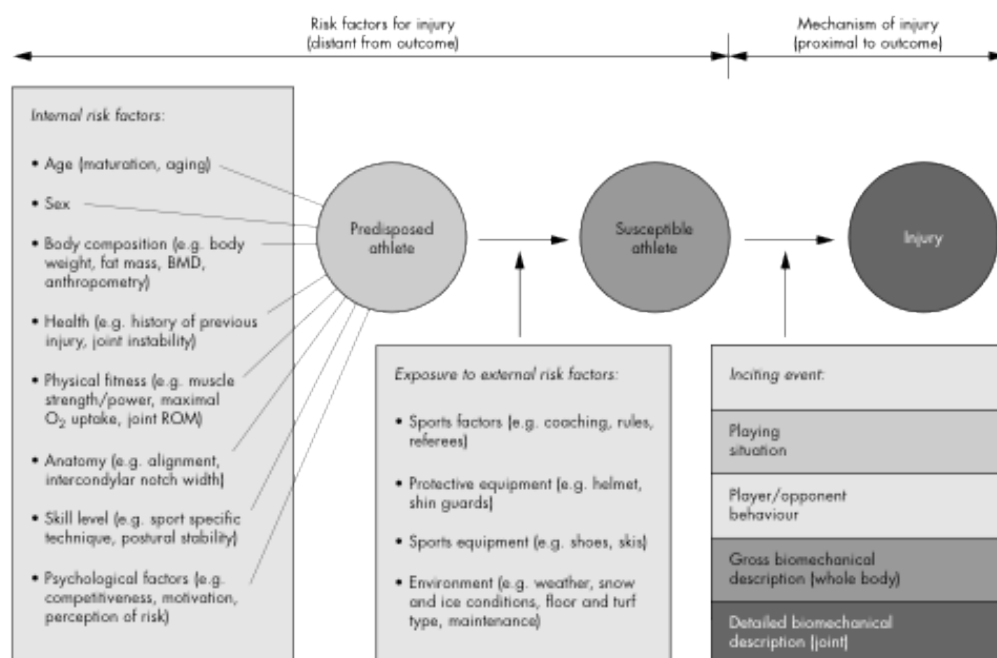
To understand the injury mechanisms and enable prevention of sports injuries, a comprehensive model has been developed (Bahr & Krosshaug, 2005). The model describes a list of possible internal risk factors that predisposes an athlete to injury. The risk factors for an ACL injury will be discussed more thoroughly below. The risk factors for an injury however include age and sex, the physical condition of the athlete

<sup>1</sup> From "Biomechanics of Knee Ligaments" by Woo, Debski, Withrow & Janaushek, 1999, Am J Sports Med, 27 (4), pp 537, © 1999 American Orthopaedic Society for Sports Medicine.

as well as health, body composition and anatomy. The skill level and psychology of the player will also affect the internal risk factors. The next step in the causation of injury is the athletes' exposure to different external risk factors. These will affect how susceptible the athlete is to injury. The nature of the sport will be one external risk factor, this for example being quantity and force of body contact (tackles, hits etc.), the rules of the game and how the referee enforces them. The choice of equipment and protective equipment, and the playing environment are also external risk factors. The last step in the injury causation model is a detailed description of the inciting event. It is split up into four parts:

1. Close description of the playing situation, at the moment of injury.
2. Behavior of the player and a possible opponent, at the moment of injury.
3. Gross biomechanical description, explaining the position of the whole body.
4. Detailed biomechanical description, explaining the position of joint.

In the next part of the theory, the risk factors and the mechanisms causing a non-contact ACL injury will be taken on, mainly based on a comprehensive model for injury causation (figure 3.3).



**Figure 3.3:** The comprehensive model for injury causation<sup>2</sup>.

<sup>2</sup> From "Understanding injury mechanisms: a key component of preventing injuries in sport" by Bahr & Krosshaug, 2005, Br J Sports Med, 39, pp 327, © 2005 BMJ Group.

### **3.3 Internal Risk Factors**

Internal risk factors are inherent in the athlete that could affect the risk of sustaining an injury. Some of the risk factors can be controlled through training or experience, whereas others are not possible to affect. Thus some people are more predisposed to sustain an ACL injury than others, and some are better to prevent them than others.

Age is also a risk factor for ACL injury. It has been reported (cadaver materials) that younger specimens of the ACL can resist greater loads compared to older specimens (Woo et al., 1991). The younger specimens were however grouped from 22-35 years, which is a common age-range for top-athletes in team sports. So we cannot with certainty confirm that a “younger” ACL in team sports can resist greater loads than an “older” one, and we do not know when the weakening of the ACL starts. But before full maturation, in adolescence, the muscle strength and recruitment do not follow the growth of the limbs, which leads to increased neuromuscular deficit (Hewett & Myer, 2011). This seems to be one of the main reasons for the greater ACL injury susceptibility in adolescent- and female athletes, and will be processed later on. The muscles surrounding the knee joint functions as support to the ACL, thus muscular strength and activation is paramount to control the extreme loads the knee joint can be exposed to. Muscular strength seems to increase up to about the age of 30 years, and keep a steady level until the age of 40-50 years (Larsson, 1982). It is known that females lose muscular strength at an earlier age compared to males, because of sarcopenia (Kamel, 2003). For female non-athletes muscular strength is greatest at the age of 21-30, and at the age of 41-50 there is a significant decrease in muscular strength in muscles of the lower extremities (Akbari & Mousavikhatir, 2012). The same study saw the function decreased significantly at the age range of 31-40. The lower extremity muscle strength and cross-sectional area has also proven to decrease in sedentary males at the age of about 40-55 years (Larsson, 1982). This age-induced decrease in muscle strength can however be delayed by systematic resistance training (Macaluso & De Vito, 2004; Walker, et al., 2011). The age of elite athletes ranges from about 18 to 35 years, and it would be fair to assume organized and systematic resistance training is a routine for athletes at this level. Thus the strength of the athletes can increase through out most of the career and amount of age-induced decrease in muscle strength is rather small, if not non-existing.

### **3.3.1 Health**

Athletes, who previously have been exposed to an ACL-rupture, have an increased risk for re-rupture. Regardless of the surgical procedure, former ACL-injured female athletes in high-force and pivoting team sports, has a re-injury rate of about 20% and about 10% went on to rupture the opposite "healthy" ACL (Bak et al., 2001; Myklebust et al., 2003a). When rupturing the ACL, there are typically other coincident injuries to the knee joint as bone bruise in 80-90% and meniscus injury in 75% (Engebretsen, Arendt, & Fritts, 1993), which are known to induce degenerative changes to the knee joint (Nakamae, Engebretsen, Bahr, Krosshaug, & Ochi, 2006). Thus one of the long-term consequences that follow ACL injuries is osteoarthritis (OA), and 6-12 years after the ACL injury, half of the injured athletes will develop OA, and up to 75% experience decreased knee function (Lohmander, Ostenberg, Englund, & Roos, 2004; Myklebust et al., 2003a). This shows that a previous injury to the ACL can affect the susceptibility to sustain another injury, and that athletes in top-level team sports often suffer from pain and limited function in the previously injured knee.

Specific knee joint laxity leads to different types excessive motion in the knee. Athletes with specific knee joint laxity can often display hyperextension and anterior-posterior tibiofemoral translation, as well as increased valgus-varus motion and internal-external rotation (Ford et al., 2003; Myer, Ford, Paterno, Nick & Hewett, 2008). Knee joint laxity will also lead to diminished proprioception in the knee, thus weakened ability to activate surrounding muscles in time to protect the knee against potentially dangerous forces (Rozzi, Lephart, Gear, & Fu, 1999). Female athletes in general display greater knee joint laxity than males, and this may contribute to the greater incidence of ACL injuries in female athletes (Rozzi et al., 1999).

The physical fitness level of the athlete can affect susceptibility to future ACL injury. First functional joint stability or joint stability during activity is partly controlled by the neuromuscular activation of the muscles across the joint. The fine-tuned dynamic and static muscle activation is essential to secure stability. When related to muscle/neuromuscular factors in the lower extremities, there seems to be two concerns, the activation and the strength. A well-known group of researchers has



categorized neuromuscular deficits into four groups of risk factors in jump landing (Myer, Ford & Hewitt, 2004; Myer, Jensen, Ford & Hewett, 2011b). For athletes that lacks sufficient muscle activation during strenuous motion, the ligaments, and in particular the ACL, will function as main stabilizers and not the surrounding muscles (Ford et al., 2003). The athletes show inability of dynamic control of the lower extremities in the frontal plane during landing and cutting movements, hence varus/valgus motion, are categorized as “*ligament dominant*” athletes (Myer et al., 2011b). Female athletes often show greater inability of dynamic control, due to late muscle activation of the lower extremity (Ford et al., 2003).

Athletes with limited muscle strength in the backside of the lower extremities will also strive to maintain dynamic knee stability. The characteristic of the jump and cutting technique is that the athlete has a more upright position (low hip and knee flexion) and a “harder landing”, Myer et al., 2011b, categorize this pattern as “*Quadriceps dominant*”. A deeper jumping technique will result in greater moments and greater activation of the hip extensors (Hamstrings, Gluteus Maximus). Lack of strength can therefore be the reason for the upright technique (vertical spine), usually seen in female athletes (Ford et al., 2005; Chappell, Creighton, Giuliani, Yu, & Garrett, 2007). Greater activation and strength in the quadriceps muscles compared to the hamstring muscles, can lead to an anterior tibial translation that will increase ACL-load. A co-activation of the hamstrings muscles will create a greater compression on the knee joint, reducing the anterior tibial translation, valgus/varus motion and further stabilizing the joint (Lloyd & Buchanan, 2001; Besier et al., 2003). Female athletes show greater inability to activate the muscles protecting the knee, and thus greater dynamic instability (Wojtys, Huston, Schock, Boylan, & Ashton-Miller, 2003; Ford et al., 2005)

Side to side differences in lower extremity constitutes a risk factor. This is a combination of, lack of strength and muscle activation in one leg compared to the other. It is made visible by uneven contact timing of the feet and diverse dynamic control of the limbs when landing (Myer et al., 2011b). This pattern was labeled as “*leg dominant*”. Athletes that rely on the stronger leg may put excessive strain on the “strong” knee, as well as the “weak” knee would be exposed due to decreased

absorption and stabilization abilities in the knee during high force motion (Ford et al., 2003).

The deficits in muscle strength and neuromuscular activation may be reduced with strength-, plyometric-, proprioceptive- and neuromuscular training. A jump-landing training program (technique training) can display a positive effect on landing mechanics and muscle strength in the lower extremities, increasing hamstrings muscle power and the hamstring-to-quadriceps peak torque ratio, reducing possible side-to-side differences and increasing knee stability in general (Hewett, Stroupe, Nance, & Noyes, 1996). Dynamic neuromuscular training, through plyometric and movement training, core strengthening, balance training, resistance training and interval speed training, has shown to increase a line of factors, such as the dynamic knee stability, jump height, speed and squat-strength (Myer et al., 2005). So using technique based prevention training, performing jump and landing exercises with deep hip- and knee flexion in a two-footed landing, and proper knee alignment (knees over toes) as well a narrow stance in cutting movements is considered key elements in preventing ACL injuries (Myklebust, Engebretsen, Brækken, Skjølberg, Olsen, & Bahr, 2003b; Myer et al., 2005).

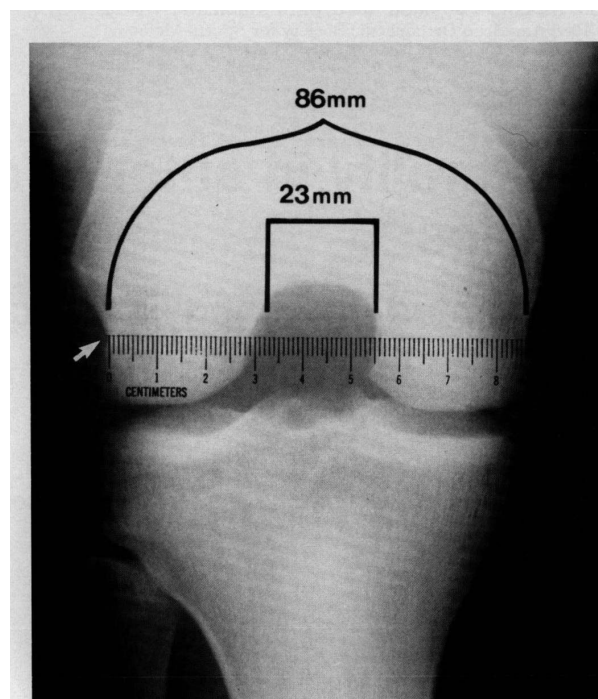
Muscular fatigue may also be a risk factor of future ACL injury. Fatigued muscles appear to absorb less energy, before they reach the amount of stretch that can cause injury (Mair, Seaber, Glisson, & Garrett Jr., 1996). This can leave a greater amount of energy needed to be absorbed from other structures, such as the ACL. The incidence of ACL injuries is greater during matches compared to training in handball players, however more cases of injury occurrence during second half compared to the first half has not been registered (Myklebust et al., 1997; Faunø & Wulff Jacobsen, 2006). So the effect of muscular fatigue on ACL injuries cannot be certain.

### **3.3.2 Anatomy**

There seem to be discrepancies about the supposed anatomic risk factors for future ACL injury, but common for most of them is that they usually are genetic or developed during maturation. Obvious gender differences in lower extremity anatomy

are reported, including limb alignment, joint laxity and muscle development (Griffin, et al., 2000).

The size of the femoral notch seems to play a part in the risk of future ACL injury. A measurement technique using radiographs of the femoral notch has been developed to determine the risk of ACL injury (figure 3.4). Notch Width Index (NWI) is the ratio of the size of intercondylar notch divided by the condylar width at the level of the popliteal groove (Narrowest part of the condyle). A ratio  $\leq 0.18$  and  $\leq 0.20$  is related to intercondylar notch stenosis for females and males respectively (Souryal & Freeman, 1993). A small NWI was significantly correlated to non-contact ACL injuries in cutting and pivoting sports. During motion, flexion/hyperextension of the knee and rotation of the tibia, the ACL can impinge over the femoral condyle. In a stenotic knee there can be a greater chance of impingement, which increases the stress of the ACL, and thus increase the ACL-rupture risk (Souryal & Freeman, 1993; LaPrade & Burnett, 1994).

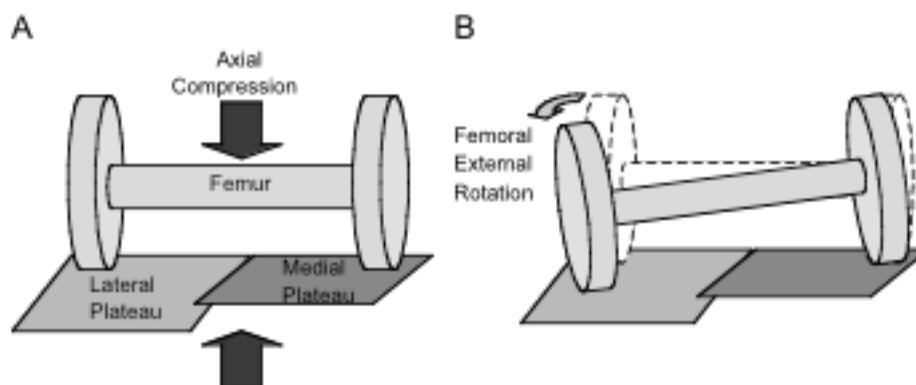


**Figure 3.4:** Radiograph as used for measurement of NWI<sup>3</sup>.

<sup>3</sup> From "Bilaterality in anterior cruciate ligament injuries: Associated intercondylar notch stenosis", Souryal, Moore & Evans, 1988, Am J Sports Med, 16, pp. 450, © 1988 American Orthopaedic Society for Sports Medicine.

Athletes with a stenotic knee is also considered to have a smaller ACL compared to those with greater notch width, and a smaller ACL will in turn have tolerate less strain before rupture (LaPrade & Burnett, 1994; Chandrashekar et al., 2005). Later research has also confirmed that with smaller notch width a smaller ACL often comes along. A small notch width combined with normal ACL-size may increase impingement, which in turn can deteriorate the ACL-function and increase the injury risk (Simon, Everhart, Nagaraja, & Chaudhari, 2010).

Researchers have in recent years been looking at the tibial plateau slope, and its possible connection to ACL injuries. A steeper posterior Lateral Tibial Slope (LTS) is found to be a possible risk factor. When forces act on the knee joint, the femur will slide off the plateau posterior, as seen in figure 3.5, using the Medial Tibial Slope (MTS) as pivot point (Simon et al., 2010). This will in turn force an external rotation of the femur on the fixed tibia, increasing the strain on the ACL (Simon et al., 2010; Şenışık, et al., 2011).



**Figure 3.5:** A simplified figure of the sliding of the femur on the lateral tibial plateau. (A: Before loading B: After loading)<sup>4</sup>.

The LTS is proven to be significantly greater in ACL-injured athletes compared to non-injured athletes (Stijak, Herzog, & Schai, 2008; Simon et al., 2010). Şenışık, et al., (2011) saw that there were no differences in LTS between football players and sedentary people, but nonetheless observed greater LTS in players who ruptured their ACL compared to the non-injured players. This suggests that a greater posterior LTS increases risk of sustaining an ACL injury in team-sport athletes.

<sup>4</sup> From “A case-control study of anterior cruciate ligament volume, tibial plateau slopes and intercondylar notch dimensions in ACL-injured knee”, Simon, Everhart, Nagaraja & Chaudhari, 2010, J Biomech, 43 (9), pp. 1705, © 2010 Elsevier Ltd.

The postural stability can affect the lower extremities during motion. Dysfunction of the core muscles can provide greater loads on the lower extremities. The trunk musculature cannot answer the presented dynamic demands, leading to a lateral trunk position that will alter the posture of the limbs into an unbeneficial position. The non-advantageous posture can lead to direct increased load on the knee joint and also inhibit adjacent muscles to perform their protective work of the joint (Myer et al., 2011b). The same study categorizes this neuromuscular deficit as “*Trunk dominance*”.

Anterior pelvic tilt is hypothesized to have a few different effects on the limb alignment. Anterior pelvis tilt may cause greater valgus of the knee, internal rotation of femur and subtalar pronation, as well as lengthening the hamstrings muscles and thereby reducing their function to prevent anterior tibial translation and hyperextension of the knee; motions that may be related to increased ACL-strain and -injuries (Shultz, Nguyen, & Beynnon, 2007). It might not be the pelvic position itself, but rather the limb alignment changes that cause the increased risk. An increase in femoral anteversion may also affect the limb alignment, as it can weaken the effect of the gluteus medius as an abductor of the hip. This can create a hip adduction and thus valgus motion in the knee (Shultz et al., 2007).

As mentioned earlier, prevention programs focusing on jump/landing-technique has turned out beneficial on ACL injuries and dynamic knee control. Seeking to automate two-footed landings with knees over toes, and a narrow the stance in landings and cutting movements has shown a trend towards decreased incidence of ACL injuries in handball players (Myklebust et al., 2003b). The same study compared the elite players who followed the program with the ones who failed to do so, and elicited a significant decrease in ACL injury incidence. Earlier a jump-training program found significant reduction in peak landing force and increased knee stability, measured as valgus and varus moments, during a volleyball block jump (Hewett et al., 1996). This was supported by (Myer et al. 2005), who also found increased knee stability after a six weeks jump landing technique program, consisting of deep landings (high knee flexion) and proper knee alignment. No doubt technique training is beneficial for reducing ACL injury incidence and increasing dynamic knee stability. Myklebust et al. (2003b) showed best results with the elite player compared to players in lower

divisions, probably because of closer follow up and a greater number of training hours. Implementing jump-landing technique training for younger players could be more beneficial, as technique patterns for landing and cutting movements are more effectual (Myklebust et al., 2003b).

For handball players, up to eighty percent of ACL injuries occur in matches versus training (Myklebust et al., 2003b). Greater incidence of ACL injuries during match is similarly observed in football and volleyball (Engström, Johansson, & Törnkvist, 1991; Ferretti, Papandrea, Conteduca, & Mariani, 1992). Elite athletes have more training hours than match hours, and should be more focused and motivated during matches, so these results may seem a bit peculiar. However an explanation may relate to greater intensity and more unanticipated situations during a match.

Myklebust et al., (1997) showed a greater incidence of ACL injuries in training among elite players than in division 2 and 3, which was explained by higher intensity. Greater valgus and internal rotation moments have been exhibited during unanticipated movements, probably due to lack of great and specific activation of the muscles stabilizing the knee, which occurred during preplanned movements (Besier et al., 2003). Thus unanticipated movement may increase the risk of sustaining an ACL injury.

### **3.4 External Risk Factors**

#### **3.4.1 Environment & Equipment**

The playing-surface and choice of footwear can affect the risk of future ACL injury. The friction coefficient will be determined by interaction of the sole of the footwear on the playing-surface. A greater friction coefficient will cause higher torsional resistance and greater strain on the ACL during motion, thus increasing the risk of an ACL injury (Heidt, Dormer, Cawley, Scranton, Losse, & Howard, 1996). Studies have shown a greater incidence of general injuries on artificial grass/turf compared to natural grass (Árnason, Gudmundsson, Dahl, & Jóhannsson, 1996), as well as a higher incidence of general and ACL injuries in indoor football compared to outdoor (Hoff & Martin, 1986; Rochcongar, Laboute, Jan, & Carling, 2009). Indoor-sports arenas can have different playing-surface, and the artificial floors seem to have and

higher incidence of ACL injuries compared to wooden floors, most likely because of the generally greater friction found in artificial floors (Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2003). In outdoor sports the weather conditions can also affect the risk, low evaporation and high rainfall has proven to reduce the incidences of non-contact ACL injuries in Australian footballers (Orchard, Seward, McGivern, & Hood, 1999). In the National Football League, athletes obtained 95% of the non-contact ACL injuries in dry weather conditions (Scranton, et al., 1997). Cold weather has also been associated with lower incidence of ACL injuries on both natural and artificial grass, probably due to reduced friction (Orchard & Powell, 2003).

This means that the surface can affect to which degree an athlete is exposed to an ACL injury, then the temperature and the amount of precipitation again can change the properties of the surface. For an team sport athlete this is a dilemma, on one side she wants high friction between the shoe and surface, for greater response to direction changes, cutting maneuvers, jumping, feinting etc. and at the same time not expose oneself to an unnecessary increased injury-risk. Through long experience in the sport, elite athletes find the footwear that suits them the best, regards to response from ground and not inhibiting the performance, but whether they consider the injury risk is rather doubtful. So the friction between athlete and surface, which is the main external non-contact ACL injury risk factor, can be further altered by the choice of footwear in training and match situations.

### **3.5 Mechanisms of Injury**

#### **3.5.1 Playing Situation**

As mentioned earlier up to eighty percent of ACL injuries in female team handball occurs in match situations (Myklebust et al., 2003b), and there has been found a similar distribution in football and volleyball (Engström et al., 1991; Ferretti et al., 1992). In a match the handball-player is in an attacking situation 80-90% of the time an ACL injury occurs (Myklebust et al., 1997; Olsen et al., 2004). In the study by Olsen (2004), three former or present national team handball coaches analyzed video recording of 20 ACL injuries in match, and most of them (15) occurred during high-speed motion. In football 62 of 105 (59%) players was situated in the opponents half

of the field, when they sustained the injury, and 18 of those cases was inside the penalty box (Faunø & Wulff Jakobsen, 2006). Eighty-eight players could determine their role at the time of the injury, and 50 of them recalled having an offensive role.

### **3.5.2 Player/Opponent Behavior**

It has been reported that ACL-injured handball players did not recall the injury-situation as dangerous, as being impeded or perturbed by an opponent, or as an abnormal or unfamiliar movement (Myklebust et al., 1997). However newer studies has shown that in most attacking situations the athlete is handling the ball and is faced towards the goal, in close proximity of an opponent (Myklebust et al., 2003b; Olsen et al., 2004). A video analysis of 10 ACL ruptures in handball and basketball, displayed indirect contact in 4 of the cases, 7 of the players were in ball-contact and the last 3 had just shot or passed the ball (Koga, et al., 2010). Seven players were performing a cutting motion, and the last three were landing on one leg when rupturing the ACL (Koga, et al., 2010). The video analysis by Olsen et al., (2004) reported indirect contact in 6 (30 %) of the injury-situations, meaning a contact to another body part than the lower extremities. Almost half of the players were considered to be out of balance when injured. Twelve of the handball players were perturbed (trying to avoid collision, pushed or held) in a way that their opponents inhibited balance and/or coordination. So from a view that stated opponents presence would not affect the risk of an non-contact ACL injury, it seems to have moved towards that almost half of the injuries occurs when players is perturbed by an opponent in some way, causing indirect- or non-contact ACL injuries.

In the studying of ACL injuries in football about 80% of the injuries were in non-contact situations (Faunø & Wulff Jakobsen, 2006; Rochcongar et al., 2009). A major part (62%) of ACL injuries occurred in situations where the player was in ball contact, and in 92 of the 105 cases, the players performed either a turn/cut or a landing from an aerial duel (Faunø & Wulff Jakobsen, 2006). Rochcongar et al., (2009) saw that players were performing a pivoting motion in 34.5% of the injury situations. About 10% of the football players experienced an ACL injury when being tackled by an opponent (Rochcongar et al., 2009). In eleven of 105 cases, an opponent fouled the injured player so harsh, that he was penalized with a yellow or red card



(Faunø & Wulff Jakobsen, 2006). So also for football players it seems that opponents tackling or forcing a cutting motion, in some way contributes to the inciting event.

### **3.5.3 Gross Biomechanical Description**

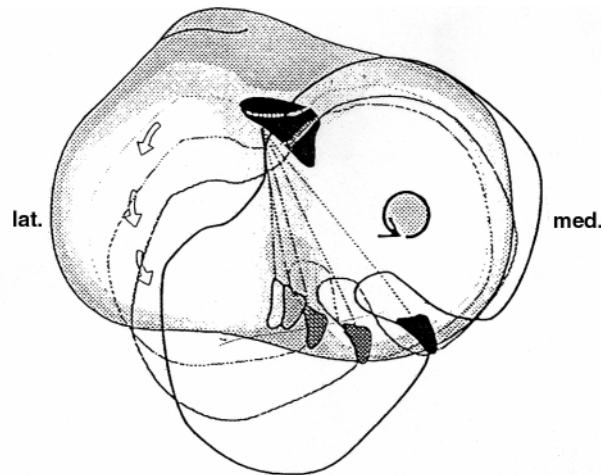
In team sports the usual injury-risk movement patterns is when changing or stopping high-speed motion such as cutting, landing, decelerating or pivoting. For these situations the foot is fixed on the surface, more distal compared to the knee, and the weight is usually distributed a hundred percent on the injured leg (Olsen et al., 2004). As mentioned above the athletes were familiar with the movement in the injury-situation, as they had performed the movement many times before, neither did they recall anything unusual about the moment of injury (Myklebust, 1997, 1998), however other athletes remembered to have a wider foot stance than usual, when rupturing the ACL (Olsen et al., 2004). This may be because an opponent is forcing the player out of balance.

### **3.5.4 Detailed Biomechanical Description**

When analyzing a video or questioning an athlete after an ACL injury. Establishing the exact time the ACL rupture took place has been a great challenge. It can be difficult to determine on a video clip, and sometimes athletes do not feel any significant pain when it ruptures or only have a vague memory of the situation. This has been a challenge until recent inauguration of Model-Based Image-Matching (MBIM).

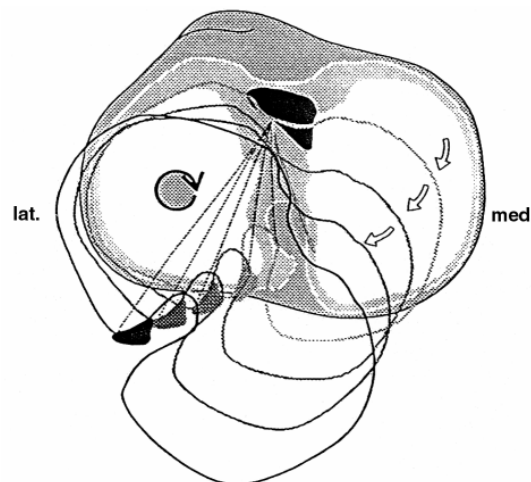
There seems to be no doubt that when rupturing the ACL in a non-contact incident, athletes are planting the foot with a straight or slightly flexed knee and valgus during both cut-movement and landing (Olsen et al., 2004). There also seem to be external or internal rotation of the tibia at the time of rupture. This is supported by Ebstrup & Bojsen-Møller (2000), who reported either, valgus and internal femoral rotation or varus and external femoral rotation in the knee at ACL rupture. Matsumoto & Seedhom (1993) saw that in a flexed knee, a femoral external rotation would force the meniscus to move up to 12-14 mm antero-posteriorly on the lateral side, and about 4 mm on the medial side. This means that the meniscus will pivot about a point located near the medial side (figure 3.6). That movement strains the ACL, and the femur will

slide on the tibia to equalize this strain. When the knee is in varus, the medial femur condyle will remain in the concave part of the tibial condyle, and compression forces from the bodyweight will inhibit any slide thus increasing the strain on the ACL (Ebstrup & Bojsen-Møller, 2000).



*Figure 3.6: The left knee joint in varus load, seen from above. During an external rotation of the femur, pivoting will occur about the medial plateau of the tibia<sup>5</sup>.*

When the knee is loaded in valgus, seen in figure 3.7, the medial part of the knee joint seem to be unloaded, and the medial passive structures will be put to tension, the pivot point will shift to the near lateral side of the meniscus, and the ACL will be endangered by the internal rotation of the femur (Ebstrup & Bojsen-Møller, 2000).



*Figure 3.7: The left knee joint in valgus load, seen from above. During a femoral internal rotation, pivoting will occur about the lateral plateau of the tibia<sup>5</sup>.*

<sup>5</sup> From “Anterior cruciate ligament injuries in indoor ball games”, Ebstrup & Bojsen-Møller, 2000, *Scand J Med Sci Sports*, 10, pp. 115, © 2000 Munksgaard.

However through MBIM, researchers have been able to estimate the time point when the ACL injury occurs, as well as a detailed kinematic description of the knee joint at the time of injury (Koga, et al., 2010; Koga, Bahr, Myklebust, Engebretsen, Grund, & Krosshaug, 2011). As mentioned above other studies have stated that a slightly flexed knee with a combined valgus motion and tibial rotation is present when the ACL ruptures, this is confirmed by Koga et al, (2010). In the 10 handball and basketball cases studied by Koga et al., (2010), there was no valgus at the initial contact (IC) with the ground, this however increased to an average of  $12^\circ$  and an estimated  $GRF_{PEAK}$  of  $3.2 \times \text{Body Weight (BW)}$  40 mSec after IC. At the same time points an external tibial rotation of  $5^\circ$  was changed to  $8^\circ$  of internal tibial rotation. However the MBIM showed the players had on average a  $17^\circ$  external tibial rotation at 300 mSec after IC. A MBIM study estimated a 9 mm anterior tibial translation in an ACL rupture situation after 30 mSec (Koga, et al., 2011), as well as the consistent abrupt increase in valgus and internal rotation in all 10 cases, suggests that ACL ruptures occurs within 40 mSec after IC (Koga, et al., 2010). Koga, et al., (2011) suggests that valgus motion and lateral compression will force internal tibial rotation and anterior tibial translation, finally resulting in an ACL rupture.

The model is comprehensive and elicits the important factors concerning an ACL injury, in team sports and especially in handball. It may give athletes and coaches an overview of the internal and external risk factor, which can decrease the susceptibility to injury. It aims to thoroughly describe the mechanisms during the inciting event causing an ACL injury, both according to athlete and opponent behavior.

### ***3.6 Screening for potential future ACL injury***

#### **3.6.1 3D motion analysis/Lab based screening**

As the female participation in sports has increased, so has the amount of ACL injuries, thus development of a reliable screening tool and effective prevention programs have been natural. A great deal of internal and external risk factors is the basis of the degree of exposure, and combined with the inciting event, causing the injury (Bahr & Krosshaug, 2005). The movement pattern of the inciting event has

been investigated, and this of course is a complex movement with a combination of motions (Olsen et al., 2004). Dynamic knee stability in VDJ has shown to be a good indication of the ACL injury risk, and knee abduction moment has been proven to predict ACL injuries with 73% specificity and 78% sensitivity (Hewett et al., 2005). 3D-motion analysis has over the last decade been accepted as providing reliable and accurate measurements. Now the three-dimensional motion analysis systems are recognized as “the gold standard” in screening of neuromuscular control in athletes (McLean et al., 2005; Stensrud et al., 2011).

### **3.6.2 2D analysis/clinic based**

The problem with the three-dimensional motion analysis systems is its application to a large scale of athletes, because of the great cost of time, space and finances. Therefore, the molding of new ways of screening for dynamic knee instability have seen daylight, and one of them being two-dimensional video analysis. The joint-angles for the lower extremities are determined by estimation of the hip, knee and angle joint center in every frame of the video recording. Even though the 2D joint angles are estimated to be greater than the 3D-analysis angles, there is good correlation 2D and 3D valgus angles, suggesting that the inter-subject differences will be exposed in even with the 2D approach (McLean et al., 2005). Earlier it has been clarified that athletes in motion will have an internal or external rotation of the femur in relation to the tibia (Ebstrup & Bojsen-Møller, 2000; Koga, et al., 2010). This rotation can obscure the “real” picture of the motion, because a flexion of the knee may look like a valgus motion on a 2D-video recording (McLean et al., 2005). The same study has seen that trunk lean can occlude the hip joint during motion and thus complicate the precise determination of the joint. This is not the problem in side step and side jump, but can be an issue in a VDJ, impeding the 2D-analysis. Newer research comparing 2D and 3D-valgus analysis, found a very good correlation ( $r = .87$ ), without mentioning some of the issues stated above, and they did not report any difference in magnitude of the valgus angles (Myer, Ford, Khoury, Succop, & Hewett, 2010). There can be some issues in using the 2D-analysis, but it seems to be a valid screening tool to differentiate athletes in dynamic knee control. However the 2D measurement is not good enough to provide precise description of knee valgus magnitudes (McLean et al., 2005).

### **3.6.3 Field-based**

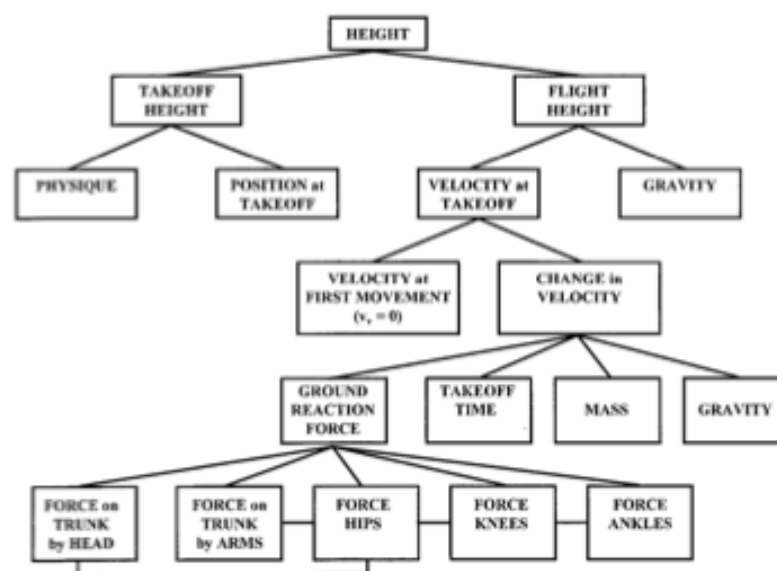
In order to facilitate screening athletes for dynamic knee control, the reliability of subjective assessment have been undertaken (Stensrud et al., 2011). A single observer assessed athletes, according to a scale formed by a group of experienced physiotherapists. Having looked at three different movements Stensrud et al., (2011) found that a single leg VDJ not was a good way to assess poor knee control. However VDJ and Single Leg Squat (SLS) revealed neuromuscular deficits in athletes, and showed good correlation with 2D-video analysis, as valgus FPPA increased with assumed reduced knee control (good-, reduced- and poor-performance). Even though the VDJ and the SLS both were found useful, they had poor correlation as only half of the athletes assessed with poor knee control in each test were evaluated to have poor knee control on the other test. Thus both tests should be used to reveal the most athletes with poor knee control, as the tests seem to challenge different aspects of physique and motor control (Stensrud et al., 2011). The difficulties with the field-based assessment are the inter-observer differences that will occur, based on experience and “screen-training”. Application of this method in a prospective study could be the next step to acknowledge as a valid screening tool.

Different scoring systems have been developed in the chase to facilitate screening methods. A Landing Error Scoring System (LESS) during VDJ did not seem to work, as the whole scale (0-17) was not used and the different categories (Excellent, good, moderate and poor landing technique) could not predict increased risk of future noncontact ACL injury (Smith, et al., 2011a). Another scoring system, a “clinician friendly nomogram” seems more reliable as it shows high sensitivity and specificity in predicting high knee abduction moment (KAM) in female athletes (Myer et al., 2010; Myer, Ford & Hewett, 2011a). The nomogram includes measurements of knee valgus and flexion ROM during a VDJ from two-dimensional video recordings, and measurements of tibia length, mass and quadriceps-to-hamstring ratio. Thus the nomogram is in need of video recording equipment, and should be compared and associated with other clinic-based approaches according to time, space and cost. An application of this method could not support this, as there was no difference in predictive KAM, between ACL injured athletes and their matched controls (Goetschius, et al., 2012). There has been methodological and design concerns about

this study, compared to the ones used in the development of the nomogram. If the existing differences are enough to obscure the outcome is unsure and debated (Myer, et al., 2013). So it could seem like the nomogram, though theoretically good, is not likely applicable as a screening tool.

Thus the laboratory-based method still is the favored tool to assess ACL injury risk, but researchers should keep striving to further develop a useful clinical/field friendly method, that can embrace a larger number of athletes.

### 3.7 Jump performance



**Figure 3.8:** The deterministic model of vertical jump<sup>6</sup>.

#### 3.7.1 Ground Reaction Force

The mechanisms that determine the height in a jump are presented in a deterministic model (figure 3.8). Change in vertical velocity is vital to alter jump height. Ground reaction force (GRF) is one of the determining factors of vertical velocity together with takeoff time, mass of the athlete and gravity. The force from the head and arms on the trunk in turn determines GRF, as well as force developed from the hips, knees and ankle joints.

<sup>6</sup> From "Anatomy, Mechanics and Human Motion", by J. G. Hay and J. G. Reid, 1988, Englewood Cliffs, NJ: Prentice-Hall. © 1988

During a VDJ the hip extensors contribute the most to create force, then the plantar flexors and the knee extensors contribute the least (Ford et al., 2005; Smith et al., 2011b). Even though a sole increase in the knee extensor moment, have been displayed in situations where the jump height increased (Ford et al., 2005; Smith et al., 2011b). In a more bounce type of jump, compared with the Counter Movement Jump (CMJ), the athletes increase jump height with augmented recruitment of the plantar flexors and knee extensors (Bobbert, Huijing, & van Ingen Schenau, 1987), however Smith (2011b) failed the show the same increased recruitment of the knee extensors in the bounce drop jump. The peak verticalGRF (vGRF) is increasing when athletes perform a bounce drop jump, suggesting a more intense force production (Smith et al., 2011b). Track and field athletes have shown ability to develop greater peak vGRF compared to volleyball players, but the volleyball players still jumped higher in a drop jump test (Kollias, Panoutsakopoulos, & Papaiakevou, 2004). When testing athletes jump performance there are two options regarding the arms, first keeping the arms akimbo or have no restrictions, which usually leads to the athlete performing an arm swing. According to the deterministic model above, the arms will affect the trunk with a vertical momentum and eventually increase the jump height. Studies have found that an active arm swing will help the athlete to increase jump performance as much as a 20% (Feltner, Frascchetti, & Crisp, 1999; Feltner, Bishop, & Perez, 2004).

### **3.7.2 Takeoff Time**

Laws of the physics, and the 2<sup>nd</sup> law of Newton ( $F = ma$ ) decide jump height. Rewritten the 2<sup>nd</sup> law of Newton looks like this:  $Ft = mv$ . Where the Force created over time is an impulse that affects the velocity of the mass (Caldwell, Robertson, & Whittlesey, 2004). So it is important to state that jump height is not decided by the peak vGRF, but by the total force produced during the contact time. This is why there seems to be discrepancies about the takeoff time and its influence on the jump height, in the literature. On one side, decreased contact time is related to increased jump height (Bobbert et al., 1987; Smith et al., 2011b). This is suggested to be because of better use of the Stretch Shortening Cycle (SSC)(Ford et al., 2005). Before the amortization phase, the extensor muscles are stretched and therefore activated. The stretch will increase the force in the tendomuscular complex and storage of this elastic

energy that will result in a more powerful subsequent concentric phase (Newton & Kraemer, 1994).

However, female the athletes investigated by Ford et al., (2005) had no change in contact time when they increased the jump height. As mentioned above athletes with greater peak vGRF do not necessarily jump higher, but athletes that utilize less time in contact with the ground, often increase their peak vGRF, and in most cases it means an enhancement of performance (Bobbert et al., 1987: Smith et al., 2011b), but as mentioned above volleyball players have shown ability to jump higher, despite displaying less peak vGRF (Kollias et al., 2004), because they developed greater force during the whole, longer contact time i.e. created a greater impulse. The laws of the physics implies that great force combined with long contact time would be the most way beneficial to achieve the greatest jump height. But these two factors work in opposite directions, as high force often means short contact time and vice versa. Too much flexion in hip and knees in a VDJ will also create great moments that will inhibit the development of vertical force, thus negative effect on jump performance. Every athlete will have an optimal joint flexion angle to perform best. For football players the optimal flexion angle may not be as great as for volleyball players (Smith, Ford, Myer, Holleran, Treadway, & Hewett, 2007, Moran & Wallace, 2007), hence the execution of a jump in the specific sport. Greater flexion angles will create longer displacement, thus greater impulse. Greater flexion angles will create greater external moments, so if an athlete cannot overcome this, force will dissipate.

Athletes utilize muscles around the hip, knee and ankle joints to perform a VDJ, but mainly improve their jump performance by increasing the recruitment of the knee extensor muscles (Ford et al., 2005: Smith et al., 2011b). When increasing the recruitment from plantar flexors and hip extensors it is often connected with a technique change to a more bounce or upright style of jumping (Bobbert et al., 1987). By swinging the arms during the takeoff phase, it will affect the trunk and the athlete will generate further vertical force (Feltner et al., 1999, 2004). A more rapid jump, with shorter stance time and less amortization phase is related to greater jump height, as the athlete can be utilizing the SSC better (Ford et al, 2005), but it however looks like athletes will perform best using a jump technique that is similar to the one used in their specific sport (Kollias et al., 2004).



## **4 Method**

### **4.1 Design**

The present study was a part of a large ongoing prospective cohort study in which all female handball and football teams of the Norwegian elite series and the Norwegian National Team were invited to participate.

The aim of the prospective cohort study was: *to evaluate anatomical, neuromuscular and biomechanical risk factors for ACL injury in a prospective cohort study of elite female team handball and football players.*

The prospective cohort study consisted of different tests to evaluate anatomical, neuromuscular and biomechanical risk factors. The present study focused on one of the biomechanical tests, where the athletes' dynamic knee control in a VDJ was examined during two conditions (with and without an overhead target).

The relevant test will be described thoroughly below, while the rest of the tests are not in concern of this study, and will only be mentioned.

The study started with a series of screening tests at the Norwegian School of Sport Sciences (NIH) in June 2007. Injuries were recorded during all team activities during the 2007/08 season. Additional screening was conducted on promoted teams/new players every season since 2007, and will continue for an unknown number of years. Moreover, female football players from the Norwegian elite series have been included in the study since 2009. It is expected that 40-50 new players is included pr year. The amount of handball and football players tested now amasses to approximately 700.

### **4.2 Subjects**

The data in the present study included handball players tested in 2009 and football players tested in 2010. A total of 72 handball- and football players were examined (36 from each sport). There was no difference in age between players in the two sports. The handball players however had significantly greater height and bodymass compared to the football players (data displayed in table 4.1).

**Table 4.1:** Anthropometric data for the two groups separately, and for all of the athletes

Players	Age (years)	Height (cm)	Min. (cm)	Max. (cm)	Bodymass (kg)	Min. (kg)	Max. (kg)
Handball	22.4 ± 4.0	172.3 ± 6.0 <sup>b</sup>	160.5	183.0	69.9 ± 7.3 <sup>b</sup>	54.9	85.3
Football	22.3 ± 4.0	164.8 ± 5.7	154.0	175.5	62.2 ± 6.7	50.3	76.5
All	22.4 ± 4.0	169.0 ± 7.0	154.0	183.0	66.1 ± 8.0	50.3	85.3

Values are mean ± SD, minimum and maximum value. Handball n = 36, Football n = 35, All n= 71,

<sup>b</sup> Significant difference between sports at  $p < 0.05$

### 4.3 Testing procedure

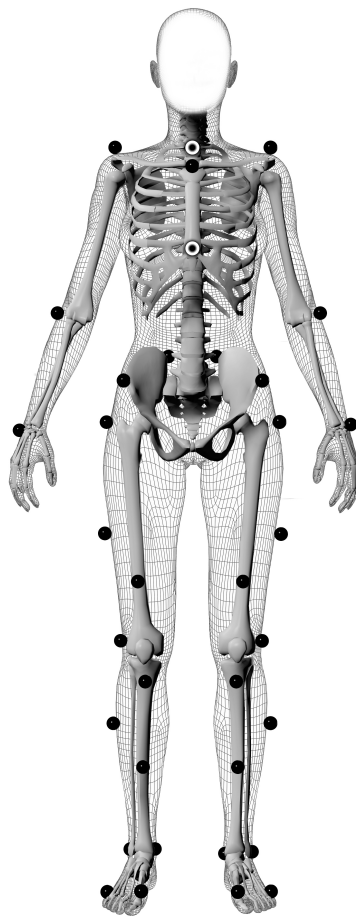
The athletes tested in this study came from all across the country, and players from Trondheim, Kristiansand, Bergen and Tromsø were flown in the day before testing. The athletes were paired and performed tests in a random order at the 8 different test stations. The athletes used about 30-60 min. at each test station.

### 4.4 Test stations

1. Marker placement.
2. Sidestep cutting and vertical drop jumps.
3. Marker removal and anthropometric measures.
4. Isokinetic strength testing.
5. Single leg squat and navicular drop.
6. Shoulder tests, genu recurvatum and testing for generalized joint laxity.
7. Hamstrings mobility, hip abductor strength, hip anteversion, KT-1000 star reach test.
8. Balance, legpress.

The present study was based on the data from the two-legged maximal VDJ. The data was collected using 3D motion analysis to describe lower extremity movement patterns (see below):

In preparation for the 3D motion analysis, 35 individual reflective markers were attached to specific palpable landmarks (figure 4.1). The height (in centimeters) of the barefooted athlete was found. Warm up consisted of 5-8 min easy-moderate ergometer bicycling and 3 submaximal countermovement jumps. Before the trials started, the 3D motion analysis system was calibrated according to the recommendation from the manufacturer. The recordings started with a static trial of the athlete, standing on the force plates, in anatomical position with the arms horizontal. This was to determine the anatomical coordinate systems and provided us with the lower limb alignment of the athlete, centre of mass (COM) as well as the bodyweight.



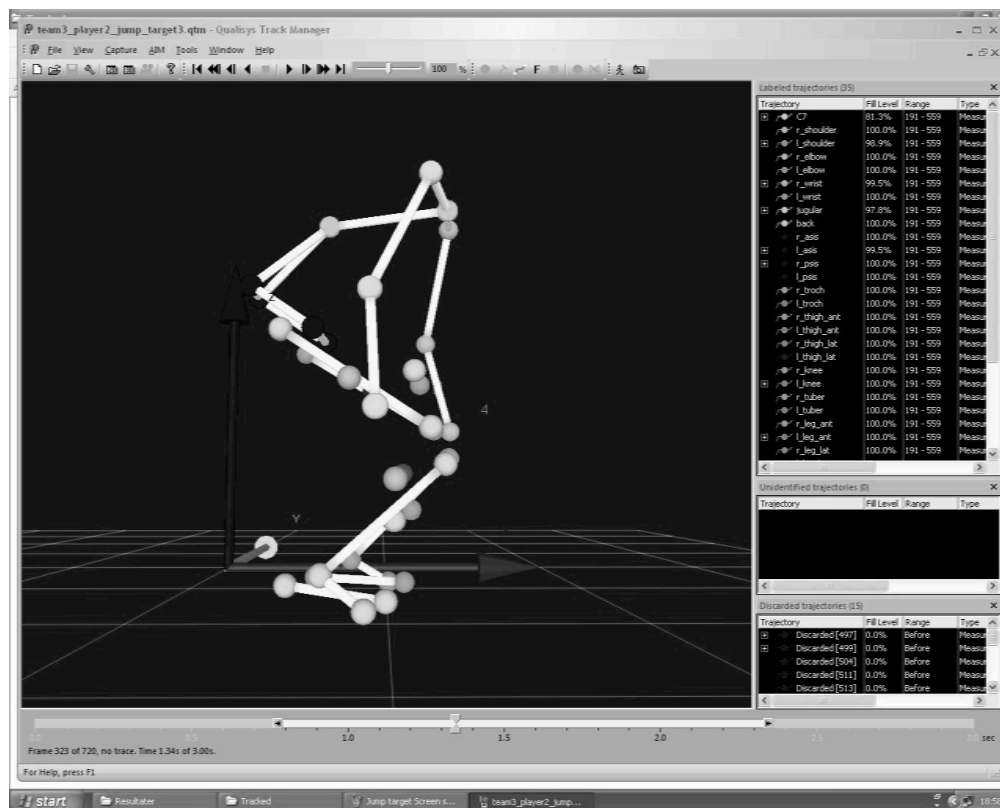
**Figure 4.1:** A model of the marker placement on the athlete. The half colored markers are located posterior on the athlete.

The subject was standing on a 30 cm high box and was instructed to drop directly down off the box, with a foot on each of the force plates and immediately perform a maximal vertical jump. The athlete performed five approved VDJs without an

overhead target. The athletes were free to select their own jump-technique in terms of knee angle (sagittal plane) before reversing the motion. There were no restrictions to the arm-use, so athletes found it natural to swing the arms to use them as upwards thrust. The no restriction of arm-use was a way to replicate the method of Myer et al., (2005), where the subjects were instructed to reach for a basketball rebound in both situations.

The data-collection of the knee joint motion commenced at the instant of contact on the force plate (Initial Contact) and until takeoff (TO). For the VDJ to be approved, the athlete had to drop and not jump down from the box and hit with one foot on each force platform. The test leader encouraged the athlete to drop from the box and not jump down from it. If a reflex-marker was lost during the data-collection the jump was not approved.

Of the five approved jumps, the last three recordings with acceptable tracking were used. Acceptable tracking was defined as no disappearing of markers for more than 10 following frames of the recording. A technician controlling the computer quickly looked through the recording to accept the tracking (figure 4.2).



**Figure 4.2:** Caption of the computer screen after the recording of a VDJ.

The reflex-camera system required the marker to be visible to at least two of the eight cameras, to be positioned in the coordinate system.

This was an issue especially for the pelvis-markers, as they disappeared from the reflex cameras at the reverse-point (amortization phase) of the jump, because of great hip flexion. Some athletes were given illiac crest-markers, bilaterally, to compensate for the hidden pelvis-marker. The athletes then performed the VDJ with target. A lightweight bar (Swix C3 skipole), attached to a high-jump measuring pole, was placed horizontally as an overhead target (Figure 4.3).



**Figure 4.3:** *The vertical drop-jump. In this project the drop down from box/stool is 30cm, the athletes has placed a foot on each force platform and will try to reach for the overhead target with the head.*

The athletes were required to reach the target with the top of their head. Test leaders adjusted the height of the bar, so the first height was reasonable, and it was raised

when touched. The athlete moved on to a new height when she succeeded in reaching the bar, but was finished when failing to reach it in two succeeding attempts. The athlete was expected to reach maximum height within ten jumps.

#### **4.5 Data Collection**

Since this study used subjects from the same cohort as Kristianslund & Krosshaug (2013), the data collection is the same.

In brief, however, an eight camera, infrared optical tracking system (ProReflex, Qualisys Inc., Gothenburg, Sweden) recorded the motion with a frame rate of 240 Hz. Two Amti force plates (AMTI LG6-4-1, Watertown, MA 02472, USA) measured the ground reaction forces and center of pressure at a frame rate of 960 Hz. The marker trajectories were tracked and calculated in Qualisys Track Manager (Qualisys Inc., Gothenburg, Sweden). The contact phase was defined as unfiltered vertical ground reaction force exceeding 20 N. Marker signal smoothing and interpolation was done using the generalized cross validation package in the cubic mode (Woltring, 1986). An interpolating quintic spline was then fitted to the data points, and first and second order derivatives were subsequently obtained from this higher order spline. If one pelvic marker was missing for more than 20 frames, an interpolation established on the position of three other pelvic markers was used. The cut-off frequency for force and marker data was set to 15 Hz.

Joint kinetics and kinematics were calculated using a standard 3D iterative Newton/Euler method (Bresler & Frankel, 1950). An optimization procedure involving singular value decomposition (Söderkvist & Wedin, 1993) was utilized to derive the segment embedded reference frames for the thigh and shank. The joint centers of the ankle and knee were determined by former methods (Davis, Ounpuu, Tyburski, & Gage, 1991), and the ankle joint centre location was defined according to (Eng and Winter, 1995). The hip joint centers were estimated similar to (Bell, Pedersen, & Brand, 1990). The attitude angle convention was used to estimate relevant clinical angles for the hip, knee and ankle joint (Woltring, 1994). The inertia parameters were estimated based on 46 measures of segment heights, perimeters and widths (Yeadon, 1990). This method divides the body into an eleven-segment model, where each segment consists of subsegments, where the different density of each

segment is taken into account. For each segment the mass and location of mass centre are calculated and the total centre of mass (COM) can be determined. In this project COM was used to determine the jump height of the athlete. The height of COM at the static recording deducted from the COM at the top of the jump was the jump height of the athlete, i.e. displacement of COM.

All calculations were done using Matlab® (MathWorks Inc., Natick, MA, USA).

Of the three jumps with acceptable tracking with and without overhead target, the one with the best performance (jump height), during each condition, was reported.

We investigated hip, knee and ankle joint kinetics and kinematics. The position of COM of the body and of the torso, relative to knee and foot placement, was correlated with lower extremity joint moments.

#### **4.6 Ethical considerations**

Since this was a part of a bigger project, it has already been reviewed and approved by the *Regional Committee for Medical Research Ethics* and reported to *Norwegian Social Science Data Services*.

The risk of getting an injury in this project was equal to or smaller than the risk of injury in the regular training practice the players attend. A study has shown that the risk of injury is significantly smaller in training practice compared with match situations, about  $\frac{3}{4}$  of all ACL injuries occurred during matches (Smith et al., 2011b). Squat-jumps are a well-known exercise for team handball and football players, and the other tests are also regularly performed in elite team handball.

#### **4.7 Statistics**

In SPSS (PASW Statistics 18.0.2), the data was checked for normality using a Shapiro-Wilks Test. Paired T-Tests were used for analyzing differences between jump and target-jump situations and side-to-side differences for normal data. For data that violated the assumption of normality (displayed in table 4.2), the Wilcoxon's Matched Pairs Signed Ranks Test was used. When comparing the two groups of athletes,

football and handball players, with each other, the Independent T-Test was used for normal data, and the Mann-Whitney U Test was used for data violating the assumption of normality. Microsoft Excel was used to produce graphs and tables. Bland-Altman plots were produced in MedCalc v12.2.1.

**Table 4.2:** *The data violating assumption of normality during non-target and target conditions.*

<b>Data violating assumption of normality</b>	<b>Non-Target Condition</b>	<b>Target Condition</b>
<b>Wilcoxon's Matched Pairs Signed Ranks Test</b>		
<i>Range of Motion</i>		
-Right Knee Flexion	x	
-Left Knee Flexion	x	
-Right FPPA	x	x
-Left FPPA	x	x
-Right Hip Flexion	x	x
-Left Hip Flexion	x	x
-Right Ankle Flexion	x	
<i>Moments</i>		
-Right Knee Valgus		x
-Right Knee Flexion	x	
-Right Ankle Flexion	x	x
<b>Mann-Whitney's U Test</b>		
<i>Contact Time</i>		
-Football	x	x
-Handball	x	x



## 5 Results

### 5.1 Dropout

Of the seventy-two athletes tested in this study, one football player was missing a back marker during the recording, which prevented a complete description of the COM. This led to error in the data, and the player had to be excluded. That brought the total to 71 athletes, 36 handball players and 35 football players.

### 5.2 Jump Height

On average all athletes increased their jump height significantly from 41.2 cm in Non-target-VDJ to 43.8 cm in Target-VDJ, the increase was 2.6 cm (95% CI: 1.8, 3.4;  $p < 0.01$ ), which is a 6.3 % increase (table 5.1). The handball players displayed an increase from 41.6 cm to 44.8 cm (95% CI: 2.0, 4.4;  $p < 0.01$ ) across the two jumping conditions, which was a 7.7 % improvement. The football players showed a 4.9 % improvement as they altered jump-height from 40.8 cm to 42.8 cm (95% CI: 1.2, 2.8;  $p < 0.01$ ). The handball players also jumped significantly higher than the football players in target conditions (4.6 %,  $p = 0.04$ ).

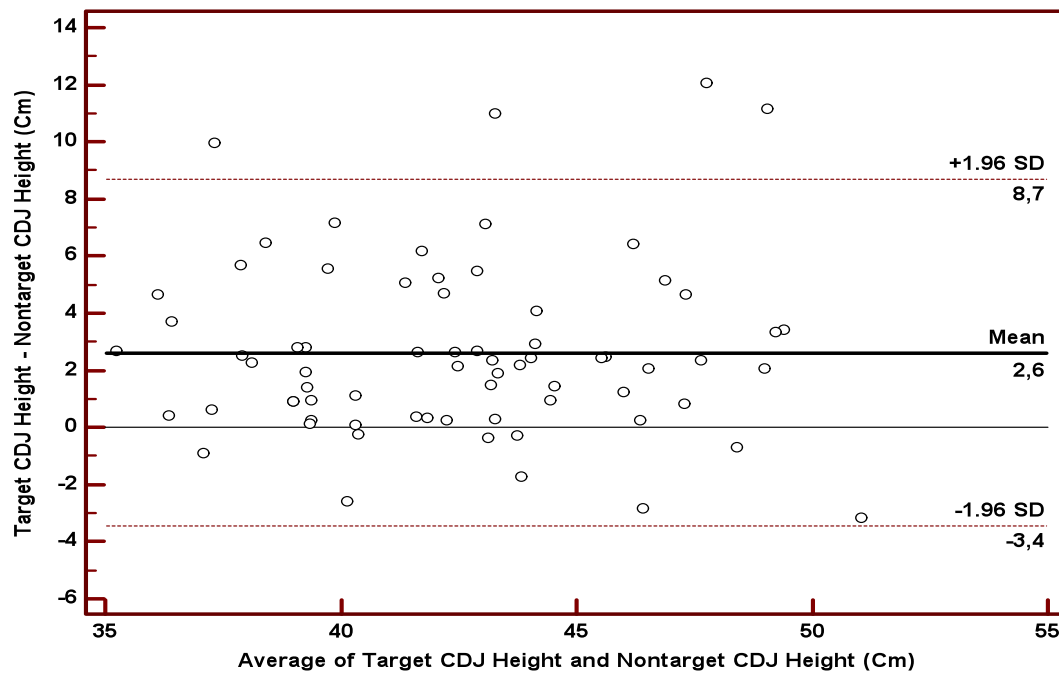
**Table 5.1:** Results of the performance during Non-target VDJ conditions and Target VDJ conditions.

Performance	Non-target VDJ	Target VDJ	Difference & CI
Jump Height (cm)	41.2 $\pm$ 0.5	43.8 $\pm$ 0.5**	2.6 (1.8 – 3.4)
- Handball	41.6 $\pm$ 0.7	44.8 $\pm$ 0.8** <sup>b</sup>	3.2 (2.0 – 4.4)
- Football	40.8 $\pm$ 0.7	42.8 $\pm$ 0.6**	2.0 (1.2 – 2.8)

Values are mean  $\pm$  SEM, Handball n = 36, Football n = 35, All n = 71, \*\* Significant target difference at  $p < 0.01$ , <sup>b</sup> Significant sport difference at  $p < 0.05$

In the Bland-Altman plot (figure 5.1) the first parameter explains the difference in jump height of the two jumps (Non-target and Target VDJ) and the second parameter is the average jump height of the two jumps. The plot shows an even spread and the mean difference is 2.6 cm, and the 95% limits of agreement runs from -3.4 cm to 8.7 cm. Ninety-five percent of the athletes are located between the limits of agreement,

four athletes (5.6 %) lie above the limits of agreement. The plot also shows that one seventh of the athletes (10) had no positive effect in using an overhead-target, while the rest (61 athletes) altered their jump height positively. Thus 85% of the athletes increased their performance when jumping for an overhead target. The greatest change was 12 cm from Non-target VDJ to Target VDJ.

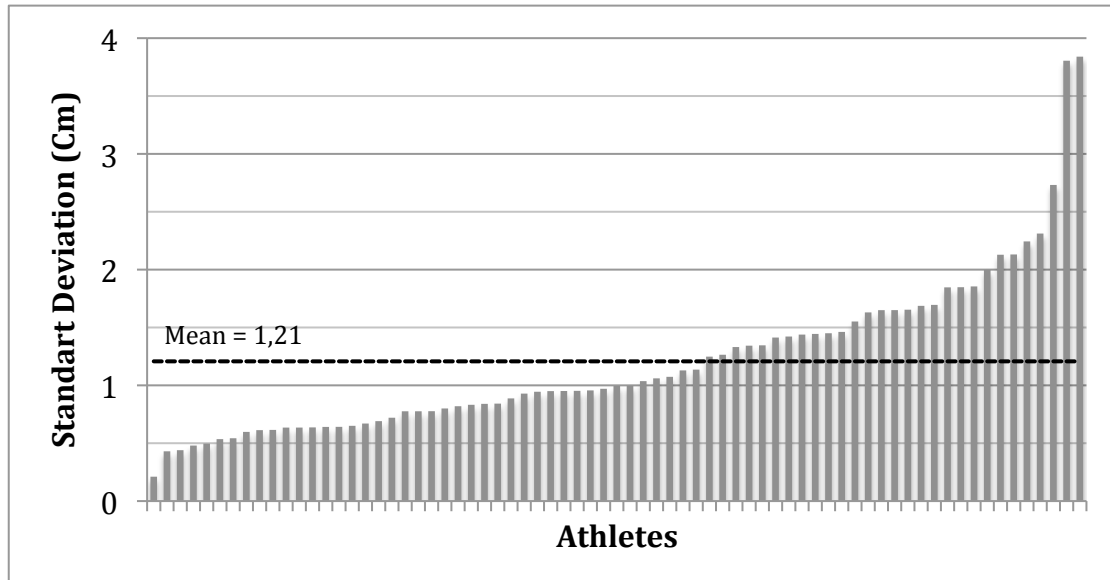


**Figure 5.1:** Bland-Altman Plot of Jump Height. The average of the Non-target and Target VDJ are displayed on the X-axis. The difference of the Non-target and Target VDJ are displayed on the Y-axis. The full line indicates the mean difference and the dotted lines indicate 95% limits of agreement.  $N = 71$ , Mean = 2.6,  $SD \pm 1.96 = 8.7$  cm & -3.4 cm.

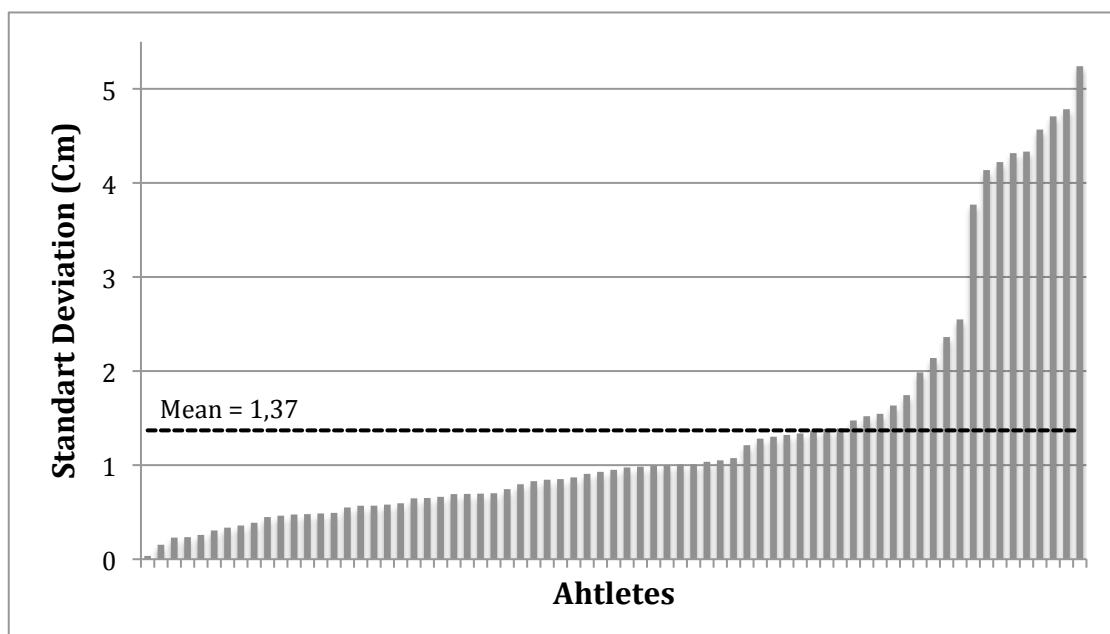
### 5.3 Variation between jumps

The players had three accepted jumps for each of the two VDJ conditions. To get an overview of the variation in the three jumps, the Standard Deviation for each individual athlete is presented in figure 5.2 and 5.3. The Standard Deviation for the three Non-target VDJs varies between 0.2 cm and 3.8 cm, with a mean SD of 1.21 cm for all of the athletes. For the three Target VDJs the SD ranges between 0.0 cm and 5.2 cm and has a mean of 1.37 cm for all athletes. This is a 13.2 % increase in the SD and a greater range in target-situations compared with non-target situations. The variation-pattern is very similar for the two conditions, however the for Target VDJ

there is an increased number of athletes with a greater SD. Only two players showed variation over 3 cm during Non-target VDJs, but nine players had this variation in the Target-VDJs.



**Figure 5.2:** The Standard Deviation for each athlete across the three accepted Non-target VDJ. N=71



**Figure 5.3:** The Standard Deviation for each athlete across the three accepted Target VDJ. N=71

## 5.4 Knee Control

### 5.4.1 Valgus

There was only a small insignificant change from Non-target to Target situations, in the valgus knee angles at IC. As displayed in table 5.2, the same was applicable for the comparison of the right and left side at IC. There was a significant increase of maximum knee valgus during the contact phase of the Target VDJ. There was a 0,5° increase for both the right and left side (95% CI: 0.1, 0.9;  $p = 0.02$ ) (see table 5.3).

**Table 5.2:** Initial Contact angles of the knee joint during Non-target VDJ and Target VDJ.

Initial Contact Angle			Non-target VDJ	Target VDJ	Difference & CI
<b>Valgus angle</b>	Knee valgus (°)	Right	2.6 ± 0.4 & 3.2	2.8 ± 0.4 & 3.6	0.2 (-0.4 – 0.8)
		Left	2.4 ± 0.5 & 4.0	2.7 ± 0.5 & 4.2	0.2 (-0.2 – 0.6)
	FPPA (°)	Right	-0.5 ± 0.5 & 4.1	-0.4 ± 0.6 & 4.7	0.0 (-0.8 – 0.8)
		Left	-1.6 ± 0.5 & 4.4	-1.6 ± 0.6 & 4.7	0.0 (-0.6 – 0.6)

Values are mean ± SEM & SD for Non-target and Target VDJ and mean & Confidence Interval for Difference,  $n = 71$ ,

**Table 5.3:** Maximum angles of the knee joint during Non-target VDJ and Target VDJ.

Maximum Angle			Non-target VDJ	Target VDJ	Difference & CI
<b>Valgus angle</b>	Knee valgus (°)	Right	9.9 ± 0.6 & 4.7	10.4 ± 0.6 & 4.7**	0.5 (0.1 – 0.9)
		Left	10.8 ± 0.7 & 5.9	11.3 ± 0.7 & 5.9*	0.5 (0.1 – 0.9)
	FPPA (°)	Right	6.2 ± 0.7 & 6.3	8.3 ± 0.9 & 7.3**	2.2 (0.8 – 3.6)
		Left	6.8 ± 0.9 & 7.2	7.7 ± 1.0 & 8.1	0.9 (-0.5 – 2.3)

Values are mean ± SEM & SD for Non-target and Target VDJ and mean & Confidence Interval for Difference,  $n = 71$ , \* Target differences at  $p < 0.05$ , \*\* Target difference at  $p < 0.01$

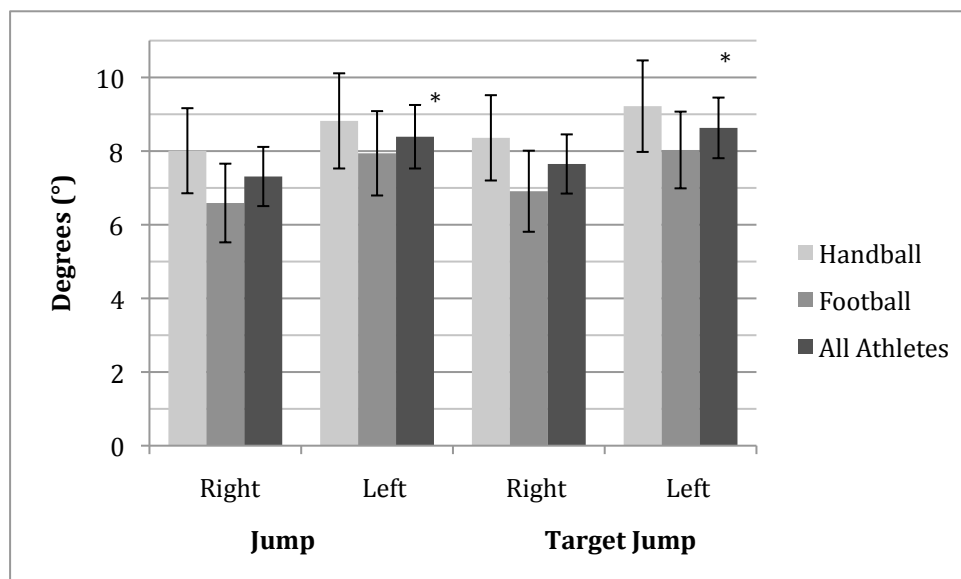
Despite the increase in maximum valgus angle there were no significant differences in the valgus ROM between the two VDJ conditions. However a significant side-to-side difference in both non-target situations and target situations for the knee valgus ROM was found. The athletes had about one degree greater valgus in the left knee, 7.3° vs. 8.4° (95% CI: 0.3, 1.9;  $p = 0.02$ ) and 7.7° vs. 8.6° (95% CI: 0.2, 1.8;  $p < 0.027$ ), as shown in table 5.4.

There were no significant differences between the two sports, in terms of valgus ROM. Handball players seemed to always have slightly greater valgus than the football players, and showed a tendency to increase valgus ROM in the right knee in both jump situations ( $p = 0.08$ ), compared to the football players (figure 5.4).

**Table 5.4:** Range of Motion of the knee joint during Non-target VDJ and Target VDJ.

Range of Motion			Non-target VDJ	Target VDJ	Difference & CI
<b>Valgus angle</b>	Knee valgus (°)	Right	$7.3 \pm 0.4$ & $3.4$	$7.7 \pm 0.4$ & $3.5$	$0.3$ ( $-0.2 - 1.0$ )
		Left	$8.4 \pm 0.4$ & $3.7^a$	$8.6 \pm 0.4$ & $3.5^a$	$0.2$ ( $-0.2 - 0.6$ )
	FPPA (°)	Right	$6.7 \pm 0.5$ & $4.3$	$8.8 \pm 0.7$ & $5.7^{**}$	$2.1$ ( $0.9 - 3.4$ )
		Left	$8.3 \pm 0.7$ & $5.7^a$	$9.3 \pm 0.7$ & $5.8$	$0.9$ ( $-0.2 - 2.2$ )

Values are mean  $\pm$  SEM & SD for Non-target and Target VDJ and mean & Confidence Interval for Difference,  $n = 71$ , <sup>a</sup> Side to side differences at  $p < 0.05$ , <sup>\*\*</sup> Target difference at  $p < 0.01$



**Figure 5.4:** The mean valgus ROM for the whole group, as well as for the different sports. The error bars indicates the 95% CI. Handball  $n = 36$ , Football  $n = 35$ , All Athletes  $n = 71$ , \* Side-to-side differences at  $p < 0.05$ .

## 5.4.2 FPPA

At IC there was neither changes in the FPPA between the two jumping conditions nor any side-to-side differences despite showing a tendency ( $p = 0.11$ ). The maximum FPPA however increased for both the right and left side during Target VDJs, for the right side the significant increase was  $2.2^\circ$  (95% CI:  $0.8, 3.6$ ;  $p < 0.01$ ) and for the left

side the increase was  $0.9^{\circ}$  ( $-0.5, 2.3$ ;  $p = 0.17$ ). Therefore there was a significant increase of  $2.1^{\circ}$  (95% CI:  $0.9, 3.4$ ;  $p < 0.01$ ) in the FPPA ROM in the right knee from non-target to target situations. For the left knee, there was also an increase in FPPA ROM, it was however not statistical significant (see table 5.4). For the FPPA ROM there was side-to-side differences during the Non-target VDJs, the angle was  $1.7^{\circ}$  (95% CI:  $0.1, 3.3$ ;  $p = 0.03$ ) greater on the left side, the side-to-side differences during the Target VDJs was not statistically significant.

The Knee Valgus and FPPA methods presented relatively even measurements of the knee abduction ROM angle. Both methods showed side-to-side differences in Non-target VDJs, but in Target VDJs it was only the valgus measurements that showed a significant side-to-side difference. The FPPA measurement however revealed a target difference in the right knee abduction ROM angle. The variation for the FPPA measurements was however greater compared to the Valgus measurements,  $SD_{\text{Valgus}} = 3.5^{\circ}$ ,  $SD_{\text{FPPA}} = 5.4^{\circ}$ .

### 5.4.3 Moments

Hip flexion moments decreased significantly with  $0.12 \text{ Nm/kg}$  (95% CI:  $0.02, 0.22$ ;  $p = 0.02$ ) on the right side during target conditions. On the left side the hip flexion moments increased, but however not significantly ( $p > 0.2$ ) (see table 5.5). The moment on the left side were significantly greater than on the right side in target conditions ( $p = 0.03$ ).

Knee valgus moments increased from No-target to Target conditions. There was a  $0.05 \text{ Nm/kg}$  significant increase on the right side (95% CI:  $0.01, 0.09$ ;  $p < 0.01$ ), and a  $0.04 \text{ Nm/kg}$  significant increase (95% CI:  $0.00, 0.08$ ;  $p = 0.047$ ) on the left side. The valgus moments on the right side were also significantly greater than on the left side in Overhead-target situations ( $p = 0.035$ ).

**Table 5.5:** The moments for the joints in the lower extremities. The moments were adjusted for the athletes' bodymass.

Moments		Non-target VDJ		Target VDJ	Difference & CI
<b>Hip</b>	Flexion (Nm/kg)	Right	2.97 ± 0.05	2.85 ± 0.06*	0.12 (0.02 – 0.22)
		Left	2.98 ± 0.07	3.01 ± 0.08 <sup>a</sup>	0.03 (-0.07 – 0.13)
<b>Knee</b>	Valgus (Nm/kg)	Right	0.41 ± 0.02	0.45 ± 0.02 <sup>a</sup> **	0.05 (0.01 – 0.09)
		Left	0.37 ± 0.02	0.40 ± 0.02*	0.04 (0.00 – 0.08)
	Flexion (Nm/kg)	Right	2.49 ± 0.06	2.65 ± 0.07**	0.16 (-0.62 – 0.94)
		Left	2.55 ± 0.07	2.70 ± 0.07**	0.15 (-0.44 – 0.74)
<b>Ankle</b>	Flexion (Nm/kg)	Right	2.06 ± 0.06	2.19 ± 0.07**	0.13 (-0.65 – 0.91)
		Left	2.05 ± 0.06	2.13 ± 0.06**	0.09 (-0.50 – 0.68)

Values are mean ± SEM, n = 71, <sup>a</sup> Side to side differences at p < 0.05, \*\* Target difference at p < 0.01,

\* Target difference at p < 0.05

For both knee and ankle flexion there was a significant increase in the target situations for both the right and the left side. The knee flexion moments increased in target situations with 6.4% and 5.9% for the right and left side respectively. The ankle flexion moments also increased in the target conditions, the increase was 6.3 % on the right side, and 3.9% the left side (A closer description of results are shown in table 5.5)

## 5.5 Jump Technique

### 5.5.1 Joint Angles

Table 5.6 displays that the athletes had no significant changes in IC hip and ankle flexion angle from Non-target situation to Target situations. There was a significant increase in both right and left IC knee flexion angle. Even though the flexion angle was on the left side, the increase in both knees was 2.2° (95% CI: 1.0, 3.4; p < 0.01).

**Table 5.6:** Initial contact flexion angles in the lower extremity during the two VDJ-conditions.

Initial Contact Angle			Non-target VDJ	Target VDJ	Difference & CI
Flexion angle	Hip (°)	Right	49.4 ± 1.0	49.8 ± 1.0	0.4 (-1.2 – 2.0)
		Left	50.7 ± 1.1	51.0 ± 1.0	0.3 (-1.1 – 1.7)
	Knee (°)	Right	35.6 ± 0.8	37.8 ± 0.8**	2.2 (1.0 – 3.4)
		Left	36.5 ± 0.9	38.7 ± 0.9**	2.2 (1.0 – 3.4)
	Ankle (°)	Right	-12.1 ± 0.8	-12.5 ± 0.8	0.4 (-0.8 – 1.6)
		Left	-12.9 ± 0.8	-12.6 ± 0.7	0.3 (-0.7 – 1.3)

Values are mean ± SEM, n = 71, \*\* Target difference at p < 0.01

When analyzing the maximum flexion angles during the VDJs, there is a significant decrease in hip and knee flexion angle (table 5.7), this means the athletes do not go as deep in Target VDJ situations as in Non-target VDJ situations. The hip flexion decrease was 4.2° (95% CI: 2.6, 5.8; p < 0.01) for the right side, and 4.3° (95% CI: 2.7, 5.9; p < 0.01) for the left side. The knee flexion angle decreased 3.0° (95% CI: 1.4, 4.6; p < 0.01) on the right side, and 3.2° (95% CI: 1.6, 4.8; p < 0.01) on the left side. In the ankle there were no significant changes in the flexion angles.

**Table 5.7:** Maximum flexion angles in the lower extremity during the two VDJ-conditions.

Maximum Angle			Non-target VDJ	Target VDJ	Difference & CI
Flexion angle	Hip (°)	Right	76.1 ± 1.7	71.9 ± 1.6**	4.2 (2.6 – 5.8)
		Left	77.4 ± 1.8	73.1 ± 1.7**	4.3 (2.7 – 5.9)
	Knee (°)	Right	89.5 ± 1.5	86.5 ± 1.2**	3.0 (1.4 – 4.6)
		Left	90.2 ± 1.5	87.0 ± 1.2**	3.2 (1.6 – 4.8)
	Ankle (°)	Right	35.5 ± 0.6	35.2 ± 0.6	0.4 (-0.2 – 1.0)
		Left	35.0 ± 0.4	34.5 ± 0.5	0.5 (-0.1 – 1.1)

Values are mean ± SEM, n = 71, \*\* Target difference at p < 0.01

## 5.5.2 Contact Time

Contact time was greater in non-target conditions compared to target conditions. The athletes reduced the contact time from 0.39 seconds during Non-target-VDJs to 0.36 Sec (95% CI: 0.01, 0.05; p < 0.01). The calculations showed no significant differences between the two groups of athletes (table 5.8), but they both individually



decreased the contact time significantly, during target conditions, with 0.03 sec and 0.04 sec for the handball and football players respectively (thoroughly description in table 5.8).

**Table 5.8:** *The athletes contact time with the force platforms before toe-off.*

<b>Contact Time</b>	<b>Non-target VDJ</b>	<b>Target VDJ</b>	<b>Difference &amp; CI</b>
Contact Time (Sec)	0.39 ± 0.01	0.36 ± 0.01**	0.03 (0.01 – 0.05)
- Handball	0.38 ± 0.02	0.35 ± 0.02**	0.03 (0.01 – 0.05)
- Football	0.41 ± 0.02	0.37 ± 0.02**	0.04 (0.02 – 0.06)

Values are mean ± SEM, Handball n = 36, Football n = 35, All n= 71, \*\* Significant target difference at  $p < 0.01$ .

## 6 Discussion

One of the main goals of this study was to investigate the effect of an overhead target on performance in a VDJ in female elite athletes. The hypothesis was confirmed as the athletes on average increased the jump height significantly by  $2.6 \pm 0.4$  cm during target conditions.

The other purpose of the study was to investigate if greater knee valgus could be provoked by the presence of an overhead target, in athletes performing a VDJ. The athletes did however not show any significant increase in knee valgus ROM in target situations. The maximum knee valgus angle increased by  $0.5 \pm 0.2^\circ$  in target situations. This was a statistically significant increase, but the clinical relevance is likely insignificant.

In future studies and tests, the use of overhead target could be implemented, as it seems to trigger the extrinsic motivation and therefore increases the performance. However, the overhead target does not seem to have any notable effect on the dynamic knee stability and therefore the use of it seems to have no relevance in this matter.

### 6.1 *Jump Height*

The difference in jump height found in this study was greater than the one found in an earlier study (Ford et al., 2005), where the female football players altered their jump height from  $36.8 \pm 1.1$  cm to  $37.3 \pm 1.0$  cm during the overhead target condition. Maturation and the competition level of the athletes can probably explain the higher average and increase in this study. Elite football and handball players should on an average be more matured, and have a greater amount of total training hours than collegiate football players.

The handball players had a greater jump height, compared to football players in both Non-target VDJ and Target VDJ conditions, as well as a greater increase. The fact that handball players jumped higher than the football players can probably be explained by the requirements of the two sports. In handball jump shots and jump blocks are present in a game in almost every attack, whereas jumping in football

games occurs only a fraction of the times during aerial duels. So the ability to perform jumps near maximum effort is required more often in handball than in football.

The Bland-Altman plot was used to see if there was an agreement between the difference of the two jumping conditions ( $\Delta$ -jump height), and the average of the two jumps. It seems feasible to think that athletes with low average jump height would have the greatest potential to increase jump height, because less motivated subjects and subject with sub-optimal jump technique will have poorer performance on the non-target VDJ. Thus having greater advantage of the extrinsic motivation (overhead goal) and the repetitions of the VDJs. The plot however showed an even spread, which means that the gain in jump height between tasks ( $\Delta$ -jump height) is not explained by the athletes average jump height. Thus athletes with greater average are not more likely to increase their performance when reaching for an overhead target, than athletes with lower average jump height. The athletes alter their jump height to more or less the same extent, independent of their average jump height (true value).

The variation of the three approved jump for each condition was showed in figure 5 and 6. In the Non-target conditions the mean variation was 1.21 cm. The main part of the athletes had a variation ranging from 0-2.5 cm, with three cases above that range. There was a similar pattern for the Target conditions, however a greater amount of athletes were above that range, and with a greater variation (up to 5.2 cm). This is however not surprising, because we raised the target with one or two centimeters each time the player obtained an approved jump. Thus some of the athletes could have reached the target without maximal effort.

## **6.2 Valgus Motion**

Previously both 3D-valgus measurements and FPPA has been used to investigate knee abduction angles, and both measurement methods were included in this study for better comparison. The athletes did not show any notable change in initial contact knee valgus angles during target conditions, but they showed a significant  $0.5^\circ$  increase in maximum knee valgus angle for both legs, with  $p < 0.01$  for the right side and  $p = 0.02$  for the left side. Despite the change in maximum knee valgus angle there was only a slight, insignificant, change in the knee valgus ROM during target

conditions. The magnitude of maximum knee valgus angle found in this study was only one third of what female athletes earlier have displayed (Ford et al., 2003). A possible explanation may be because athletes in our study was more matured and experienced compared to the female athletes in the study by Ford et al., (2003). However Hewett et al., (2005) found clear differences in valgus angles between healthy athletes and athletes who went on to rupture their ACL. The injured athletes displayed a mean of  $9^{\circ}$  valgus angle in VDJs, which was a bit less than the valgus angles found in our study. In a recent study handball players displayed maximum valgus angles of  $5.6^{\circ}$  and  $11.5^{\circ}$  during VDJ and sidestep cutting respectively (Kristianslund & Krosshaug, 2013). Thus athletes in our study have valgus angles more similar to the ones seen in sidestep cutting. Recently athletes have, through MBIM, displayed that knee valgus angle increased  $12^{\circ}$  from neutral position, 40 mSec after IC in ACL injury situations (Koga et al., 2010), and valgus angles in team sports of  $14-15^{\circ}$  (Krosshaug, Slauterbeck, Engebretsen, & Bahr, 2007). Thus the valgus angles in our study are the about same as angles exhibited in future ACL injured athletes during VDJ, sidestep cutting motions and in ACL injury situations where athletes were performing cutting motions or one-legged landings. This should in turn mean that athletes performing the VDJ in our study are at great risk of an ACL injury when tested. However there also seems to be another predictor of ACL injury risk, as we shall discuss later on.

In a study by Ford et al., (2003) they looked at side-to-side differences and found much greater knee valgus angle in the dominant leg ( $15.1^{\circ}$ ,  $p < 0,001$ ), which was more than twice as great valgus angle. A possible mechanism may be athletes (over-) relying on their “strong” leg, and thus the valgus angle is greater on the dominant side. In our study the athletes displayed significant side-to-side differences ( $p < 0.05$ ) by  $1.1^{\circ}$  for Non-target VDJs and  $0.9^{\circ}$  for Target VDJs. For both situations the greatest ROM angles were found in the left knee. We only differentiate between right and left leg, and the small differences in our study may be because of decreased strength or ability of activation in the muscles surrounding the left knees of the athletes. The maturation of the athletes in our study and probably a longer background from strength- and neuromuscular training may have diminished the imbalance between the legs, and subsequently reduced the risk of sustaining a leg dominant ACL injury (Myer et al., 2011b).

The measurement of the maximum FPPA showed a significant increase ( $p < 0.01$ ) in the right knee, in target situations, but not for the left knee. The FPPA ROM showed the same results, as there was a significant increase ( $p < 0.01$ ) in target situations for the right knee. The FPPA ROM also displayed a significant ( $p = 0.03$ ) side-to-side difference in non-target situations, but not in target situations. The FPPA data is fortunately quite similar to the 3D-datas, with the same changes, but the FPPA however shows greater variation in the measurements. These differences in variation will be discussed later on.

One can say that our hypothesis is supported as the maximum knee valgus angle increased during target conditions. However the change in maximum valgus angle of  $0.5^\circ$ , which is about a 4-5% difference, is not a large enough response for us to call it a clinical relevant change, as we would have liked over 8% difference to manifest it as clinical relevant. During sport specific sidestep cutting, a reduction of about  $4-5^\circ$  in knee valgus angle has seen to be followed by a 19% decrease in knee valgus moments (Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2012b). So the use of a target only seems to have a small influence on the dynamic knee control in a VDJ.

### **6.3 Moments**

The players increased their knee valgus moments significantly, with about 0.05 Nm/kg in target conditions. The knee valgus moments seen earlier in injured athletes were almost twice the size, compared to those in our study (Hewett, et al., 2005). In our study the valgus moments were greater on the right side during both conditions, thus other parameters than just the valgus angle must decide the knee valgus moments, since valgus angles were greater on the left side. The study by Kristianslund et al., (2012b) lists parameters that affects the valgus moment in sidestep cutting, some of them applicable to VDJ, but not measured in our study, and some of them only applicable to the motion performed in that study (sidestep cutting). Changes in the knee valgus angle have the biggest impact on the valgus moment, but athletes in this study may change other (unmeasured) parameters during target conditions, hence the greater valgus angles on the left side but greater valgus moments on the right side. Parameters such as hip internal rotation, hip abduction and

lateral torso flexion may have a negative effect (increase) on valgus moments, whereas toe landing could decrease the moments (Kristianslund et al., 2012b).

Despite the significant change, the magnitude of the valgus moments found in this study seems to be negligible, as a recent study, based on players from the same cohort, measured knee valgus moments up to 200% greater than seen in our study (Kristianslund et al., 2012a). The athletes were tested in a more game like situation, performing a one-legged cut motion. Hence handball and football players can resist much higher knee valgus moments during a game compared to the moments measured in a VDJ. Thus, the magnitude and difference in valgus moments cannot be considered to be clinical relevant.

Even though our athletes demonstrated valgus angles that could be linked to increased ACL injury risk, the minor valgus moments found in our study indicates that the athletes can and do resist much greater external forces, thus other more game like motions, than the VDJ, may be better ways to challenge the dynamic knee stability.

## **6.4 Technique**

### **6.4.1 Joint Angles**

The athletes had a significant change in both hip and knee flexion ROM. The hip flexion ROM was changed with  $-4.6^{\circ}$ , and the knee flexion ROM was changed by  $-5.3^{\circ}$ . The reduction in the hip ROM was caused by a significant decrease in the maximum hip angle and no change of initial contact angle. The knee ROM decrease was caused by a significant increase of initial contact angle and a significant reduction of the maximum angle. Thus the athletes did not go as deep in the Target situations as they did during the Non-target VDJ.

The study by Ford et al. (2005) found a small increase in maximum knee flexion angles for the female athletes, using an overhead target. Similar to this study, Smith et al. (2011b) found a decrease in knee flexion angle during hurdle conditions, meaning a more upright jump. The same changes were however absent in the overhead target (Vertec) conditions (Smith et al., 2011b). In our study the athletes had less hip flexion

during target conditions, which has been seen in hurdle conditions, but not for overhead target conditions (Ford et al., 2005, Smith et al., 2011b).

The findings in our study are comparable to those obtained for male athletes in the study of Ford et al. (2005) and the hurdle results in the study of Smith et al. (2011b).

The athletes use a more upright jump technique to achieve greater jump height, by decreasing the ROM flexion angles in both the hip and knee joints.

The vertec is a different type of overhead target, compared to the study by Ford et al. (2005) and our study. The height of the overhead target can be raised, but the vertec will present a result no matter the jump height and therefore not pushing the athletes in the same way. This type of extrinsic motivation can be what is driving the athletes to reach the target and provoke the differences in jump height and therefore an overhead target could beneficially be implemented in training and testing of VDJ as it most likely will affect performance positively.

#### **6.4.2 Contact Time**

The athletes executed the jumps quicker in target situations, explained by a statistical significant change in contact time. This is in agreement with other studies, where the male athletes had a significant decrease in stance time and a concurrent increase in jump height (Ford et al., 2005). Another study showed similar results using a hurdle (Smith et al., 2011b), the athletes increased the ground reaction force as well as decreasing the stance time from. These “new” findings are in disagreement with the results from earlier studies, where a decrease in contact time has been closely related to a reduction of jump height and vertical velocity at takeoff (Bobbert et al., 1987; Walsh, Arampatzis, Schade, & Brüggemann, 2004). For the female athletes in the study of Ford et al. (2005) there was no change in stance time despite increasing their jump height significantly. The athletes executed the jumps during the target situations using a more upright technique, with less flexion in hips and knees. Thus they had shorter contact time, and probably better utilization of the SSC, due to a more bounce like jumping technique (Horita, Komi, Nicol, & Kyröläinen, 2002). This could be a reason for the subsequent improvement in performance. Even though our study found a decrease in contact time with greater jump height, it is difficult to say if this is a direct reason for the improvement of performance.

## **6.5 3D vs. 2D screening**

The 3D-measurement is recognized as “the gold standard” of screening athletes for neuromuscular deficits (McLean et al., 2005; Stensrud et al., 2011). When we compared the two measurement methods we found greater variation in the FPPA-measurements despite relatively equal valgus-values. The greater variation in FPPA-measurements can probably be explained by the method of capturing data. The FPPA-method uses a “locked” two-dimensional coordinate system, which views the athlete from the front. This means that if the athlete performs a hip internal rotation, or is not precisely situated perpendicular to the coordinate system, any flexion of the knee could be perceived as knee valgus. This “crosstalk” is probably the explanation of the greater variation in FPPA measurements. The three-dimensional measuring method has its own coordinate system for the thigh in relation to the leg based on the static pre-measurement in anatomical position. This will diminish, if not eliminate, any crosstalk between knee valgus and knee flexion.

The study by Hewett et al., (2005) saw that both valgus angles and moments were good predictors of future ACL-injuries, however recently researchers revealed poor correlation between valgus angles in VDJ and valgus motions in sidestep cutting (Kristianslund & Krosshaug, 2013). Our findings also indicate that the valgus motion can be near injury angles (Krosshaug et al., 2007, Koga et al., 2010), without notable valgus moments. Consequently these results can obstruct the use of video camera or subjective assessment in VDJ, as external knee moments are hard to assess using a video camera, and not comparable to sidestep cutting motions, where ACL injures often occurs (Kristianslund & Krosshaug, 2013).

## **6.6 Handball vs. Football players**

The reason for comparison between handball and football players, was to see if the two groups of athletes could be recognized as one cohort, or if they exhibited differences on a scale, that would indicate that we had to recognize them as two cohorts.



## **6.7 Jump Height Method**

There is no consensus in measuring jump height during VDJ and other jumping tasks in science and training. Some studies measure the performance as the Vertical velocity at TO or the peak vGRF (Chappell & Limpisvasti, 2008; Smith et al., 2011b), whilst others express the performance in centimeters/meters (Bobbert et al., 1987; Ford et al., 2005; Ford et al., 2007; Rønnestad et al., 2008). Most methods include measuring displacement of some sort of a mark, in the study by Ford et al., (2005) jump height was measured as displacement of the right and left trochanter marker, from standing position to maximum height. Rønnestad et al., (2008) measured displacement of COM from force developed and the body mass of the athlete. What seems to be the key in choosing the right method will be the velocity of the movement. If an athlete is starting from a static start position, jump height measured from the force platform will be a good method. Because measurement of the height and speed of COM at IC is needed in a VDJ, the measurement from the force platforms will be insecure, thus displacement of COM will exhibit a more stable measurement. From Newton's 3<sup>rd</sup> law we know there is a direct mechanic relationship between the forces an athlete exerts into the ground and the height the COM will attain. The study by Ford et al., (2007) used a method similar to the one used in our study to determine jump height in overhead target VDJ and found excellent within-session and between-session reliability.

## **6.8 Relevance for Lower Level Athletes**

Even though this project is analyzing a large amount of players (approximately 700), compared to other studies, there can be difficulties in generalizing the results because athletes are absolute elite handball and football players. Participation in handball and football has increased in Norway over the last 9 years with 64.637 to 72.051 and 85.680 to 105.631 respectively (Norges Idrettsforbund, Årsrapport 2002, 2003) (Norges Idrettsforbund, Årsrapport 2011, 2012), i.e. more players are participating in lower levels. Top athletes are professional and have been training almost every day for years, in their team sport, strength- and coordination training. Thus they should/could first; be better on the test because they are used to perform at a maximum level and second; be less prone to (ACL-) injuries than high school and

college athletes and lower level players. There were however fewer ACL injury incidences on lower level handball (Myklebust et al., 1997). Higher intensity on elite level may be the reason for the greater number of ACL injury incidences, thus elite athletes are in need of better strength and neuromuscular control to restrain the external forces, and so the focus on prevention work is probably more needed on elite level.

## **6.9 Limitations of the Study**

The athletes circled between eight different test-stations, and started at different stations. Some started at the 3D motion station, while others had been through other physical demanding test before performing the VDJs. Thus some athletes could have been more fatigued when performing VDJs, due to previous activity.

Even though the athletes had about 10 minutes of rest, for the marker-placement, and a 5-8 minutes warm-up period before they started the VDJ.

Athletes performing approximately ten jumps during each condition can seem like a lot of jumps, but players should have a sufficient 15-20 seconds of rest to recover between jumps. It is known that sprinters are less fatigued than long distance runners and untrained after a series of 50 intermittent drop-jumps (type-II vs. type-I muscle fibers), with a 20 seconds rest between jumps (Skurvydas, Dudoniene, Kalvénas, & Zuoza, 2002). No such studies involve athletes from the team sports that include high-force jumping, but since they are used to repeated jumping motions in their sports it's plausible to assume they are more fatigue resistant than the athletes tested in the study by Skurvydas et al. 2002. Using countermovement jump, it has been shown that a 6 seconds rest is sufficient to perform over 50 jumps at 95% of max jump height, and over 100 jumps at 95% of max jump height given an 8 seconds rest for physical active male subjects (Pereira, Morse, Ugrinowitsch, Rodacki, Kokubun, & Fowler, 2009). Thus we can believe that the fatigue induced during the VDJ is minimal and will not affect the results of the jump height.

The athletes were not used to perform this specific type of VDJ. Thus some improvement in jump-technique during the testing could have occurred, and since the athletes always finishes with the target-jumps, a possible improvement in technique

would most likely have been present at the time of the target-jumps. Thus affecting the target-jump performance positively. The order of the jumps could also have affect the target-jump performance negatively, since the athletes could have been less motivated after performing a series of VDJs. Test leaders however always tried to inspire the athlete to exert maximal effort.

There are no restrictions on the arm-use during the VDJ. Earlier studies have shown that arms can generate an upwards thrust, and help gaining a greater jump-height (Feltner et al., 1999, 2004). Jumping with arms akimbo will exclude the different arm-techniques that might occur, and the thrust from the arms would have been ruled out and only the thrust from the lower extremities would have been a factor. The present study aimed to find a jump that replicates Hewett, et al., (2005): *"Subject were instructed to jump down from the box and immediately perform a maximal vertical jump, raising both arms as if they where jumping for a basketball rebound"*. The free arm-use would also feel more natural for the athletes, alternatively to having arms akimbo.

Marker movement on the skin of the athlete during the task performed could interfere the data. However excellent test-retest reliability have been found in e.g. valgus angles and moments for athletes performing VDJ with an overhead target (Ford et al., 2007). The reliability however declines with increased complexity of the task performed. The fact that the markers were placed by the same experienced person for all athletes in this study, should secure that marker placement is constant.

## **6.10 Implementation**

The use of an overhead target in VDJ seems to stimulate the athletes' extrinsic motivation, and push them to an even better performance. Thus there should be no hesitation in applying an adjustable overhead target to VDJ, when using it in plyometric training. The overhead-target VDJ can be used without achieving notably greater external knee forces, i.e. increasing the ACL injury risk.

Since we only demonstrated small differences in valgus angles and moments between jumping conditions, the use of an overhead target in screening tasks seems redundant.

The changes were not enough to replicate a more game like situation with the use of an overhead target. It however seems like the sidestep-cutting task is overtaking the VDJ as a 3D-screening tool, due to the more relevant movement pattern.

## **7 Conclusion**

The female athletes improve jump height with 2.6 cm when reaching for an overhead target in a Vertical Drop-Jump. Thus an overhead target will likely have a positive effect on jump performance.

The athletes perform the overhead-target Vertical Drop-Jump with 0.03 seconds less Contact Time and with about  $4.3^{\circ}$  and  $3.1^{\circ}$  less flexion in hip and knee joints respectively, hence they utilize a quicker and more erect jump-technique.

At the same time female athletes exhibit inconsiderable changes in dynamic knee control when reaching for an overhead-target in Vertical Drop-Jumps. The athletes display  $0.5^{\circ}$  greater valgus angle, and 0.05 Nm/kg greater external valgus moments. These changes were statistically significant, but likely clinical insignificant.

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## Glossary

ACL	Anterior Cruciate Ligament
BW	Body Weight
COM	Centre of Mass
FPPA	Frontal Plane Projection Angle
GRF	Ground Reaction Force
IC	Initial Contact
KAM	Knee Abduction Moment
LESS	Landing Error Scoring System
LTS	Lateral Tibial Slope
MBIM	Model Based Image Matching
MCL	Medial Collateral Ligament
MTS	Medial Tibial Slope
NWI	Notch Width Index
OA	Osteoarthritis
PCL	Posterior Cruciate Ligament
ROM	Range of Motion
SLS	Single Leg Squat
SSC	Stretch-Shortening Cycle
TO	Toe Off
vGRF	Vertical Ground Reaction Force