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Reliability and methodological concerns of vertical drop jumping and sidestep cutting tasks – Implications for ACL injury risk screening

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别怕 有我在

Don't be afraid, I'm with you.

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List of papers

This dissertation is based on the following original research papers, which are referred to in the text by their Roman numerals:

I. Reliability of knee biomechanics during a vertical drop jump in elite female athletes. Submitted to *Gait & Posture*

II. Reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks.

Submitted to *Gait & Posture*

III. The effect of thigh marker placement on knee valgus angles in vertical drop jumps and sidestep cutting. *Journal of Applied Biomechanics* 2015:31(4):269-274.

IV. The effect of overhead target on the lower limb biomechanics during a vertical drop jump test in elite female athletes.

Resubmitted with major changes to *Scandinavian Journal of Medicine and Science in Sports*

Abbreviations

ACL	Anterior cruciate ligament
CI	Confidence interval
CMC	Coefficient of multiple correlations
cm	Centimeter
GRF	Ground reaction forces
HJC	Hip joint center
Hz	Hertz
ICC	Intraclass correlation coefficient
kg	Kilogram
LESS	Landing Error Scoring System
m	Meter
MBIM	Model-based Image-matching
mm	Millimeter
Nm	Newton meter
OA	Osteoarthritis
OSTRC	Oslo Sports Trauma Research Center
SD	Standard deviation
STA	Soft tissue artifact
TE	Typical error
VDJ	Vertical drop jump
2D	Two-dimensional
3D	Three-dimensional

Summary

Background

Vertical drop jumping (VDJ) and sidestep cutting tasks have been suggested as movement screening tasks to identify players with increased risk for ACL injury. Previous studies have investigated the reliability of kinematics and kinetics in the two tasks, however the validity is questionable because of small sample sizes or inappropriate statistical methods.

The common method for assessing motion characteristics in such tasks is marker-based 3D motion analysis. However, soft tissue artifacts, particularly at the thigh segment, may largely influence the measurements. Thigh marker placement is typically not standardized and the influence on the measured knee kinematics are not yet quantified.

It is commonly assumed that movement screening tasks may be improved by making them more challenging, e.g. by adding an overhead target to the VDJ test, in order to distract the focus from keeping proper knee control, while attempting to reach the target. The overhead target can thus possibly provoke knee loading and increase the sensitivity of the task to assess ACL injury risk, but this has not yet been showed in a scientific study.

The aims of this thesis were therefore to investigate the reliability of the selected lower limb biomechanical measurements from the VDJ and sidestep cutting tasks, as well as the effect of thigh marker placement, and the inclusion of an overhead target on joint kinetics and kinematics.

Methods

Paper I and II investigated the reliability of VDJ and sport-specific sidestep cutting task respectively. Multiple-session design, on average separated by two weeks, was employed as both within-session and between-session reliability was examined. **Paper III** was a methodological study on the effect of thigh marker placement, and utilized the dataset of **Paper I and II** from the first session. Six different thigh marker clusters were compared. In total 41 athletes, including 19 elite female handball and 22 elite female football (soccer) athletes (mean ± SD: 21.6 ± 4.0 years old, 168 ± 5 cm, 66 ± 8 kg) were recruited for the testing. **Paper IV** was a methodological study on the effect of an overhead target in the VDJ task. The data of the prospective cohort study from 2009 to 2014 was utilized. A total of 523 athletes, including 363 elite female football and 160 elite female handball athletes (mean ± SD: 21.0 ± 4.0 years old, 168 ± 6 cm, 65 ± 8 kg) were tested.

Main results

Paper I: Most of the discrete knee biomechanical variables achieved good to excellent reliability in both within-session (ICC: 0.87 to 0.98) and between-session (ICC: 0.40 to 0.90) analyses. Moreover, we found a moderate to strong between-session consistency of athlete rankings (Spearman's rank correlation: 0.56 to 0.90). The knee flexion angle varied considerably within-session in some athletes, however the task can reliably measure knee valgus angles and internal rotation angles, which are believed to be related to the ACL injury mechanism.

Paper II: All the discrete lower limb biomechanical variables achieved good to excellent reliability in both within-session (ICC: 0.75 to 0.99) and between-session (ICC: 0.55 to 0.92) analyses. The between-session consistency of athlete rankings were moderate to strong (Spearman's rank correlation: 0.54 to 0.89). We furthermore found adequate reliability to be attained from three trials. There were no statistical differences in the reliability of handball- and football-specific sidestep cutting tasks in most variables, except the between-session reliability of a few frontal plane biomechanical variables, where the handball task was more reliable.

Paper III: We found systematic differences up to almost 15° of peak valgus between the marker clusters in the VDJ task. The Spearman's rank correlation coefficient varied from 0.51 to 0.97 among the six marker clusters. In addition, the cross-talk effect varied considerably between the marker clusters. The results of the current study indicate that the choice of thigh marker cluster can have a substantial impact on the magnitude of knee valgus angle, as well as the trial ranking.

Paper IV: The overhead target increased jumping height by 5.8%. Strong to moderate positive Spearman's rank correlations (0.58 to 0.95) were found between two conditions. Furthermore, the overhead target lead to statistically significant changes in many of the lower limb biomechanical variables examined. However, the effect sizes were small (<0.43).

Conclusions

The knee biomechanical variables of the VDJ task, as well as the lower limb biomechanical variables of sport-specific sidestep cutting tasks achieved good to excellent reliability in both within-session and between-session analyses for elite female handball and football athletes. The ranking of athletes based on biomechanical measurements in the two tasks can be reliably reproduced. Large differences in frontal plane knee kinematics were found between the different marker clusters. A standardized thigh marker cluster, including clear guidelines for placing non-anatomical markers, is needed to minimize the variation of the knee valgus angle measurement. Moreover, the differences in joint kinetics and kinematics between the targeted and non-targeted VDJ was small and therefore likely clinically insignificant.

Sammendrag

Bakgrunn

Vertikale tobens fallhopp og finter/retningsforandringer har blitt foreslått som screeningtester for å identifisere spillere med økt risiko for fremre korsbåndskader. Noen få tidligere studier har undersøkt reliabiliteten til kinematiske og kinetiske variabler fra slike øvelser. Validiteten av disse resultatene er imidlertid usikker på grunn av få forsøkspersoner eller feil bruk av statistiske metoder.

Den mest brukte metoden for å vurdere biomekanikk i disse øvelsene er markørbasert bevegelsesanalyse. Imidlertid kan bløtvevsartifakter, spesielt på lårsegmentet, påvirke målingene. Plasseringen av markørene på låret er normalt ikke standardisert og i hvilken grad dette påvirker den målte knebevegelsen er foreløpig ikke kvantifisert.

Det er normalt å anta at slike screeningtester kan forbedres ved å gjøre dem mer utfordrende, f.eks ved å introdusere et overhengende mål som utøveren instrueres i å strekke seg etter i fallhopptesten, for å flytte oppmerksomheten fra knekontroll til å nå opp til dette objektet. Det overhengende målet kan derfor potensielt bidra til økt knebelastning og øke sensitiviteten til screeningtesten.

Hensikten med denne avhandlingen var derfor å undersøke reliabiliteten til utvalgte biomekaniske variabler for underekstremitetene gjennom et tobens fallhopp og to idrettsspesifikke retningsforandringer. I tillegg ønsket vi å undersøke effekten av markørplassering på lårsegmentet på knekinematikk, og til slutt i hvilken grad et overhengende mål kan endre biomekaniske variabler i fallhopptesten.

Metode

Artikkel I og II undersøkte reliabiliteten av fallhopptesten samt de idrettsspesifikke retningsforandringene. Vi gjennomførte to repeterte testsesjoner med to ukers mellomrom og både reliabilitet innenfor og mellom disse sesjonene ble undersøkt.

Artikkel III var et metodestudie der effekten av markørplassering på knevalgus ble studert på bakgrunn av testsesjon 1 fra artikkel I og II. Seks ulike markørkombinasjoner ble sammenliknet. I alt 19 kvinnelige håndballspillere og 22 kvinnelige fotballspillere fra eliteserien (gjennomsnitt ± SD: 21,6 ± 4,0 år, 168 ± 5 cm, 66 ± 8 kg) ble rekruttert til testingen. **Artikkel IV** var et metodestudie hvor effekten av et overhengende mål på biomekaniske variabler i underekstremiteten i fallhopptesten ble undersøkt. I denne studien ble data fra en prospektiv kohortstudie i perioden 2009 til 2014 benyttet. Totalt 160 kvinnelige håndballspillere og 363 kvinnelige fotballspillere fra eliteserien (gjennomsnitt ± SD: 21,0 ± 4,0 år, 168 ± 6 cm, 65 ± 8 kg) ble testet.

Resultater

Artikkel I: De fleste diskrete biomekaniske variablene for kneleddet oppnådde god til utmerket reliabilitet i både innen-sesjon (ICC: 0.87 til 0.98) og i mellom-sesjon (ICC: 0.40 til 0.90) analyser. Videre fant vi moderate til sterke assosiasjoner mellom sesjoner for rankingen av utøvere basert på de samme variablene (Spearmans rangkorrelasjon: 0.56 til 0.90). Knefleksjonsvinkelen varierte betydelig mellom sesjoner for enkelte utøvere, men til tross for dette kunne testen måle valgusvinkler og internrotasjonsvinkler med høy reliabilitet - variabler som antas å være relatert til skademekanismen for fremre korsbåndskader.

Artikkel II: Alle de diskrete variablene for underekstremitetene oppnådde god til utmerket reliabilitet i innen-sesjon (ICC: 0.75 til 0.99) så vel som i mellom-sesjon (ICC: 0.55 to 0.92) analyser. Mellom-sesjon rangkorrelasjonen av utøvere var moderat til sterk (Spearmans rangkorrelasjon: 0.54 til 0.89). Videre fant vi adekvat reliabilitet for kun tre repeterte retningsforandringer. Det var ingen forskjell i reliabilitet for håndballspesifikke og fotballspesifikke retningsforandringer, med unntak av noen få frontalplans-variabler for mellom-sesjon analysene, der håndballspillerne viste høyere reliabilitet.

Artikkel III: Vi fant systematiske forskjeller opp mot 15° i maksimal knevalgus mellom de seks markørkombinasjonene. Spearmans rangkorrelasjon varierte fra 0.51 til 0.97. I tillegg varierte «crosstalk-effekten» betydelig mellom markørkombinasjonene. Resultatene fra denne studien viser at valg av markørkonfigruasjon kan ha en betydelig påvirkning på både størrelsen av de målte valgusvinklene så vel som rangering av spillere basert på deres knevalgus.

Artikkel IV: Introduksjonen av det overhengende målet økte hopphøyden med 5.8%. Moderate til sterke positive Spearmans rangkorrelasjoner (0.58 to 0.95) ble funnet mellom de to betingelsene. I tillegg var det signifikante forskjeller i flere av de målte biomekaniske variablene. Imidlertid var effektstørrelsene små (<0.43).

Konklusjoner

De biomekaniske variablene som ble målt i både fallhopptesten så vel som de to idrettsspesifikke retningsforandringene viste seg å ha god til utmerket reliabilitet i både innen-sesjon og mellom-sesjon for kvinnelige elitespillere i håndball og fotball. De biomekaniske målingene i de to retningsforandringene hadde moderat til høy rangkorrelasjon mellom sesjoner og kan derfor potensielt benyttes til screening.

Store forskjeller i frontalplanskinematikk ble funnet mellom de 6 ulike markørkonfigurasjonene. En standardisert markørplassering, som inkluderer klare retningslinjer for plassering av alle markører, inkludert de som ikke er plassert over anatomiske landemerker, er nødvendig for å begrense variasjonen i målinger av knevalgus. Videre fant vi at forskjellene i leddkinetikk og kinematikk i fallhopptesten, henholdsvis med og uten et overhengende mål, var små og dermed sannsynligvis ikke av klinisk betydning.

Introduction

Physical activity and sports participation is encouraged because of the numerous positive effects including, improving physical and mental health (Thorlindsson et al., 1990; Pate et al., 1995; Richman & Shaffer, 2000; Rossi et al., 2012). Specifically, regular physical activity reduces the risk of hypertension, obesity, and other serious diseases as well as premature mortality in general (Pate et al., 1995; Oja et al., 2010; Drake et al., 2012). However, sports participation is also associated with a risk of being injured, which may have consequences for both the athlete and for the society. In Scandinavian countries, 10-19% of acute injuries in the hospital emergency department are sports injuries (Lindqvist et al., 1996; Ytterstad, 1996).

ACL injuries in sports

Anterior cruciate ligament (ACL) injuries have received growing attention in past decades because of complicated surgical treatment, lengthy rehabilitation and long-term musculoskeletal consequences (Engebretsen, 2014). Therefore, scientific research with the aim of preventing ACL injuries has been announced as a major focus from the International Olympic Committee (Renstrom et al., 2008).

Anatomy

The ACL is one of the four main ligaments of the knee (Figure 1). It has its origin at the femoral notch of the knee with attachments on the inner surface of the lateral femoral condyle and, its insertion at a fossa located anterior and lateral to the medial tibial spine (Duthon et al., 2006). The primary function of the ACL is to restrain the anterior displacement of the tibia relative to the femur, and its secondary function is to restrain knee rotation and frontal plane angulations (Beynnon et al., 2005; Amis, 2012). The ACL is composed of two bundles; the anteriomedial and the posterolateral bundle (Duthon et al., 2006). The two bundles contribute different amounts to anterior tibial restraint as the knee flexes (Amis & Dawkins, 1991; Jordan et al., 2007).

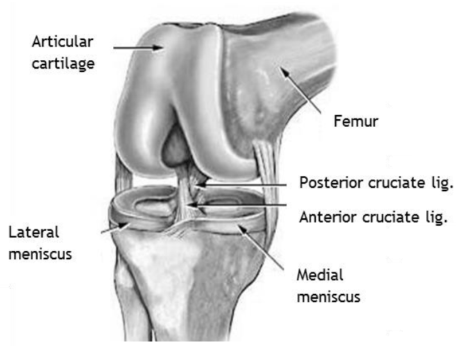


Figure 1. Anatomy of a right knee and the location of ACL (adapted from Nilstad, 2014, PhD thesis)

Consequences

An ACL rupture causes an immediate deterioration in knee stability. After ACL rupture, it is common that the athlete experiences 'giving way' episodes during activity (Lysholm & Gillquist, 1982), especially in pivoting motion (Noyes et al., 1983). Non-surgical rehabilitation program may produce individuals who can compensate for the absence of an ACL without episodes of giving way after return to pre-injury activities (Grindem et al., 2012; Frobell et al., 2013; Lynch et al., 2015). However, surgical treatment is often required for ACL deficient athletes to regain knee function (Engebretsen, 2014). There are several different surgical techniques, however it is still not clear which surgical method is better and always hard to predict the surgical outcome (Tashman et al., 2004; Lam et al., 2011; Goldsmith et al., 2013; Lubowitz et al., 2015).

After ACL reconstruction, long and intensive rehabilitation is necessary. Unfortunately, return to sports is not guaranteed because of potential knee problems such as dynamic instability, reduced range of motion, or pain (Myklebust et al., 2003; Myklebust et al., 2005; Thomee et al., 2015). Moreover, deficits in quadriceps strength may persist after surgery and rehabilitation (de Jong et al., 2007; Eitzen et al., 2010). After a few years, the persistent symptoms after an ACL injury may finally cause a withdrawal from sports (Söderman et al., 2002). In the long term, an ACL injury may cause a significant risk of developing early osteoarthritis (OA) within 15-20 years after injury (Gillquist & Messner, 1999; Lohmander et al., 2004; von Porat et al., 2004; Myklebust et al., 2005). However, the relationship between OA and ACL injury is not completely understood, and the exact relative risk is not clear (Øiestad et al., 2009).

Injured athletes do not fully recover following the ACL injury regardless of the treatment plan. With both surgical and non-surgical treatment, 20-30% of ACL injured athletes will suffer from thigh muscle group strength deficit, loss of range of motion and knee re-injury in two years after injury (Grindem et al., 2015). Either if the ACL is reconstructed or not, there is a noticeable chance of having a long-lasting knee problem, which emphasizes the importance of injury prevention and the need for identifying high risk athletes.

Injury prevention model

A four-sequence injury prevention model has been described by van Mechelen et al. (1992) with the aim of preventing sports injuries (Figure 2). The first step is to identify the incidence and severity of the injury problem. The second step involves identifying the risk factors and injury mechanisms that play a part in the occurrence of injury. The third step is to develop and implement injury preventive measures, and the final step involves evaluating the effectiveness to see if the injury rates are reduced. The following sections will review previous researches on ACL injuries accordingly to the injury prevention model.

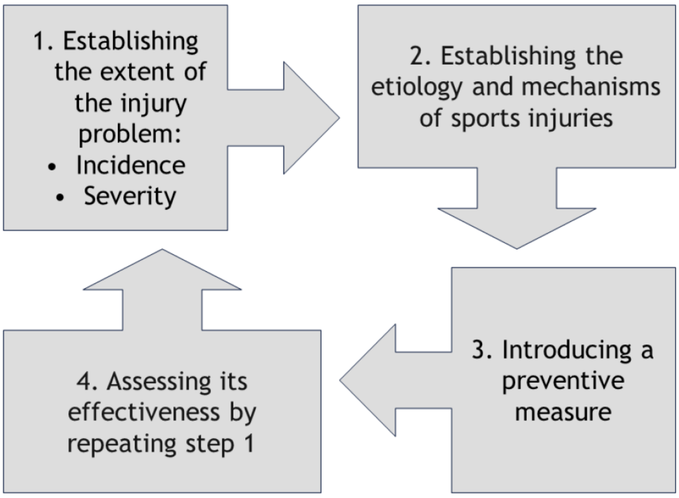


Figure 2. The 4-step sequence of injury prevention research (adapted from van Mechelen et al., 1992)

Incidence

The annual incidence of ACL injury is 62 per 100 000 citizens, according to the national surgical registries (Granan et al., 2008). Eighty percentages of the ACL injury incidents happened during sport (Lind et al., 2009). ACL injuries in sports are of particular concern in female athletes as they have 3-5 times higher risk than male athletes in different sports (Myklebust et al., 1998; Arendt et al., 1999; Renstrom et al., 2008; Walden et al., 2011).

Female handball and football have high injury incidences compared to other sports. In the US, overall incidences were 0.31 and 0.27 injuries per 1000 player-hours in female football and basketball, respectively (Agel et al., 2005). In Norway, Myklebust et al. (1997) reported that the injury incidence for elite female handball athletes was as high as 0.97 injuries per 1000 player-hours. Renstrom et al. (2008) reported that football had the second highest injury rate per 1000 player-hours, which was slightly lower than gymnastics. Combining studies conducted before 2008, it was suggested that 5-10% of the female handball and football athletes would sustain an ACL injury each season, which is equal to one entire team in a typical European league

(Myklebust et al., 1997; Myklebust et al., 1998; Faude et al., 2006; Tegnander et al., 2008). After ten years of performing injury prevention in Norwegian elite female handball, the ACL injury incidence dropped from 1.08 cases per team in the 1989-99 season to 0.42 cases per team in the 2010-11 season (Myklebust et al., 2013).

Causation

Understanding the injury causation is the second step of the injury prevention model, providing the foundation for designing effective injury preventive measures (van Mechelen et al., 1992). Considering the multifactorial nature of sports injuries, Meeuwisse (1994) suggested a model, later revised by Bahr & Krosshaug (2005) describing how an injury likely results from a complex interaction between intrinsic and extrinsic risk factors (Figure 3). This information explains why a particular athlete may be at risk in a given situation (internal and external risk factors for injury), and how injuries happen (mechanism of injury) (Bahr & Krosshaug, 2005). An athlete will be predisposed for injury from internal risk factors and will further be susceptible from external risk factors. However, the inciting event is the final trigger for the injury situation. The inciting event can be described at different levels, i.e. the specific sport situation, the athletes' behaviour and movement, and gross detailed information on biomechanics (Bahr & Krosshaug, 2005).

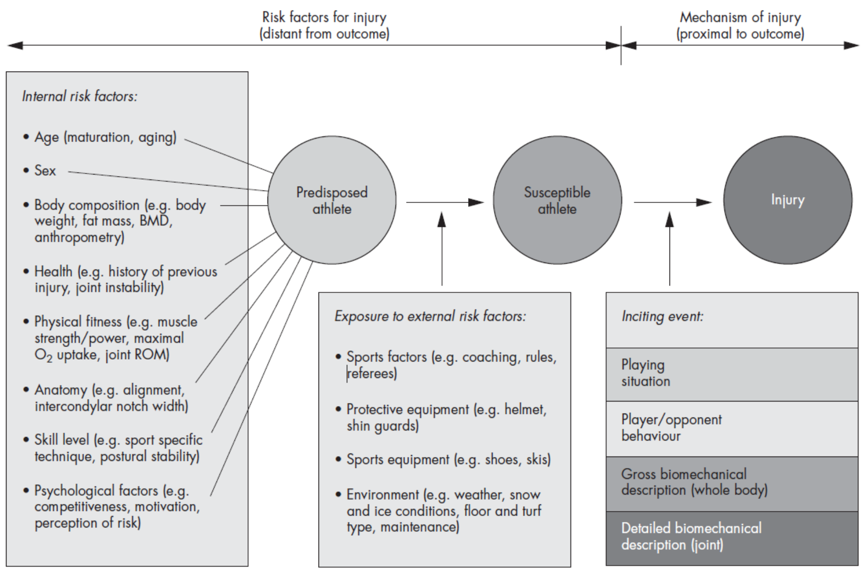


Figure 3. A comprehensive model for injury causation (adapted from Bahr & Krosshaug, 2005)

The ACL injury situation can be classified as contact or non-contact (Myklebust et al., 1997; Boden et al., 2000; Olsen et al., 2004). In a non-contact injury, there is no direct contact between other athletes or objects and the injured knee. Over 70% of ACL injuries have been classified as non-contact, however in females this rate has been described as high as 95%, for instance in Norwegian female handball (Myklebust et al., 1997; Boden et al., 2000; Agel et al., 2005).

A number of different methodological approaches have been used to describe the inciting event (Krosshaug et al., 2005a), including interviews of injured athletes, analysis of video recordings of actual injuries, clinical examinations, measurements from "close to injury" situations, etc. Each approach has its own strengths and weaknesses (Krosshaug et al., 2005a). For example, the usefulness of athlete interviews or video analysis may be limited by the precision of the method (Olsen et al., 2004; Krosshaug et al., 2007c).

Clinical and diagnostic examinations can reveal the injury mechanism retrospectively. Bone bruises in the lateral femoral condyle or posterolateral corner of the tibia are typical MRI findings after ACL injury (Speer et al., 1992; Viskontas et al., 2008). It has been suggested that these bone bruises result from lateral compression during the injury, and that knee valgus is likely involved in the injury mechanism (Speer et al. 1992).

Cadaveric studies can simulate the ACL rupture. DeMorat et al. (2004) found that 4500N of simulated quadriceps loading could rupture six out of 12 cadaveric specimens. It is hypothesized that the pull of the patellar tendon causes an anterior translation of tibia that may strain the ACL and cause injury alone. This injury mechanism was suggested as the quadriceps drawer theory (DeMorat et al., 2004). However, McLean et al. (2004a) found that the ACL loading in sidestep cutting motions generally exceeded the strain threshold reported by DeMorat et al. (2004). Another cadaveric study showed that ACL rupture occurred via excessive compressive loading where the magnitude was similar to the jump-landing in sports (Meyer & Haut 2005). The excessive compressive loading could induce anterior translation of the tibia and internal tibia rotation, which could cause an ACL rupture (Meyer & Haut, 2008). However, the usefulness of a cadaveric study is often hampered by the fact that specimens have structurally degraded and may have lower bone density (Wall et al. 2012).

Measurements of real injury situations are the most valuable because they will tell what is actually occurring during an injury. In recent decades, sports injury situations have often been televised and therefore possible for video analysis. Qualitative video analysis have shown that a high portion of ACL injuries occur in sidestep cutting and single-leg landings (Olsen et al., 2004; Boden et al., 2000; Krosshaug et al., 2007a; Boden et al., 2009; Hewett et al., 2009). Among 20 televised ACL injury cases in female handball, 60% of the cases happened in sidestep cutting and 20% in single-leg landings (Olsen et al., 2004). Krosshaug et al. (2007a) reported that unilateral loading was observed at the time of injury, even if the athlete had both feet on the ground. Boden et al. (2009) and Hewett et al. (2009) described the pattern of flatfooted landing, valgus collapse and lateral trunk lean in injury situations. Visual estimates of joint kinematics from video will have substantial errors such as a systematic underestimation of knee flexion angle (Krosshaug et al., 2007c). Moreover, detailed and precise measurements of joint kinematics were missing from visual estimation. In summary, the typical injury pattern is that the athlete is out of balance and has an extended leg with subsequent knee valgus collapse. However, an alternative approach is needed to estimate the detailed joint kinematics.

A model-based image-matching (MBIM) technique is an alternative approach (Krosshaug & Bahr, 2005; Krosshaug et al., 2007b) to characterize the joint kinematics during injury situations. This method can reconstruct the 3D bony motion from matching a skeleton model onto the video image sequence. This technique has been validated with 3D marker-based motion analysis as the gold standard where good agreement and reliability were achieved (Krosshaug & Bahr, 2005; Mok et al., 2011).

Koga et al. (2010) estimated 3D kinematics and vertical ground reaction forces for ten non-contact ACL injury cases using the MBIM technique. These were injuries occurring in either a handball or basketball matches. The injury situation was either during cutting or a single-leg landing. Based on joint kinematics, Koga et al. (2010) concluded that the injury mechanism included a valgus motion in combination with internal tibial rotation within 40ms after initial contact. The valgus loading caused a lateral compression, and the lateral compressive load, as well as the anterior force vector caused by quadriceps contraction, caused a displacement of the femur relative to the tibia. As a consequence, the lateral femoral condyle shifted posteriorly and the tibia translated anteriorly and rotated internally, resulting in ACL rupture. The sudden external rotation was likely occurring after the ACL was torn (Figure 4). This injury mechanism seemed to be a combination of valgus loading and a quadriceps drawer.

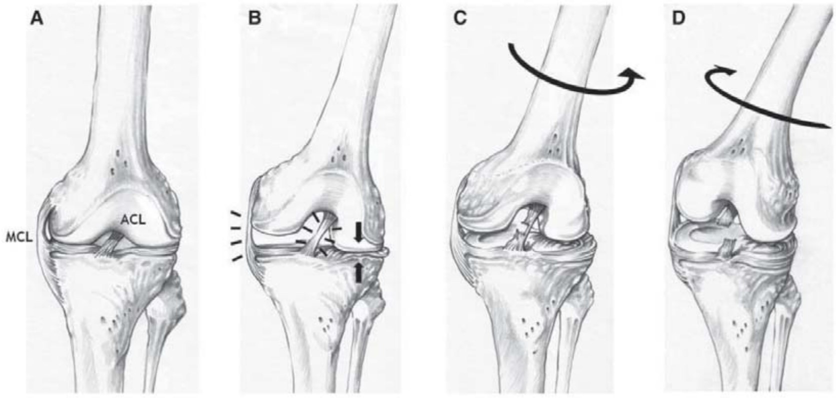


Figure 4. Mechanism of ACL injury as suggested by Koga et al. (2010). A, an unloaded knee. B, when valgus loading was applied, the medial collateral ligament became taut and lateral compression occurred. C, this compressive load, as well as the anterior force vector caused by quadriceps contraction, caused a displacement of the femur relative to the tibia where the lateral femoral condyle shifted posteriorly and the tibia translated anteriorly and rotated internally, resulting in ACL ruptured. D, after the ACL was torn, the primary restraint to anterior translation of the tibia was gone. This caused the medial femoral condyle to also be displaced posteriorly, resulting in external rotation of the tibia.

To summarize, knee valgus and internal rotation seem to be important components of the injury mechanism. Thus, the control of knee frontal and transverse plane motion is likely to be important for ACL injury prevention. Moreover, a consistent pattern of excessive knee valgus collapse in sport motions could be an indicator of high-risk athletes. This could possibly be reflected by the biomechanical variables measured in a typical movement screening task, such as VDJ task.

Biomechanical risk factors

As female athletes have a higher risk of sustaining an ACL injury compared to male athletes, researchers have attempted to identify risk factors of non-contact ACL injury by comparing movement patterns between male and female athletes. Female athletes seem to perform jump-landing motions with reduced hip and knee flexion, and increased adduction and internal rotation of the hip (Ford et al., 2003; Pappas et al., 2007; Sigward & Powers, 2006; Mendiguchia et al., 2011). In general, females seemed to have less sagittal plane motion and greater frontal and transverse plan motion. This is coherent to the excessive knee valgus collapse identified from the injury mechanism investigation. These biomechanical variables were significantly different between males and females, however their associations with an ACL injury remains unclear.

One investigation found an association between discrete biomechanical variables and the ACL injury risk. Hewett et al. (2005) tested 205 American high school female athletes during a VDJ task and a 3D motion analysis technique was used to capture the joint kinematics and kinetics. They recorded nine non-contact ACL injuries, and found that knee valgus angles at initial contact and peak values, as well as peak knee abduction moments, were associated with ACL injury. Knee abduction moments had a sensitivity of 78% and a specificity of 73% to predict an ACL injury status (Hewett et al., 2005). However, the findings were based on nine injury cases only, which is far less than a suggested minimum number of injury cases (20-50) for detecting moderate to strong association between the risk factor and injury (Bahr & Holme, 2003).

To date, there is no solid evidence regarding the association between any biomechanical risk factors and ACL injuries.

Prevention

Introduction of preventive measures is the final step of the injury prevention model. Although ACL injury risk factors are still not entirely clear, prevention strategies and training programs have been developed based on existing knowledge. These approaches have included injury education (Ettlinger et al., 1995), balance training (Caraffa et al., 1996; Söderman et al., 2000), plyometric training (Heidt et al., 2000) and multicomponent training (Hewett et al., 1999; Myklebust et al., 2003; Olsen et al., 2005; Walden et al. 2012).

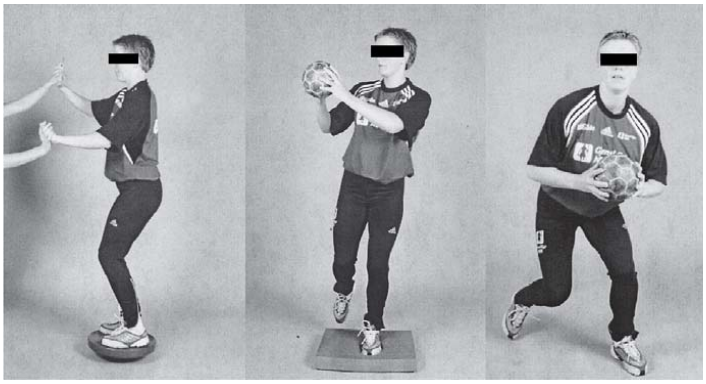


Figure 5. Example of a wobble board exercise (left), a mat exercise (middle) and a floor exercise (right) (adapted from Myklebust et al., 2003)

In Norway, a 15-mins five-phase training program with three different balance exercises focusing on neuromuscular control and planting/landing skills was developed for female handball athletes (Figure 5) (Myklebust et al., 2003; Olsen et al., 2005). Two separate studies resulted in a 40% (Myklebust et al., 2003) and 80% (Olsen et al., 2005) reduction in the ACL injury incidence, respectively. In the US, Mandelbaum et al. (2005) investigated the effect of a multi-component warm-up program on knee injury rates among adolescent female football athletes. ACL injuries were reduced by more than 77% after two years of intervention. Walden et al. (2012) implemented another 15-mins neuromuscular warm-up program targeting core stability, balance and proper knee alignment. The results showed a 64% reduction in the rate of ACL injury in the intervention group compared with the control group.

In particular in football, a comprehensive injury prevention warm-up program, as known as "11+", is promoted by the international football federation. It consists of a basic warm-up, balance training, core training and landing technique training. The effectiveness of the "11+" has been shown to be excellent (Soligard et al., 2008; Steffen et al., 2013). This injury prevention program focuses on all kinds of neuromuscular injuries, including ACL injury.

Recent systematic reviews and meta-analyses have suggested that neuromuscular training combined with injury education can reduce the incidence rate of ACL injuries by 50%, on average (Gagnier et al., 2013; Taylor et al., 2015). Since these programs focus on movement patterns such as knee-over-toe posture, avoiding knee valgus and regaining postural control from perturbations, these movement patterns may correlate to ACL injury risk. Therefore, we still need to investigate the suggested biomechanical risk factors to identify high-risk athletes and to improve existing programs aiming to reduce ACL injury risk.

Injury risk screening

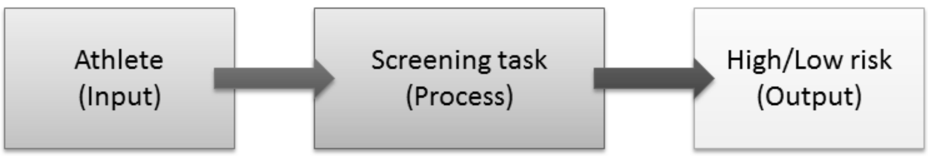


Figure 6. The functional flow of a screening tool.

The purpose of an injury risk screening task is to assess and detect characteristics of an athlete that predispose him/her to increased risk of injury (Figure 6). Such a task is an important tool for sport coaches and medical staff to identify high-risk athletes in order to prevent injuries, as well as providing feedback for movement impairment (Myer et al., 2013b). The high-risk athlete can thereby be prescribed a targeted injury prevention program, a better effect can be achieved from the injury prevention programs (Myer et al., 2007; DiStefano et al., 2009). Moreover, a screening task can be used to investigate the risk factors of an injury through a prospective cohort study. Ideally, the screening task should be reliable, simple and cost-effective.

Vertical drop jump task

The VDJ is the most common and traditional task for assessing knee control (Ford et al., 2003; Hewett et al., 2005; Barber-Westin et al., 2010; Myer et al., 2011a; Noyes et al., 2005; Myer et al., 2013b; Nilstad et al., 2014a; Nilstad et al., 2014b). This task can be used to examine the frontal plane knee control in a standardized and well-controlled situation. Since knee abduction moment in the VDJ task was found to be associated with a future ACL injury (Hewett et al., 2005), the VDJ task was used to develop a clinic-based nomogram to estimate the peak knee abduction moment, in order to identify high ACL injury risk in female athletes (Myer et al., 2011a; Myer et al., 2011b). The clinic-based nomogram employs the clinically obtainable measures of knee valgus motion, knee flexion range of motion, body mass, tibia length and quadriceps-to-hamstrings ratio. This clinical technique was found to predict high knee abduction moments (>25.25 Nm) in female athletes with high sensitivity (85%) and specificity (93%) (Myer et al., 2011b).

Later on, Goetschius et al. (2012) applied the nomogram on 1855 female high school and college athletes with 20 injury cases. They found no relationship between estimated peak knee abduction moments and ACL injury, and concluded that the clinic-based nomogram cannot identify athletes with increased ACL injury risk. Myer et al. (2013a) claimed that their differing results could be explained by the methodological variations. For instance, in the study by Goetschius et al. (2012), the jump involved horizontal movement and the foot separation was not standardized before drop down. Myer et al. (2013a) claimed that this would hamper the effect of provoking knee valgus collapse. Moreover, Goetschius et al. (2012) employed a nested matched case-control design, which would mask out the consideration of weight and height in the nomogram. The nomogram is therefore sensitive to methodological variations, and thus affects the usefulness of the screening task, which is supposed to detect the risk factor consistently and reliably.

Padua et al. (2009) developed an assessment system for the VDJ in order to identify high-risk athletes for ACL injury from their jump landing motion pattern. They instruct the athletes to perform a VDJ task, and the landing motions are assessed by a tool called the Landing Error Scoring System (LESS), which is a system based on 17 motion characteristics, such as a medial knee displacement and an asymmetry of initial foot contact. The LESS was proved to be reliable for both inter-rater and intra-rater (Padua et al., 2009). However, the LESS failed to identify ACL injuries (n=28) in a cohort of 5047 high school and college athletes (Smith et al., 2012). The major reason could be the low variation of the test scores. The possible test scores ranged from zero to 19. However, 75% of the athletes scored less than six, and the mean scores from the high-risk group was only 6.91 with a SD of 0.89 (Padua et al., 2009). In a subgroup of 829 elite-youth football athletes, the LESS demonstrated a good sensitivity of 86% to classify the ACL injuries (Padua et al., 2015). Nonetheless, the result may not be reliable because of the small sample size of the injury group (7 cases) which is far less than the limit of 20 cases, suggested by Bahr & Holme (2003). Therefore, further validation is needed before the LESS can be used as a valid clinical screening tool.

If the screening task is not sensitive for detecting injury risk, a possible reason could be that the task lacks the appropriate exercise intensity or inadequate choice of exercises. Thus, increasing the exercise intensity and applying sport-specific elements to the VDJ task could potentially improve the sensitivity of the test. An external focus of attention could increase the automaticity in movement control (Wulf et al., 2001; Wulf & Dufek, 2007). Therefore, an overhead target in the VDJ task can potentially increase joint loading and distract the athlete from focusing on frontal plane knee control when being tested. Previous studies have confirmed that an overhead target increases jump height (Ford et al., 2005a; Wulf & Dufek, 2007; Wulf & Dufek, 2009). Nonetheless, its effect on lower extremity biomechanics has not been adequately investigated. Wulf & Dufek (2009) concluded that the overhead target increases lower extremity joint loading, but only in a limited sample of four male and six female university students. Ford and co-workers (2005a) found significant increases in knee flexion angle and joint extensor moment in a sample of 18 collegiate female athletes. However, the effect of an overhead target on frontal and transverse plane biomechanics has never been investigated. **Paper IV** focuses on this knowledge gap.

Researchers have focused a lot of their effort on developing a simple and cost-effective VDJ screening task. 3D motion analysis is regarded as the 'gold standard' for assessing joint kinematics and kinetics in the laboratory. However, the technique is expensive and time-consuming. In order to improve the cost-effectiveness of this screening task, 2D motion analysis has been used to assess frontal plane biomechanics. Mizner et al. (2012) assessed the relationship between 2D and 3D knee frontal plane biomechanics. The knee-to-ankle separation ratio accounted for a higher variance of 3D knee abduction angle (R^2=0.350) and knee abduction moment (R^2=0.394) when compared with the frontal plane projection angle (R^2=0.145, 0.254). In addition, the intra- and inter-rater reliability was excellent, with intraclass correlation coefficient (ICC) values of 0.89 and 0.97, respectively. Ekegren et al. (2009) suggested replacing quantitative measures with a novel observational screening guideline to evaluate knee valgus motion in a VDJ by comparing screening scores to 3D motion analysis. Both intra- and inter-rater agreement was reported to be good (Kappa=0.75-0.85). Furthermore, well-educated physiotherapists were found to be able to provide reliable real-time subjective decisions on knee control (Figure 7) (Stensrud et al., 2011; Nilstad et al., 2014b). However, substantial variability in observational assessment of ACL injury risk (when knee abduction moment was used as the criterion) has been documented, thus observer "visual inspection skill" should be considered and assessed (Petushek et al., 2015). However, no study has proven that the observational assessment of the VDJ is directly associated with ACL injury.

Figure 7. Real-time observational screening (A) and 3D marker-based motion analysis during a drop-jump task (B) (adopted from Nilstad et al., 2014)

Sidestep cutting task

The majority of non-contact ACL injuries occur in a single-leg support situation with a rapid change of direction (Boden et al., 2000; Olsen et al., 2004; Krosshaug et al., 2007a). This indicates that a single-leg motion may be preferable in a screening task. Since the support from the contralateral leg would be eliminated, it could be easier to identify poor knee control due to the higher joint loading (Pappas et al., 2007; Weinhandl et al., 2010; Harty et al., 2011). In handball, sidestep cutting is a common motion used to fake the opponent going into the opposite direction (Zebis et al., 2008) (Figure 8). Cutting motions are also common in other sports including basketball, football, badminton and volleyball (Eils et al., 2004; Kimura et al., 2010). In other words, the sidestep cutting task may be better emulate the frontal plane knee loading and control, as encountered during real match situations. Therefore, sidestep cutting task may be more appropriate as a screening task to assess ACL injury risk.

Figure 8. Example of a sidestep cutting movement in an elite female handball match

To date, no cohort studies have assessed a sidestep cutting task for injury risk factor screening purposes. However, the task has been studied extensively in ACL injury related research (McLean et al, 1999; McLean et al., 2004b; Ford et al., 2005b; Sigward & Powers, 2006; Dempsey et al., 2009; Kristianslund & Krosshaug, 2013; Kristianslund et al., 2014a). For understanding the injury mechanism, McLean et al. (2004a) found that sagittal plane knee joint forces alone cannot rupture the ACL during sidestep cutting. The ACL injury motion has to be multi-planar, i.e. involving frontal and transverse planes. For risk factor investigation, sidestep cutting was used to investigate the sex difference in knee biomechanics due to the well-documented sex difference in ACL injury risk (Sigward & Powers, 2006). Results revealed female athletes displayed greater knee abduction moment and less knee flexion moments. For injury prevention, changing cutting technique may reduce knee abduction loading in sidestep cuts. Toe-landing, foot placement close to the body and upright torso posture has been found to reduce the knee abduction loading (Dempsey et al., 2009; Kristianslund et al., 2014a).

The sidestep cutting task could potentially identify other athletes than the VDJ task would. During this task, the athlete rotate around the stance foot, and the movement is not confined to the sagittal plane. Ground reaction forces (GRF) act to change movement direction, therefore the GRF has a large horizontal component and the loads are different from uniplanar tasks (Jindrich et al., 2006). In a study comparing the two tasks, the authors reported that the knee abduction moment was significantly smaller in the VDJ task and weakly correlated with the abduction moments of the cutting task (R=0.135) (Kristianslund & Krosshaug, 2013). Cowley et al. (2006) and Cortes et al. (2011) reported similar findings. In fact, the knee joint loading was approximately 6 times higher in a sidestep cutting task compared to the VDJ. Therefore, the sidestep cutting may be a better test to identify high-risk athletes.

The sidestep task can also be made more demanding and sport-specific by including a static defender (McLean et al., 2004b; Kristianslund & Krosshaug, 2013), passing/receiving a ball (Fedie et al., 2010) or performing in an unanticipated situation (Besier et al., 2001; Ford et al., 2005b). All these sport-specific elements have been shown to increase the knee loading. However, the downside of adding various elements is that it is more difficult to standardize the test properly, possibly leading to higher variability and hence reduced reliability. On the other hand, the variation in sidestep cutting technique likely reflects the variety in cutting during active game play, and hence is similar to real game situation. It is necessary to determine the minimum number of trials necessary to obtain a reliable measure for such a task. **Paper II** addresses this question.

Reliability in screening tasks

A reliable screening task is able to identify the high-risk athlete based on the screening measurements. The selection of the screening measurements refers to the injury risk factor identified by a prospective cohort study, and the prospective cohort study explores the associations between potential injury risk factors and injury outcomes. Bahr & Holme (2003) pointed out that the ability of a prospective cohort study to firmly identify potential risk factors depends on the accuracy of measurements. The information on potential risk factors must be collected with adequate accuracy, and the accuracy of the method can considerably influence the required sample size of study (Bahr & Holme, 2003). The meaning of "accuracy" has been confused by the jargon such as precision, agreement, repeatability and consistency (Hopkins, 2000; Weir, 2005; De Vet et al., 2006). After carefully interpreting the contents and ideas in Bahr & Holme (2003), their "accuracy" is the validity and reliability of a measurement.

- Validity of measurement indicates the degree to which the scores from the test measures what it is supposed to measure (Thomas et al., 2011).
- Reliability refers to the reproducibility of values of a test, assay or other measurement in repeated trials on the same individuals (Hopkins, 2000).

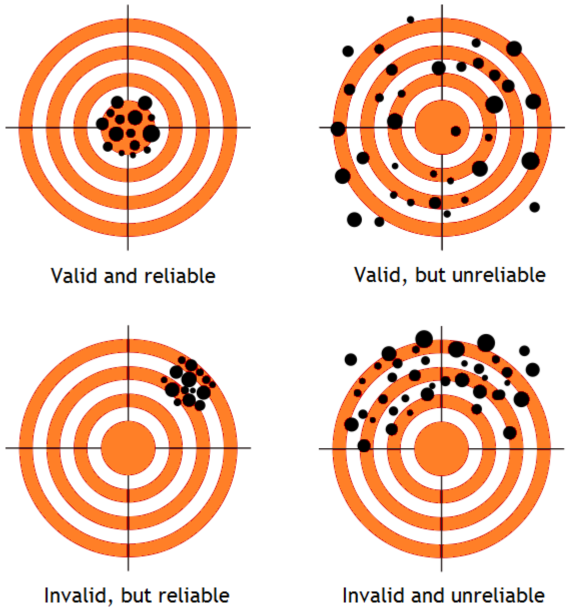


Figure 9. Illustration of the concept of validity and reliability using a dartboard

In general, validity refers to an agreement between the observed value and the true value of a measure. In detail, validity can be classified as logical validity, content validity, criterion validity, construct validity, and predictive validity (Thomas et al., 2011). Depending on measurement types and meanings, validity can have various descriptions.

The objective of a prospective cohort study is to identify the risk factor associated with injury risk. This fits to the definition of predictive validity, which means the degree to which measurements of predictor variables can accurately predict criterion scores (Thomas et al., 2011). Hewett et al. (2005) reported knee abduction moments had a sensitivity of 78% and a specificity of 73% for predicting ACL injury cases (Hewett et al., 2005). This is an example of a predictive validity metrics for injury risk factor. If the predictive validity is high, this implies that the variables were able to predict the high-risk athlete accurately. In other words, the risk factor can firmly identify the high-risk athlete from the screening.

In general, high reliability implies better precision of single measurements and better tracking of changes in measurements in research or practical settings. A measure can be reliable without being valid, which is the left-bottom situations illustrated in Figure 9. Upon retesting, low reliability indicates that large variations in measurement will occur, so that assessment outcomes cannot be meaningfully reproduced or interpreted (Downing, 2004).

A reliability value (ranged from 0 to 1) has as typical basic formula as (Weir, 2005; De Vet et al., 2006):

$$\text{Reliability} = \frac{\text{Variability between subjects}}{\text{Variability between subjects} + \text{Measurement error}}$$

Reliability of the measurement is vital for a prospective cohort study. Low reliability means the measurements includes high degree of measurement error or random error. The consequence of low reliability is the need for an increased sample size to detect association between groups. This has obvious implications for the design of prospective cohort studies that compare groups, i.e. an injury and a non-injury group (Hayen et al., 2007). The low reliability of the measurement can be compensated by the increase in sample size, both injured and non-injured group. For the sake of a screening task, the low reliability could lead to an increased number of trials to obtain enough accuracy of identifying a high-risk athlete, which is the right-top situations illustrated in Figure 9.

The reliability of a measurement can be divided into within-session and between-session reliability. Within-session reliability is the degree to which measurements taken in the same session are consistent. Between-session reliability is the degree to which measurements taken on different sessions are consistent. For the sake of a screening task, the identification of the high-risk athlete is based on the rank order within a group of athlete. The between-session consistency of subject ranking can reflect the reliability of identifying the high-risk athlete.

For instance, if the measurement error equals the variability between subjects, the reliability value becomes 0.5 (De Vet et al., 2006). The ICC is commonly used to describe reliability, however there is considerable confusion concerning both the calculation and interpretation of the ICC (Weir, 2005). The ICC values can be calculated using six different models (Shrout & Fleiss, 1979). Unfortunately, the choice is not straight forwards (Rankin & Stokes, 1998; Weir, 2005; De Vet et al., 2006). The ICC will give high reliability when the subject range is large, even if trial-to-trial variability is large (Shrout & Fleiss, 1979; Weir, 2005). Instead, Spearman's rank correlation will be unaffected by the range in the variable as it transforms the measurements to the ranking domain for the correlation calculation thus is less sensitive to between-subject variability. Therefore, Spearman's rank correlation coefficient can theoretically provide additional information on reliability.

The coefficient of multiple correlations (CMC) has been used to assess the waveform reliability (Ford et al., 2007). However, CMC coefficient measures are sensitive to the data range, indicating lower reliability for smaller range of motion such as knee valgus motion (Growney et al., 1997; Steinwender et al., 2000). Moreover, CMC is generally insensitive to systematic error (Roislien et al., 2012). An alternative to the CMC, the waveform reliability, can be quantified as the typical error of every time point. With this temporal presentation, the movement variability can be further described in a specific region such as initial contact or mid-stance. Using this approach, we could detect landing technique variation between sessions and attribute them to a specific phase of the movement. Furthermore, since the typical error (TE) has the same unit as the measurement, the variability can be directly related the measurement itself (McGinley et al., 2009). Future reliability studies may benefit from using this approach to present waveform reliability.

The reliability of knee biomechanical variables in VDJs has been investigated, both within-session and between-session (Ford et al., 2007; Malfait et al., 2014). Ford et al. (2007) utilized the ICC and TE of various discrete biomechanical variables to quantify the between-session reliability. The majority of the knee kinematic and kinetic variables were shown to have fair to excellent reliability within- (ICC from 0.67 to 0.99) and between-sessions (ICC from 0.59 to 0.92) (Ford et al., 2007) in young female high school athletes. Malfait et al. (2014) assessed the within-session reliability and showed that the knee valgus angle displayed a small variability of only 1.1°, while the variability of knee flexion angle was larger at 3.8° (Malfait et al., 2014). However, the low number of participants in the previous reliability studies is a major concern. Methodology studies of reliability in sports medicine suggest that such studies should contain a minimum of 40 subjects (Atkinson & Nevill, 1998; Hopkins, 2000). The reliability of vertical drop jump tasks have up until now only been investigated in very limited populations, i.e. one study on eight recreational athletes (Malfait et al., 2014) and one on 11 high school athletes (Ford et al., 2007). Likewise, the reliability of medial knee displacement was only reported from a study with five participants (Ford et al., 2003). Furthermore, previous studies have not investigated the reliability of a vertical drop jump task in a homogenous elite populations. Elite female handball and football cohorts are of particular interest, knowing that the risk of sustaining ACL injuries is higher compared with other groups of athletes (Myklebust et al., 1998; Hootman et al., 2007; Prodromos et al., 2007).

The reliability of lower limb biomechanical measurements during sidestep cutting has not been adequately evaluated. Recently, Sankey et al. (2015) investigated the reliability of knee loading variables in sidestep cutting. The study investigated the inter-trial, inter-session and inter-observer reliability. However, the generalizability of results is limited because the sample size was restricted to only four male and four female participants. Sigward & Power (2006) reported the between-session waveform reliability to be acceptable, but only five female football athletes were included. Ford and co-workers (2005) reported the within-session ICC coefficients for knee and ankle frontal plane kinematics to be excellent in 126 adolescent basketball athletes. However, the between-session reliability was not reported. Lastly, Kaila et al. (2007) reported the reliability of lower limb biomechanics in 15 male football athletes, but using Pearson's correlation coefficients only, which was discouraged for assessing reliability (Weir, 2005).

The current thesis will look into the reliability issue of the VDJ and sidestep cutting task which are **Paper I and II**.

Marker-based 3D motion analysis

The main aim of a 3D marker-based motion capture is to reconstruct the movement of bony segments for further analyses (Cappozzo et al., 2005). Therefore, as aforementioned, marker-based 3D motion analysis is commonly used to measure joint kinematics and kinetics in a screening task. This chapter will reveal the basic principles and limitations of the marker-based 3D motion analysis.

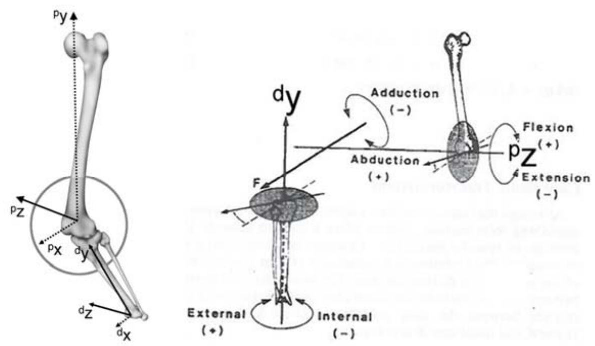


Figure 10. Proximal (px, py, pz) and distal (dx, dy, dz) anatomical coordinate frames used to describe the femur and tibial. Knee joint angles are defined by the rotation occurring about the three joint coordinates axes. (adapted from Grood & Suntay, 1983 and Cappozzo et al., 2005)

Joint kinematics are calculated from the position of markers. Reflective skin markers are put on defined anatomical landmarks of a body, and the markers work as a medium to track the underlying bone. Technically, the markers expose their 3D positions by reflecting the infra-red light emitted by the cameras which are calibrated with known location in the 3D space. Minimum three markers form a marker cluster on a segment. Relevant anatomical frames are used to described the segment orientation and well-defined by the International Society of Biomechanics (Wu et al., 2002). Using the femur and tibia as examples, the longitudinal axis links the proximal and distal joint centers, and the lateral markers near joints (lateral epicondyle for knee, and lateral malleolus for ankle) define the rotational orientation. Hence, the 3D joint kinematics can be calculated from the relative position between the anatomical frame of two contiguous bony segments, such as the femur and tibia (Figure 10). Lastly, the 3D joint kinematics are reported as different joint angles including flexion-extension, adduction-abduction and internal-external rotation. The convention suggested by Grood & Suntay (1983) is the predominant method for describing the lower limb kinematics.

Joint kinetics are calculated iteratively using a Newton-Euler approach (Winter, 1984; Davis et al., 1991). In a motor task, force platform measures the 3D GRF acting on the distal end of the foot segment. Combining the acceleration and inertial properties of the foot segment we calculate the ankle joint moments and forces. With this principle, the hip and knee joint moments can be deduced from the forces acting on the distal joint of the distal segment, and the acceleration and internal properties of the distal segment. The joint moments are expressed in a relevant coordinate system, such as the laboratory frame or local reference frames. As a result, the joint kinetics are reported as joint moments along on the axis of flexion-extension, adduction-abduction and internal-external rotation.

Methodological concerns

The estimates of joint kinematics and kinetics can be affected by many methodological issues such as marker placement (Della Croce et al., 2005), marker cluster configuration (Leardini et al., 2005), calculation approach (Robinson & Vanrenterghem, 2012; Kristianslund et al., 2014b) and signal filtering (Woltring, 1986; Kristianslund et al., 2012).

Such methodological considerations can give a paramount impact to the clinical interpretation of estimates. A remarkable example was shown by Kristianslund et al. (2012) who found that different combinations of filtering frequencies on force and movement data would result in significant differences in ranking of athletes based on their knee abduction moment. The discrepancy can lead to inconsistencies of screening results, because the risk classification is sensitive to athlete's ranking in a group. Consequently, the authors suggested that the same filtering frequency should be imposed on force and movement data.

The most significant source of error in 3D motion analysis is the skin movement with respect to the underlying bones (Cappozzo et al., 1996; Leardini et al., 2005; Peters et al., 2010). This is known as soft tissue artifact (STA). STA has been shown to be most significant in the thigh segment, compared to the foot and shank segments (Reinschmidt et al., 1997a; Reinschmidt et al., 1997b; Benoit et al., 2006; Peters et al., 2010; Tsai et al., 2011). Moreover, the STA has been found to be greatest at the bony landmarks near joint areas. In a cycling motion, the STA on the lateral epicondyle was 40 mm in 120 degrees knee flexion, while the greater trochanter had STA up to 35 mm (Cappozzo et al., 1996). Akbarshahi et al. (2010) described the dynamic behaviour of STA with respect to the knee flexion angle. For a marker on the lateral epicondyle, an unrealistic sudden shift in proximal and anterior directions was found right after heel strike in walking. Similar findings have been reported from a fluoroscopy study, where the proximal thigh showed the largest STA in daily motor tasks (Stagni et al., 2005), whereas the lateral part of the mid-thigh was the region least affected by STA, at about maximal 15 mm. However, this STA magnitude is still greater than for the foot and shank segment, which have less than 10 mm (Peters et al., 2010). These findings can be explained by the fact that the large thigh muscle group contracts significantly during motion.

The effects from STA on the lower limb kinematics are significant (Cappozzo et al., 1996; Holden et al., 1997; Reinschmidt et al., 1997a; Reinschmidt et al., 1997b; Tsai et al., 2011). In gait, the error can reach 10% on flexion-extension, 20% on abduction-adduction and 100% on internal-external rotation of the full range of motion (Cappozzo et al., 1996). In running, the root mean square differences on flexion-extension, abduction-adduction and internal-external rotation are from 4.3° to 8.5° (Reinschmidt et al., 1997b). It was suggested that most of knee rotation errors are due to STA at the thigh, with only minor STA from the shank. Moreover, the STA of the thigh is considerable at initial contact and mid-stance during running. In a cutting motion, the absolute error was the highest on the adduction-abduction angle, with 13.1 degrees reported (Benoit et al., 2006). The STA-induced error in cutting is higher than that in gait and running.

In high-impact motions such as jump-landings, the magnitude of medio-lateral STA of thigh skin marker is crucial (Akbarshahi et al., 2010; Monnet et al., 2012). A medio-lateral marker movement will result in an axial rotation of the thigh anatomical frame along the longitudinal axis. When the knee is flexed, this will result as a knee frontal plane motion. It was well demonstrated by a bi-planar fluoroscopy study from Miranda et al. (2013). An unrealistic peak of knee valgus angle was recorded during a sidestep cutting motion. Moreover, a clear discrepancy can be observed from the profiles of two-motion analysis technique (Figure 11).

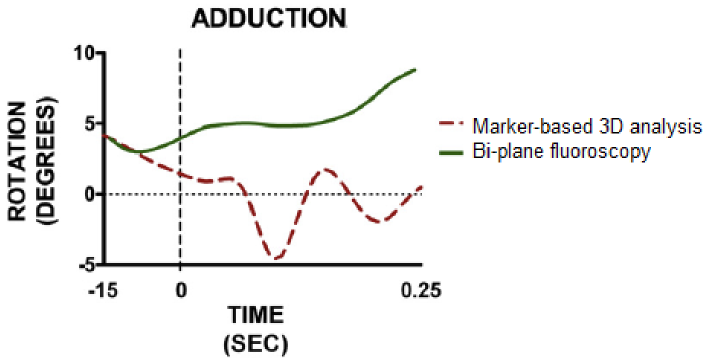


Figure 11. Knee adduction angles of jump-cutting captured by marker-based 3D analysis and bi-plane fluoroscopy (adapted from Miranda et al., 2013)

The compensation of STA is currently not feasible. Although numerous studies have attempted to model the STA in motor tasks (Dumas & Cheze, 2009; Andersen et al., 2012; Camomilla et al., 2013; Dumas et al., 2014), the subject- and task-dependency of STA make it difficult to develop a universal compensation equation (Stagni et al., 2005; Monnet et al., 2009; Camomilla et al., 2013). Therefore, the major STA preventive approach is still putting the marker on an anatomical landmark with the least STA, which was suggested as a surface marker cluster design criteria was advised almost 20 years ago (Cappozzo et al., 1997).

Marker cluster

As aforementioned, a minimum of three markers are needed to track a segment. The formation of markers is known as a marker cluster. The choice of marker clusters is not standardized, and differences have been seen between different protocols. In gait analysis, substantial differences in knee valgus measurements have been reported between five marker cluster protocols, including e.g., Plug-in-Gait and Total 3D Gait (Ferrari et al., 2008). The average difference in knee valgus angles for the entire contact phase was found to be 9.7°, while the maximum difference could be as high as 25° between two protocols (Ferrari et al., 2008). Moreover, Benedetti et al. (2013) examined the pattern similarity of lower limb joint kinematics and kinetics at seven different laboratories employing different gait analysis protocols. The pattern similarity was assessed by the coefficient of determination (R^2). Their results showed that the pattern similarity were excellent in sagittal and frontal plane rotations (R^2>0.9) (except for knee abduction-adduction), however it was poor in transverse plane for knee and hip joints (R^2=0.30). These two studies, in addition to many others, suggest that the differences originate predominantly from the location of the thigh markers, with the largest STA attributed to markers placed at the proximal thigh (Cappozzo et al., 1996; Reinschmidt et al., 1997a; Reinschmidt et al., 1997b; Baker, 2006; Ferrari et al., 2008; Peters et al., 2010).

Certain thigh marker cluster configurations, such as the Plug-in-Gait (Vicon, 2002), include the hip joint center (HJC) as a non-surface virtual marker. It is because the hip joint center is normally estimated based on pelvic markers (Bell, 1990), which are less prone to STA. Therefore, the HJC may be a better proximal anatomical marker than the greater trochanter, where a STA of close to 35 mm has been reported (Cappozzo et al., 1996; Camomilla et al, 2013). The inclusion of HJC may lead to a systematic shift of the knee valgus angle measurement, but this has not yet been investigated.

If the calculated knee flexion-extension axis is not aligned with the true knee flexion-extension axis, a so-called cross-talk error in the calculated knee valgus-varus angle will be induced as a function of knee flexion-extension (Piazza & Cavanagh, 2000). This phenomenon has been shown in both model simulation and in-vivo sport motion (Della Croce et al., 2005; Pohl et al., 2010; Piazza & Cavanagh, 2000; Robinson & Vanrenterghem, 2012). Cross-talk can be minimized by correcting the femoral coordinate system with reference to the functional knee flexion-extension axis, which is believed to be closest to the true knee flexion-extension (Piazza & Cavanagh, 2000; Schache et al., 2006). However, the coordinate frame of the thigh marker cluster defines the knee flexion-extension during dynamic trials (Grood & Suntay, 1983). The knee flexion-extension may be influenced by the erroneous axial rotation of thigh marker cluster, which is induced by STA. For instance, lateral STA of an anterior thigh marker away from the midline of the body will result in an external rotation of the thigh coordinate system along the longitudinal axis, which will result in a greater measure of knee valgus angles when the knee is flexed. Therefore, regardless of knee flexion-extension axis definition, cross-talk may appear during dynamic trials. However, no previous study has investigated cross-talk using a constant knee flexion-extension axis definition while changing the thigh marker cluster. **Paper II** focuses on this knowledge gap.

Different thigh marker clusters have been used in previous studies measuring knee valgus motion during sidestep cutting and vertical drop jumping (McLean et al., 1999; Ford et al., 2003; Hewett et al., 2005; Kristianslund & Krosshaug, 2013). Both the number of markers and their positions are different between these marker clusters. Unfortunately, there is limited information on differences between various marker clusters in high-impact movements such as sidestep cuts and vertical drop jumps. If we are to succeed with determining injury risk factors based on knee valgus angle measurements, consistency in measuring the magnitude as well as the ranking of trials is paramount. **Paper III** of the current thesis will focus on this research question.

Aims of the thesis

The intention of this PhD project was to consolidate the reliability and methodology of two commonly used movement screening tasks; the VDJ and sidestep cutting. A conceptual model of a movement screening task is proposed (Figure 12). To link up the input (athlete) and output (high/low risk) at the two ends, the task should reliably measure the biomechanical variables. In addition, the methodological concerns should be fully understood in order to avoid inconsistency of measurements. High reliability of the biomechanical variables will enhance the chance of detecting the association between the potential risk factor and injury risk. Moreover, in a screening task, high reliability implies that the screening task can consistently identify the high-risk athlete. The association of risk factors with injury, as well as its modifiability, are the determining factor in the design of effective prevention programs.

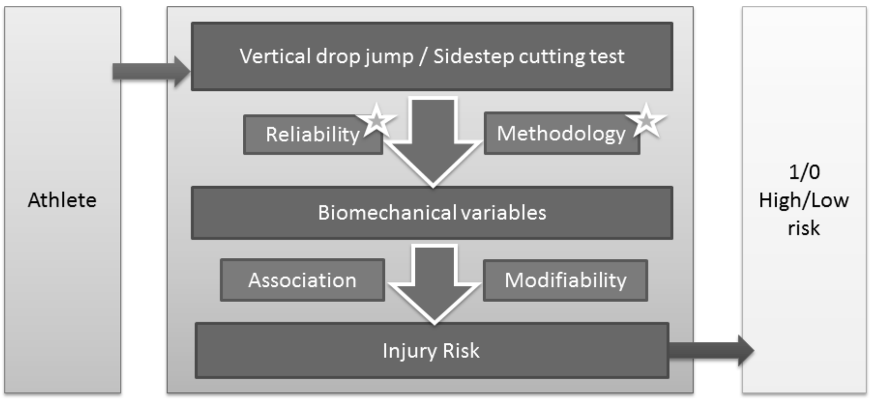


Figure 12. Conceptual model of a screening task. Stars mean the focus of the current thesis.

The specific aims of the separate papers were as follows:

I. First, to assess the within-session and between-session reliability of knee kinematics and kinetics in a vertical drop jump task among elite female handball and football athletes, and second, to quantify the within-session waveform error of measurements and between-session consistency of the subject measurements and rankings (**Paper I**)

II. First, to assess the within-session and between-session reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks among elite female handball and football athletes, and second, to determine the minimum number of sidestep cutting trials necessary to obtain a reliable measure (**Paper II**)

III. First, to quantify the differences in the calculated knee valgus angles between six different thigh marker clusters, second, to investigate the trial ranking based on their peak knee valgus angles, and third, to investigate the influence of marker clusters on the cross-talk effect (**Paper III**)

IV. To investigate the effect of an overhead target on jump height and lower limb biomechanics in all three planes of motion in a VDJ task (**Paper IV**)

Methods

The thesis is part of a project to examine potential risk factors for ACL injuries. The overall study is organized as a prospective cohort study from 2009 through 2014 at the Oslo Sports Trauma Research Center (OSTRC), Norwegian School of Sport Sciences. All data for this thesis are from the baseline testing. Athletes in Norwegian female handball and football top divisions were tested before the season and followed to study candidate risk factors for ACL injury. An invitation letter was sent to the coach of each team with information about the project and practical details related to project participation. All athletes who were expected to play in the top division during the following season were eligible for participation. Athletes with injury that kept them from full participation in team training and competition were excluded.

The risk factors studied included the biomechanics of VDJ and sport-specific sidestep cutting, muscle strength, balance and anthropometry. The thesis is based on the test of the biomechanics of VDJ and sport-specific sidestep cutting.

Study design and participants

Paper I and II investigated the reliability of VDJ and sport-specific sidestep cutting task respectively. Multiple-session design was employed because both within-session and between-session reliability were examined. Therefore, participants performed the two tasks in two separate sessions, on average separated by two weeks, ranging from three to 21 days. **Paper III** was a methodological study on the effect of thigh marker placement, and utilized the dataset of **Paper I and II**. Only data from the first session was used.

We performed a sample size calculation to estimate the minimum sample size requirement for the reliability study. The calculation was performed using the formula of Shoukri et al. (2004). The formula was specifically designed for reliability studies by setting the limit of the confidence interval width of the reliability coefficient. The width of the confidence interval was set to be 0.2 based on the reliability coefficient of VDJ reported by Ford and colleagues (2007). Based on this, with three repeated trials and mean reliability coefficient value of 0.8, the formula gave a minimum sample size requirement of 37 participants. Therefore, we aimed at recruiting more than 37 participants to avoid under-sampling and prepare for potential drop out. In total 41 participants were recruited for the testing, including 19 elite female handball and 22 elite female football (soccer) athletes (mean ± SD: 21.6 ± 4.0 years old, 168 ± 5 cm, 66 ± 8 kg). All participants were recruited from the Norwegian top handball and football division.

Lastly, **Paper IV** was a methodological study where data of the prospective cohort study from 2009 to 2014 was utilized. A total of 523 participants, including 363 elite female football (soccer) and 160 elite female handball athletes (mean ± SD: 21.0 ± 4.0 yrs old, 168 ± 6 cm, 65 ± 8 kg) were tested.

Data collection

Data were collected annually during preseason (Handball: July-August; Football: February-March) at the Norwegian School of Sport Science. Each athlete completed their testing during one day. There were eight test stations, and 60 minutes were reserved to test two athletes at each station. There were three stations for the marker-based 3D motion analysis: anthropometrics; marker placement; motion analysis of VDJ and sport-specific sidestep cutting. For this thesis, only the results from the marker-based 3D motion analysis are included.

Vertical drop jumping task

We instructed the athletes to drop off a 30 cm box and perform a maximal jump upon landing with their feet on separate force platforms (AMTI LG6-4-1, Watertown, Massachusetts, USA). They were required to perform at least three practice trials before three valid trials were collected for each athlete. At least two test operators observed the execution of the jump. They ensured that the trial was performed with maximal effort with correct foot placement on the force platforms and all markers firmly attached to the athlete's skin. If sub-maximal effort was suspected, or when jumping instead of dropping off the box (i.e. increasing the vertical center of mass position at take-off from the box), we asked the athlete to repeat the jump. This task was investigated in **Paper I, III and IV**.

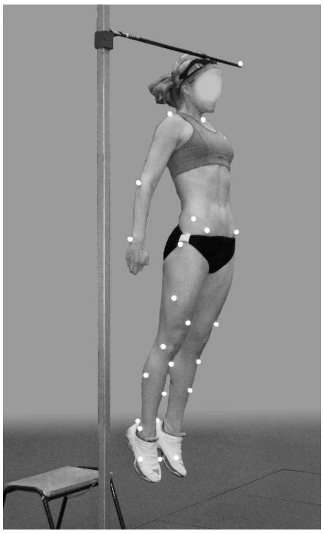


Figure 13. The target condition of VDJ task. The athlete was instructed to drop off a 30 cm box and then perform a maximal jump, aiming to reach the bar with the head.

In **Paper IV**, we included a VDJ task with an overhead target. The athletes first performed the VDJ task in a non-target condition, and subsequently the VDJ task with the overhead target. In the target condition, we set up a horizontal bar about 30 cm in front of the athlete (Figure 13). The height of the bar was set based on jump height in the VDJ task in the non-target condition. We asked the athletes to perform a VDJ and reach the bar by the head. If the athlete managed to reach the bar, it was raised in increments of five cm or less. The task ended when the athlete failed to reach the same height twice and at least three valid trials had been collected. The final three valid trials of each athlete were used for the analysis.

Sidestep cutting task

The sidestep cutting task was designed to be sport-specific. The handball athletes performed a handball-specific faking maneuver involving a static human defender (Kristianslund et al., 2012; Kristianslund & Krosshaug, 2013; Kristianslund et al., 2013; Kristianslund et al., 2014a), whereas the football athletes performed a sidestep cutting with a football through pass.

For the handball-specific protocol, the athlete used an approach run of close to six m, allowing match-like approach speed. The athlete received a lateral pass from a teammate before executing a match-like faking maneuver to pass a 170 cm tall static defender (Figure 14a). The defender adjusted her position during practice trials to ensure that the athlete stepped onto a force platform with her stance foot.

For the football-specific protocol, the athlete also used an approach run of close to six m. A teammate passed a football in a direction that forced the athlete to perform a sharp sidestep cutting maneuver in order to catch up (Figure 14b).

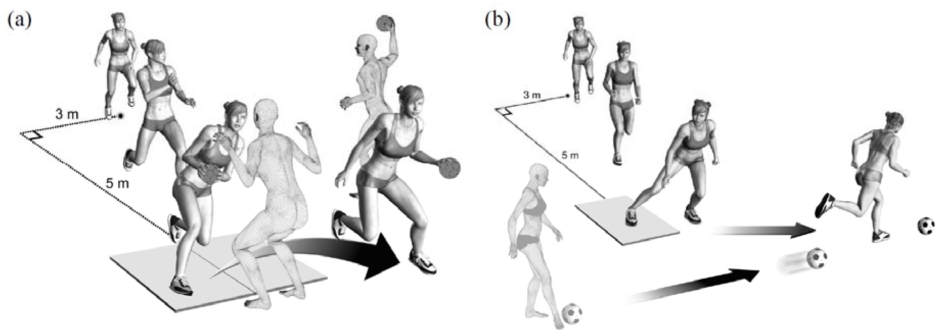


Figure 14. The testing situation of (a) the handball-specific sidestep cutting task and (b) the football-specific sidestep cutting task.

For both tasks, athletes required to perform at least three practice trials to familiarize themselves with the situation, and at least five successful trials from each side (left-right and right-left) were completed. Two test operators ensured that these trials were performed with match-like intensity with the stance foot on the force platform and all markers firmly attached to the athlete's skin. This task was investigated in **Paper II and III**.

Marker-based 3D motion analysis

The athletes wore indoor sport shoes, shorts and a sports bra. The full-body marker setup followed the standard protocol of the OSTRC using 35 retro-reflective markers (Kristianslund et al., 2012a; Kristianslund & Krosshaug, 2013; Kristianslund et al., 2014a; Kristianslund et al., 2014b; Nilstad et al., 2014a; Nilstad et al., 2014b) (Figure 15). In deep hip flexion the markers on the anterior superior iliac spine (ASIS) could be occluded. Therefore, we added two extra markers on the lateral pelvis, located eight cm posteriorly from the anterior superior iliac spine along the crista iliaca. All marker positions were defined uniquely, also those not defined by anatomical landmarks. One experienced physiotherapist, with several years practice of marker placement, was employed in both sessions.

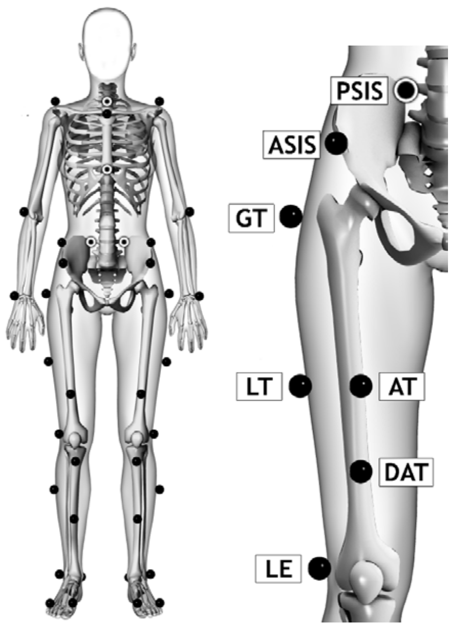


Figure 15. Left: The standard protocol of 37 reflective skin markers (adapted from Kristianslund et al., 2012a). Right: The extensive thigh marker cluster for Paper III.

In **Paper III**, we defined six different thigh marker clusters for the present study (Table 1). Marker cluster I, II and VI were based on previous studies that investigated knee biomechanics in sidestep cutting and vertical drop jumping (McLean et al., 1999; Ford et al., 2003; Hewett et al., 2005; Imwalle et al., 2009; Kristianslund & Krosshaug, 2013). Marker cluster III has no marker on the proximal thigh where we will typically find high STA (Cappozzo et al., 1996; Stagni et al., 2005; Peters et al., 2010). Marker cluster IV and V are composed of markers on the distal thigh and only one marker on the proximal thigh. In marker cluster V and VI, we included the HJC as a virtual marker., since the HJC is normally estimated based on pelvic markers (Bell, 1990), which are less prone to STA. As a result, extra thigh markers were put on the following landmarks bilaterally: lateral femoral epicondyle, greater trochanter, lateral thigh, anterior thigh and distal anterior thigh (Figure 15 and Table 2). Since all six marker clusters were attached to the subject, the current thigh marker set-up allowed all six marker clusters to be calculated simultaneously.

Table 1. Thigh skin marker cluster description

Marker cluster	Skin marker component	Number of markers	Design
I	LE, AT, GT	3	Ford et al. (2003), Hewett et al. (2005), Imwalle et al. (2009)
II	LE, DAT, GT	3	McLean et al. (1999)
III	LE, DAT, LT	3	Distal thigh
IV	LE, DAT, LT, GT	4	Distal thigh + GT
V	LE, DAT, LT, HJC	4	Distal thigh + HJC
VI	LE, DAT, LT, HJC, GT	5	Kristianslund et al. (2012)

Table 2. Definition of skin markers on the thigh segment

Skin markers	Descriptions
Lateral femoral epicondyle (LE)	Lateral epicondyle on femur. It was palpated in the standing position with the subject flexing and extending the knee.
Greater trochanter (GT)	The most superior point of the greater trochanter in standing position
Lateral thigh (LT)	2 cm anterior of the middle point of GT and LE in standing position
Anterior thigh (AT)	The most anterior point of the thigh on the same height of LT in standing position
Distal anterior thigh (DAT)	The most anterior point of the thigh at the level of the mid-point between LT and LE in standing position

We used a 480 Hz 16-camera system (Oqus 4, Qualisys, Gothenburg, Sweden) to capture the motion, while we recorded ground reaction forces using two force platforms collecting at 960 Hz. We calibrated the motion analysis system according to guidelines from the manufacturer, and calculated and tracked marker trajectories using the Qualisys Track Manager (Qualisys, Gothenburg, Sweden).

We defined the contact phase as the period where the unfiltered vertical ground reaction force exceeded 20 N. Marker trajectories and force data were filtered and interpolated using Woltring's smoothing spline in the cubic mode (Woltring, 1986), using a 15 Hz cut-off (Kristianslund et al., 2012). We calculated the hip joint center using the method proposed by Bell et al. (1990), with the anterior-posterior position of the hip joint decided by the anterior-posterior position of the marker over the greater trochanter. Furthermore, we defined the knee joint center according to Davis (1991), and the ankle joint center according to Eng & Winter (1995). Anatomical coordinate systems of the thigh and shank were determined from the static calibration trials. We defined the vertical axis in the direction from the distal to the proximal joint center, while the anterio-posterior axis was defined perpendicular to the vertical axis with no mediolateral component. The third axis was the cross product of the vertical and antero-posterior axes. Consequently, all segments had neutral internal/external rotation in the static calibration trial. We obtained technical, dynamic thigh and shank segment coordinate systems using an optimization procedure involving singular value decomposition (Soderkvist & Wedin, 1993).

We estimated inertia parameters based on 46 measures of segment heights, perimeters and widths using a modified Yeadon's method (Yeadon, 1990), with hand and foot parameters calculated with the method of Zatsiorsky & Seluyanov (1983). We calculated hip and knee joint moments with inverse dynamics using recursive Newton-Euler equations of motion as described by Davis et al. (1991) and projected onto the three rotational axes of the joint according to the joint coordinate system standard (Wu et al., 2002; Grood & Suntay, 1983; Kristianslund et al., 2014b).

We used the Grood & Suntay (1983) convention for calculating joint angles from the marker-based motion analysis. In **Paper III**, we calculated knee joint angles with the six different thigh marker clusters using otherwise identical methods. In **Paper I, II and IV**, we introduced a novel calculation of medial knee displacement. For the VDJ task (**Paper I and IV**), we calculated medial knee position as the perpendicular distance between the knee joint center and the line joining the ankle and hip joint centers, projected on the frontal plane (Figure 16). We only calculated medial knee position when the knee joint center was medial to the hip-ankle line. Otherwise, medial knee position was set to zero. The difference between the perpendicular position at the initial foot contact and the peak value was defined as the medial knee displacement. For the sidestep cutting task (**Paper II**), we calculated medial knee position as the perpendicular distance between the knee joint center and the line joining the ankle and hip joint centers projected onto the plane, defined by the knee joint flexion axis and the line joining the ankle and hip joint centers. The difference between the position at the initial foot contact and the peak value was defined as the medial knee displacement for analysis. An advantage of these conventions compared with a pure knee separation measure is that we can assess knee control individually for the left and right leg. We ran all calculations using custom Matlab scripts (MathWorks Inc., Natick, Massachusetts, USA).

Figure 16. Illustration of the medial knee position in a knee valgus posture. The difference between the medial knee position at the initial foot contact and the peak value was defined as the medial knee displacement.

Statistical analysis

In **Paper I**, we used the measurements from the right leg only for simplicity. Each trial was time-normalized from 0 to 100% of the stance phase. For every time point, we calculated the typical error based on three trials from each subject. The typical error was calculated from the standard deviation of inter-trial differences divided by the square root of two (Hopkins, 2000). The typical error represented 52% of test-retest differences of a subject in the sample group (Hopkins, 2000). Moreover, the mean curves and standard deviation were computed to represent the motion and between-subject variability.

We extracted the discrete knee biomechanical measurements including kinematics at the initial contact, peak and range of motion, the peak kinetics, and the peak vertical ground reaction force. We have reported the mean, standard deviation, and typical error of both within and between-session measurements. The ICC values for both within-session (ICC(3,k)) and between-session (ICC(3,1)) were computed (Rankin & Stokes, 1998; Ford et al., 2007). To assess the consistency of subject ranking between-sessions, Spearman's rank correlation coefficients were calculated based on the mean measurement from sessions one and two. Paired t-tests were employed to assess the difference of the mean measurement between sessions. Cohen's d was computed to assess the effect size of the mean differences.

In **Paper II**, thirty-three discrete biomechanical variables were extracted from the lower limb joint kinematics, joint kinetics and forces. We defined the dominant leg as the preferred leg when kicking a ball. We used paired t-tests and Pearson's correlations to evaluate the symmetry between dominant and non-dominant leg (Sadeghi et al., 2000). We reported the mean, standard deviation, and typical error of the within- and between-session measurements. We computed ICC values for within sessions (ICC(3,k)) and between sessions (ICC(3,1)) (Rankin & Stokes, 1998; Ford et al., 2007). The ICC values were computed based on trials 1-3, 1-4 and 1-5, separately. We used a Z-test to test the significance of difference between the ICC values of handball and football athletes. The critical value for the Z-score was set at 1.96.

To assess the consistency of subject ranking between-sessions, Spearman's rank correlation coefficients and between-session ICC were calculated based on the mean measurements from sessions one and two. Paired t-tests were employed to assess the mean difference between sessions. Cohen's d was computed to assess the effect size of the mean (Cohen, 1992). Moreover, the mean curves were time-normalized from 0 to 100% of the stance phase.

In **Paper III**, the peak knee valgus angle during the contact phase and knee valgus at initial contact were compared across the six different thigh skin marker clusters using one-way repeated-measures ANOVA. Each subject provided the measurements of right knee from three VDJ`s and five sidestep cutting trials. Each trial (N = 123 for vertical drop jumps; N = 205 for sidestep cuts) was considered as a single data point and used to construct the distribution for the statistical analyses. Post-hoc t-tests with Bonferroni correction for repeated tests were then conducted to investigate which marker clusters differed by pairwise comparison, 30 pairs in total for six marker clusters and two outcome measurements. Spearman's rank correlation coefficients were calculated to assess the consistency of trial ranking across conditions. As an indirect measure of the cross-talk, the correlation between the knee flexion-extension and varus-valgus angle profiles was used (Schache et al., 2006).

In **Paper IV**, we calculated the jump height as the difference of the vertical center of mass position between the static anatomical position and the maximal height position during the jump. Thirty selected biomechanical variables were extracted from the joint kinematics, joint kinetics and force time course for the analyses. We extracted variables for both legs and organized them into dominant and non-dominant leg. The mean of three trials for each athlete was used for the analysis of each variable. A two-way ANOVA was used to determine the main effects and interaction of leg dominance and condition on each variable. If a significant interaction was found, post-hoc paired t-tests were done separately for the dominant and non-dominant leg. We reported the mean with standard deviation for each variable from the two conditions. Moreover, we computed effect size as the mean difference divided by the pooled standard deviation to assess the clinical significance of the mean difference between conditions (Cohen, 1992). To assess the consistency of athlete ranking between two tasks, we calculated Spearman's rank correlation coefficients based on the measurements from the non-target and target condition.

The ICC classification of Fleiss (1986) was used to interpret the ICC values (less than 0.4, poor; between 0.4 and 0.75, fair to good; and greater than 0.75, excellent). The classifications of Zou et al. (2003) were used to interpret the rank correlation coefficients (greater than 0.5, moderately positive; greater than 0.8, strongly positive). Moreover, we interpreted effect size as follows: <0.2, no effect; 0.2-0.5, small effect; 0.5-0.8, medium effect; >0.8, large effect (Cohen, 1992). Statistical significance was set at $p<0.05$. Statistical analyses were performed using SPSS 18 (SPSS Inc., Chicago, IL, USA) and the statistics toolbox of Matlab (MathWorks Inc., Natick, Massachusetts, USA.

Ethics

The study was approved by the Regional Committee for Medical Research Ethics; South-Eastern Norway Regional Authority, and by the Norwegian Social Science Data Services, Norway. All athletes were recruited from the Norwegian top handball and football divisions and had signed informed consent to confirm participation in the project, including parental consent for athletes below 18 years of age.

Results and discussion

Reliability

Reliability of knee biomechanics during a vertical drop jump in elite female athletes (Paper I)

All the selected discrete variables achieved excellent within-session reliability with all ICC values greater than 0.87 (Table 3). The within-session typical errors were generally small, and errors related to the valgus angles were less than 1.0°.

The between-session ICC values for most of the selected discrete variables achieved good to excellent between-session reliability (Table 3). However, peak internal rotation moment displayed only fair between-session reliability with an ICC value of 0.40. All the rank correlation coefficients demonstrated positive correlations on the between-session athlete ranking. Peak flexion moment, peak valgus angle, peak internal rotation, medial knee displacement and jump height showed a strong between-session consistency with rank correlation coefficients greater than 0.8 (Zou et al., 2003). The flexion angle at initial contact, peak flexion angle, knee flexion range of motion, peak internal rotation and the medial knee displacement were significantly different between sessions ($p<0.05$). However, all the Cohen's d values were smaller than 0.31, indicating that the effect sizes of the mean differences were clinically insignificant.

We observed an increase in typical error during the mid-stance for the knee flexion angle, medial knee displacement, knee flexion moment, and knee valgus moment. The maximal typical error for the vertical ground reaction force was found in the first 15% of the stance phase. The within-session typical errors of the knee valgus, internal rotation angle and internal rotation moment was relatively constant throughout the whole stance phase (Figure 17 and 18). Differences in typical error waveforms were observed between sessions (Figure 17 and 18). The knee flexion angle, medial knee displacement, flexion moment and valgus moment had a higher typical error in the mid-stance phase in session two.

Table 3. Reliability values for selected knee biomechanical measurements in VDJ task.

	Session 1					Session 2					Between Session					
Variables	**Mean**	**SD**	**TE**	**ICC(3,k)**	**95% CI**	**Mean**	**SD**	**TE**	**ICC(3,k)**	**95% CI**	**TE**	**ICC(3,1)**	**95% CI**	**Spearman's rho**	**95% CI**	**Cohen's d**
Kinematics																
Knee flexion angle at IC (°)	29.66	8.45	3.05	0.90	0.83-0.94	31.50	9.11	2.88	0.93	0.88-0.96	2.40	0.79	0.64-0.88	0.78	0.58-0.89	0.21*
Peak knee flexion angle (°)	108.32	14.66	3.19	0.95	0.92-0.97	112.89	15.12	5.47	0.92	0.87-0.96	4.78	0.74	0.57-0.85	0.72	0.47-0.86	0.31*
ROM of knee flexion angle (°)	100.09	14.64	3.69	0.94	0.90-0.97	103.61	15.16	5.54	0.91	0.85-0.95	4.00	0.76	0.60-0.87	0.76	0.55-0.88	0.24*
Knee valgus angle at IC (°)	-1.07	3.54	0.70	0.97	0.95-0.98	-1.09	4.09	0.61	0.98	0.97-0.99	1.11	0.75	0.58-0.86	0.69	0.44-0.84	<0.01
Peak knee valgus angle (°)	6.92	5.21	0.77	0.98	0.97-0.99	6.60	5.60	0.94	0.98	0.96-0.99	1.51	0.81	0.67-0.89	0.81	0.63-0.91	0.06
ROM of knee valgus angle (°)	10.69	3.40	0.92	0.92	0.86-0.95	10.94	3.65	0.98	0.94	0.91-0.97	1.06	0.81	0.66-0.89	0.79	0.59-0.89	0.07
Internal knee rotation angle at IC (°)	-0.25	4.86	1.26	0.94	0.91-0.97	0.003	5.83	1.27	0.95	0.92-0.97	1.58	0.75	0.58-0.86	0.73	0.49-0.86	0.05
Peak internal knee rotation angle (°)	9.57	5.69	1.04	0.98	0.73-0.99	10.78	7.23	1.26	0.98	0.71-0.99	1.80	0.84	0.73-0.91	0.85	0.70-0.93	0.19*
ROM of internal knee rotation angle (°)	16.02	5.17	1.79	0.91	0.86-0.95	16.78	6.99	1.91	0.94	0.90-0.97	2.28	0.73	0.55-0.85	0.77	0.56-0.88	0.12
Medial knee displacement (cm)	3.54	1.86	0.59	0.91	0.86-0.95	4.22	2.46	0.86	0.92	0.87-0.96	0.87	0.75	0.58-0.86	0.80	0.62-0.90	0.31*
Jump height (m)	0.39	0.05	0.01	0.98	0.96-0.99	0.39	0.06	0.01	0.98	0.97-0.99	0.01	0.90	0.82-0.95	0.89	0.79-0.95	0.08
Kinetics																
Peak knee flexion moment (Nm)	122.37	33.60	8.26	0.95	0.92-0.97	128.08	40.57	12.31	0.95	0.91-0.97	9.96	0.85	0.74-0.92	0.90	0.80-0.95	0.15
Peak knee abduction moment (Nm)	21.44	9.82	4.21	0.87	0.78-0.93	23.64	15.12	7.06	0.88	0.80-0.93	5.63	0.69	0.50-0.83	0.72	0.49-0.86	0.17
Peak knee internal rotation moment (Nm)	12.50	4.41	1.51	0.90	0.83-0.94	12.51	4.82	1.95	0.87	0.78-0.93	2.59	0.40	0.11-0.63	0.56	0.25-0.76	<0.01
Vertical GRF (N)	930	153	53	0.91	0.85-0.95	935	157	54	0.91	0.85-0.95	56	0.69	0.48-0.82	0.73	0.50-0.86	0.03

SD: Standard deviation; TE: Typical Error; ICC: Intra-class correlation; CI: Confidence interval; IC: Initial Contact; ROM: Range of motion, GRF: Ground reaction force.

** $p<0.05$, significant difference between the mean measurements of session 1 and 2.*

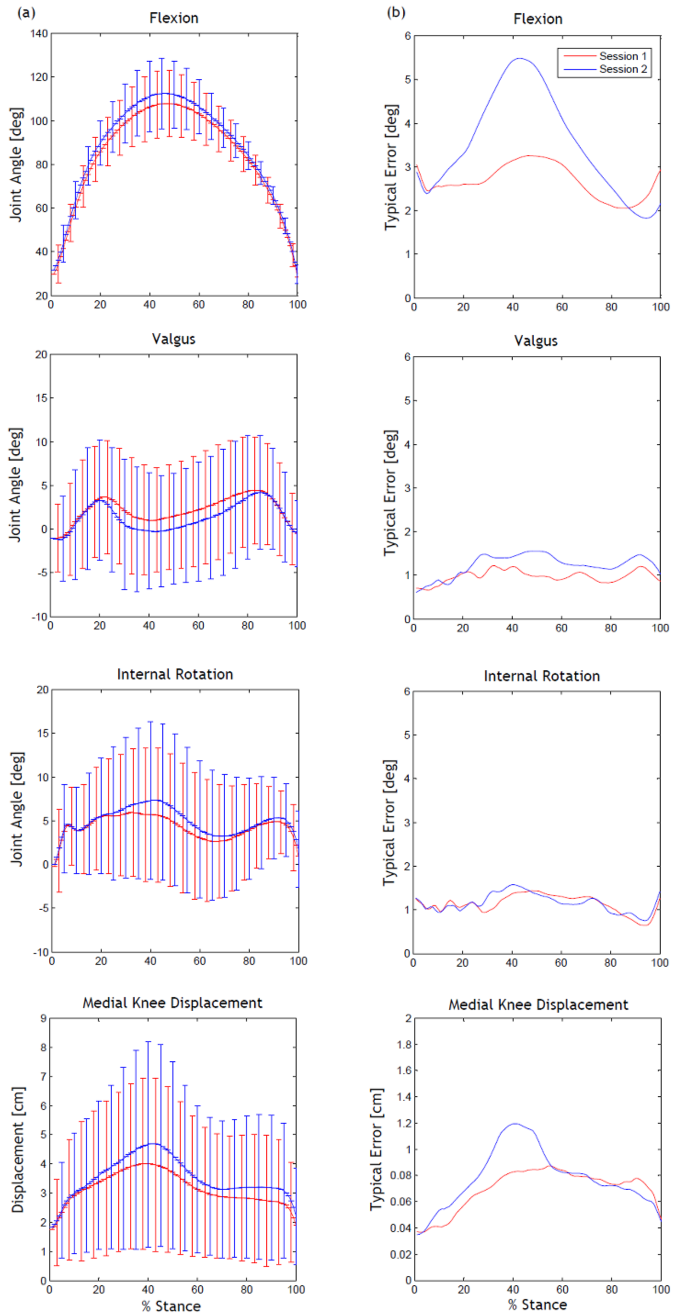


Figure 17. a) Joint kinematics mean curves with standard deviations from session 1 and 2. b) Mean typical errors. Red: Session 1; Blue: Session 2.

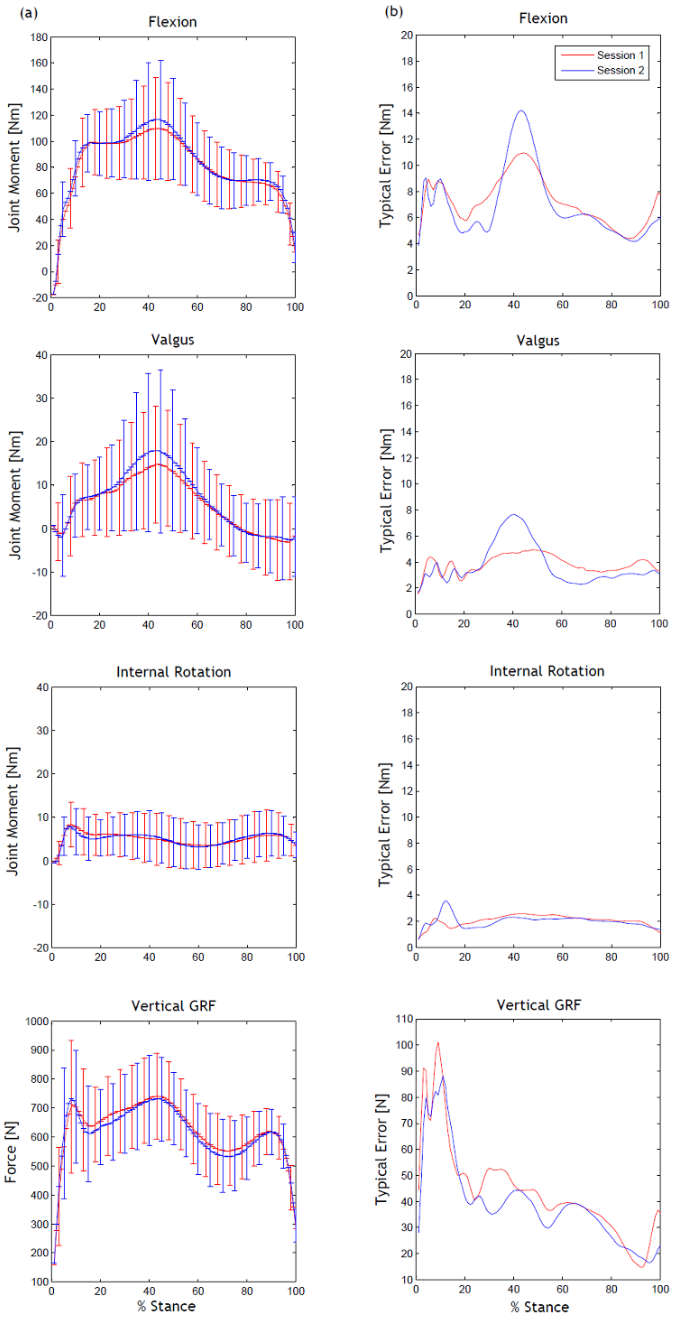


Figure 18. The kinetics and vertical GRF mean curves with standard deviations from session 1 and 2. b) Mean typical errors. Red: Session 1; Blue: Session 2.

Most of the discrete knee biomechanical variables achieved good to excellent reliability in both within-session and between-session analyses. Moreover, we found a strong between-session consistency of athlete measurements and rankings, implying that the estimates could reliably reproduce the testing results in both single-session and multiple-session studies. Furthermore, the ranking of athletes based on knee biomechanics in the vertical drop jump task can be reproduced reliably, which is critical for injury risk screening purposes. Although the knee flexion angle varied considerably within-session in some athletes, the task can reliably measure knee valgus angles and internal rotation angles, which are believed to be related to the ACL injury mechanism (Hewett et al., 2005; Koga et al., 2010).

The within-session reliability (ICC: 0.87 to 0.98) was, in general, better than between-session reliability (ICC: 0.40 to 0.90). The lower reliability between sessions is likely a result of variability in skin marker placement and changes in athlete movement (Kadaba et al., 1989; Steinwender et al., 2000; Gorton et al., 2009). Although the ICCs implies the between-session measurements were less reliable, most variables still achieved good to excellent between-session reliability except the peak knee internal rotation moment. Moreover, the between-session typical errors and effect sizes of the difference were small. Therefore, the results supported that most of the knee biomechanical variables could be reliability reproduced.

We were surprised to see a relatively large typical error of knee flexion angle and medial knee displacement during mid-stance in session two (Figure 17). However, a further analysis showed that the large typical error was generated predominantly by two athletes who performed jumps with both high and low knee flexion within the same session. When removing the two athletes from the analysis, the typical error waveform in session two was similar to session one. Landing technique standardization or instruction may likely attenuate the variation in knee flexion angles (Young et al., 1995; Marshall & Moran, 2013). In our testing protocol, the knee flexion angle was not standardized because we did not want to impose a specific jump landing technique to the athletes. Although one may speculate that some jumps amongst the athletes with high variability should be excluded due to submaximal jumping performance, the jump heights in fact turned out to be nearly identical. Hence, the jumps were correctly assessed to be valid trials according to our definition. Importantly, the knee valgus angles and internal rotation angles had a small and constant typical error throughout the whole stance phase in both sessions despite the variability in knee flexion.

An increase in typical error appeared in all the kinetic variables between 0 to 20% of the stance phase in all three directions (Figure 18). This is consistent to previous findings from Malfait and co-workers (2014) who also concluded that kinetic measurement shortly after initial contact were more variable. In contrast, they did not find a similar variation of the kinetic measurement in the mid-stance as found in our study. The larger sample size (n=41) in our study may potentially explain the greater kinetic variability which was absent in the aforementioned study (n=8).

The results obtained in this study are strikingly similar to those of Ford et al. (2007). With the exception of the internal rotation moments, the within- and between-session reliability for the two studies was close to identical (average ICC difference less than 0.1 for all common variables). In other words, good to excellent reliability of knee biomechanical measures have been obtained across two different biomechanical laboratories/protocols, including different cohorts of elite and reactional female high school athletes.

Reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks (Paper II)

Only small changes were seen when increasing the number of trials from three to five (Table 4). For the dominant leg, the mean between-session ICC value improved from 0.73 (good) to 0.75 (excellent), and the mean within-session ICC values improved from 0.91 to 0.95 (both excellent). For the non-dominant leg, the mean between-session ICC value improved from 0.75 (excellent) to 0.78 (excellent), and the mean within-session ICC values improved from 0.91 (excellent) to 0.94 (excellent). Since there were only slight improvements in reliability from increasing the number of trials, we have reported reliability measures based on trials 1-3 in the following.

Table 4. The average ICC values (interpretation from Fleiss, 1986) of all 33 variables on dominant and non-dominant leg from different sum of trials

	Leg	Trial 1-3	Trial 1-4	Trial 1-5
Within-session	**Dominant**	0.91 (Excellent)	0.94 (Excellent)	0.95 (Excellent)
	Non-dominant	0.91 (Excellent)	0.93 (Excellent)	0.94 (Excellent)
Between-session	**Dominant**	0.73 (Good)	0.74 (Good)	0.75 (Excellent)
	Non-dominant	0.75 (Excellent)	0.77 (Excellent)	0.78 (Excellent)

All 33 variables showed a statistically significant correlation between sides (mean: 0.59, 95% CI: 0.53-0.65). Only four out of all variables showed significant differences between limbs in the paired t-test (Peak hip abduction angle, the peak knee internal rotation angle, the peak knee valgus moment and the peak knee flexion moment). This implies that the majority of the variables were dependent and symmetrical between dominant and non-dominant leg in the sport-specific sidestep cutting task. For simplicity, we therefore reported reliability measures based on the dominant leg only.

There were no significant differences in the within-session ICC values of handball and football athletes. For the between-session ICC values, two variables on the dominant leg and five variables on the non-dominant leg showed a significant difference (Table 5). Since the vast majority of the ICC values showed no significant difference between handball and football, we pooled the results of handball and football athletes (Table 6). Six variables showed a significant difference between sessions. All the Cohen's d values were less than 0.21, which implies a small effect size. All the variables showed excellent within-session reliability (mean ICC: 0.91, 95% CI: 0.89-0.93), and good to excellent between-session reliability (mean ICC: 0.73, 95% CI: 0.70-0.76). In addition, all the between-session rank correlation coefficients demonstrated moderate to strong positive correlation (mean: 0.72, 95% CI: 0.69-0.76).

Table 5. The list of variables having significant difference on the between-session ICC values between handball and football athletes, significant level set at Z-score > 1.96

	Between-session ICC		
Variables	**Football**	**Handball**	**Z-score**
Dominant leg			
Peak hip adduction moment	0.37	0.90	2.71
Peak knee valgus angle	0.22	0.72	2.12
Non-dominant leg			
Peak medial shear force	0.59	0.93	2.30
Peak ankle eversion moment	0.31	0.85	2.55
Peak knee abduction moment	0.51	0.95	2.57
Peak ankle plantarflexion angle	0.88	0.52	2.07
Peak hip adduction angle	0.32	0.91	3.05

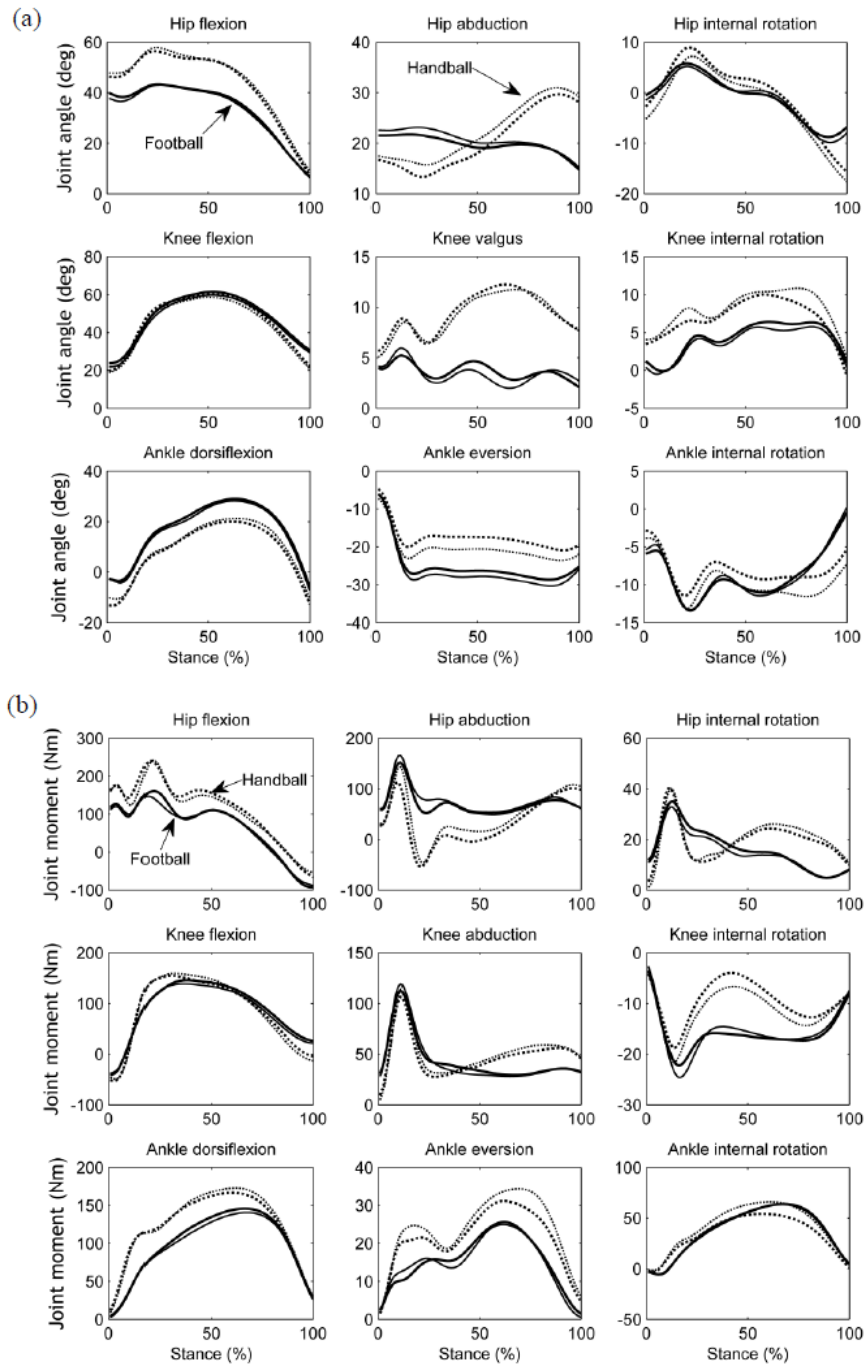
The time courses of mean joint kinematics (Figure 19a), joint kinetics (Figure 19b) and ground reaction forces (Figure 19c) were consistent between sessions. However, we could observe substantial differences in some of the waveforms between the handball- and football-specific sidestep cutting tasks.

Table 6. Within- and between-session mean typical error and reliability measurements for dominant leg from trials 1-3, pooled handball and football sport-specific sidestep cutting

	Within-session					Between-session					
	Mean	SD	Typical Error	ICC(3,k)	95% CI	Typical Error	ICC(3,1)	95% CI	Spearman's rho	95% CI	Cohen's d
Kinematics											
Hip internal rotation angle at IC (°)	-1.7	8.4	2.1	0.94	0.90-0.97	3.3	0.64	0.42-0.79	0.64	0.37-0.82	0.11
Peak hip internal rotation angle (°)	10.9	7.7	1.8	0.95	0.92-0.97	2.6	0.73	0.55-0.85	0.74	0.52-0.87	0.07
Hip abduction angle at IC (°)	18.6	7.5	1.7	0.96	0.93-0.97	2.5	0.76	0.59-0.86	0.84	0.68-0.92	0.14
Peak hip adduction angle (°)	-11.2	7.1	2.4	0.92	0.87-0.96	2.7	0.64	0.42-0.79	0.72	0.49-0.86	0.13
Peak hip abduction angle (°)	27.9	6.9	1.8	0.94	0.89-0.96	2.5	0.66	0.44-0.80	0.69	0.44-0.84	0.18*
Hip flexion angle at IC (°)	43.1	11.9	1.5	0.99	0.98-0.99	4.5	0.72	0.53-0.84	0.83	0.68-0.92	0.01
Peak hip flexion angle (°)	51.3	13.6	2.9	0.96	0.94-0.98	4.4	0.74	0.56-0.85	0.78	0.58-0.89	0.06
Knee internal rotation angle at IC (°)	4.0	8.5	2.9	0.93	0.88-0.96	3.1	0.70	0.50-0.83	0.70	0.45-0.85	0.02
Peak knee internal rotation angle (°)	12.2	5.8	1.1	0.97	0.95-0.98	1.6	0.82	0.69-0.90	0.85	0.70-0.93	0.05
Knee valgus angle at IC (°)	5.5	3.5	1.2	0.92	0.87-0.96	1.7	0.55	0.29-0.73	0.55	0.24-0.76	0.06
Peak knee valgus angle (°)	11.8	5.3	1.4	0.95	0.92-0.97	2.5	0.64	0.42-0.79	0.73	0.50-0.86	0.06
Knee flexion angle at IC (°)	21.8	7.5	2.7	0.90	0.83-0.94	2.2	0.68	0.48-0.82	0.60	0.31-0.79	0.06
Peak knee flexion angle (°)	60.5	6.9	4.1	0.75	0.56-0.85	2.6	0.63	0.41-0.79	0.65	0.38-0.82	0.09
Ankle internal rotation angle at IC (°)	-5.2	5.4	1.7	0.93	0.88-0.96	1.3	0.83	0.70-0.90	0.83	0.68-0.92	0.04
Peak ankle internal rotation angle (°)	-2.4	5.9	2.5	0.88	0.79-0.93	1.6	0.83	0.70-0.90	0.84	0.69-0.92	0.05
Ankle inversion angle at IC (°)	4.8	6.6	2.0	0.94	0.90-0.97	2.7	0.61	0.37-0.77	0.60	0.31-0.79	0.09
Peak ankle inversion angle (°)	28.4	5.9	1.4	0.95	0.92-0.97	2.1	0.71	0.49-0.84	0.74	0.51-0.87	0.21*
Ankle plantarflexion at IC (°)	8.0	12.1	5.6	0.90	0.83-0.94	4.0	0.77	0.60-0.87	0.77	0.57-0.89	0.15*
Peak ankle plantarflexion angle (°)	26.3	9.3	2.3	0.95	0.91-0.97	2.7	0.80	0.66-0.89	0.70	0.46-0.85	0.01
Peak ankle dorsiflexion angle (°)	23.2	6.8	1.8	0.94	0.89-0.96	2.5	0.69	0.48-0.82	0.54	0.22-0.75	0.09
Medial knee displacement (cm)	3.7	2.0	0.6	0.93	0.88-0.96	0.7	0.65	0.44-0.80	0.59	0.28-0.78	0.11
Kinetics											
Peak hip internal rotation moment (Nm)	43.8	14.6	5.3	0.89	0.82-0.94	5.7	0.68	0.47-0.81	0.69	0.43-0.84	0.04
Peak hip adduction moment (Nm)	50.4	82.5	15.8	0.97	0.96-0.98	24.0	0.85	0.74-0.92	0.81	0.64-0.91	0.08
Peak hip flexion moment (Nm)	243.3	95.3	19.6	0.96	0.93-0.98	35.8	0.77	0.60-0.87	0.83	0.68-0.92	0.05
Peak knee external rotation moment (Nm)	27.4	13.5	3.3	0.95	0.91-0.97	4.6	0.74	0.56-0.85	0.63	0.36-0.81	0.06
Peak knee abduction moment (Nm)	120.3	39.5	13.0	0.90	0.82-0.94	13.6	0.72	0.52-0.84	0.59	0.29-0.78	0.17*
Peak knee flexion moment (Nm)	167.3	45.2	12.3	0.94	0.89-0.96	13.0	0.80	0.66-0.89	0.74	0.52-0.87	0.00
Peak ankle internal rotation moment (Nm)	63.9	18.2	6.1	0.91	0.85-0.95	6.7	0.71	0.51-0.84	0.72	0.49-0.86	0.19*
Peak ankle eversion moment (Nm)	36.5	15.4	4.6	0.93	0.88-0.96	4.4	0.76	0.59-0.86	0.70	0.45-0.84	0.05
Peak ankle dorsiflexion moment (Nm)	164.3	30.3	6.6	0.96	0.94-0.98	7.5	0.86	0.75-0.92	0.82	0.65-0.91	0.10
Ground Reaction Forces											
Peak vertical GRF (N)	1543.2	441.8	93.3	0.97	0.94-0.98	88.7	0.92	0.85-0.96	0.89	0.78-0.95	0.10
Peak posterior shear force (N)	439.1	159.2	47.2	0.94	0.90-0.97	53.1	0.77	0.60-0.87	0.76	0.54-0.88	0.15*
Peak medial shear force (N)	738.5	154.3	42.7	0.94	0.90-0.96	37.5	0.85	0.73-0.92	0.81	0.63-0.90	0.10

SD: standard deviation; IC: initial contact; GRF: ground reaction force

* $p<0.05$, significant difference between the mean measurements of session 1 and 2.


(a)
Hip flexion
Hip abduction
Hip internal rotation
Knee flexion
Knee valgus
Knee internal rotation
Ankle dorsiflexion
Ankle eversion
Ankle internal rotation
Joint angle (deg)
Handball
Football
Stance (%)
(b)
Hip flexion
Hip abduction
Hip internal rotation
Knee flexion
Knee abduction
Knee internal rotation
Ankle dorsiflexion
Ankle eversion
Ankle internal rotation
Joint moment (Nm)
Handball
Football
Stance (%)

(c)

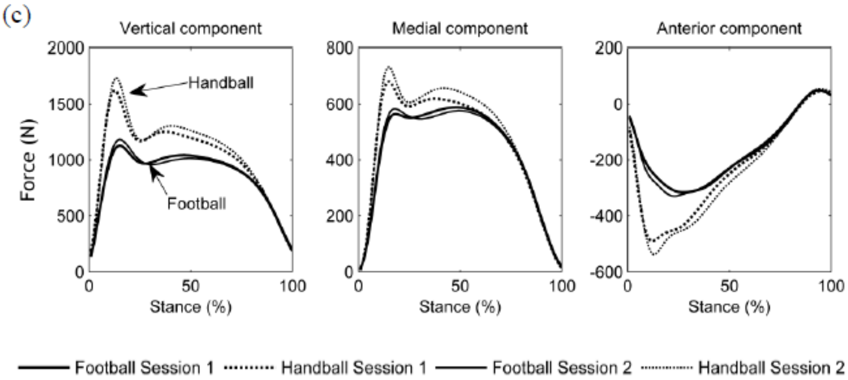


Figure 19. The time courses of mean measurements of handball and football athletes from session one and two, for (a) joint kinematics, (b) joint kinetics and (c) ground reaction forces

All the discrete biomechanical variables achieved good to excellent reliability in both within- and between-session analyses. The between-session rank correlations were moderate to strong, implying that the measurements could reliably reproduce the ranking of individuals in multiple-session studies. Furthermore, we found adequate reliability to be attained from three trials. Handball- and football-specific sidestep cutting tasks showed good to excellent reliability level in most of the variables, except the between-session reliability of a few frontal plane biomechanical variables in the football-specific sidestep cutting task. In general, the sidestep cutting task could reliably measure lower limb biomechanics.

Importantly, we found that increasing the number of trials from three to five trials only slightly improved reliability of the measurements, even if these tasks can be considered technically demanding. Thus, adequate reliability can be attained using only three trials. However, it should be noted that we required the athlete to have at least three practice trials before the official trials. In some cases, the athlete needed up to five practice trials to become familiarized and confident with the task.

As for the VDJ task, the between-session ICCs were lower than the within-session reliability. Similar findings have been reported in 3D motion analysis studies of gait, running, and VDJ (Kadaba et al., 1989; Ferber et al., 2002; Ford et al., 2007). Although several explanations exist, such as the differences in movement execution, this phenomenon can likely be explained predominantly by differences in marker placement between sessions (Kadaba et al., 1989;

Steinwender et al., 2000). Nevertheless, the between-session ICC values attain good reliability level for all discrete variables (Table 6). Moreover, the moderate to strong between-session rank correlations implies that all the discrete variables can provide a consistent ranking of athletes, which is essential for reliable screening of athletes. Coupled with the considerably small between-session typical error and effect size, the two sport-specific sidestep cutting tasks can generally provide reliable within- and between-session biomechanical measurements.

The sport-specific sidestep cutting tasks investigated in the current study showed similar reliability characteristics to the other motion tasks, except the between-session peak knee flexion angle. The between-session peak flexion angle has been found more reliable in gait (ICC: 0.80) (McGinley et al., 2009) and running (ICC: 0.93) (Ferber et al., 2002), than VDJ (ICC: 0.62) (Ford et al., 2007) and sidestep cutting (ICC: 0.63). In contrast to sidestep cutting and vertical drop jumps, gait and running are daily life motions which have been executed and developed from early childhood (Chappell & Limpisvasti, 2008; Desloovere et al., 2010; Sutherland, 1997). Therefore, the higher reliability of gait and running is likely the result of higher movement familiarization.

The two sidestep cutting tasks were found to have comparable reliability in most of the variables, but a few variables showed lower between-session reliability for the football athletes, especially on the non-dominant leg (Table 5). These were mainly frontal plane biomechanical variables of the hip and knee. A possible reason for this difference is that the faking maneuver is a fundamental motion in handball (Zebis et al., 2009) and performed repeatedly in every training session, which in turn may enhance movement consistency. Furthermore, McLean et al. (2004b) reported that a static defender could provoke the athlete to change direction more rapidly during cutting, and thereby affect frontal plane biomechanics of the lower limb. In line with the findings of McLean et al. (2004b), we found that the protocol including a static defender induced larger medial ground reaction force, larger hip abduction angle and knee valgus angle (Figure 19a and 19c). The static defender could thus potentially limit the possibility for movement variability, and thereby enhance reliability.

Implications

Paper I and II reported the reliability of biomechanical variables in VDJ and two sport-specific screening tasks. Reliable measurement are crucial in screening task, as well as in any other scientific studies such as e.g. prospective cohort studies. If the measurement is unreliable, it is difficult to detect differences that may exist between athletes (Hayen et al., 2007). In the current thesis, the reliability of measurements that may be important to understand ACL injury causation were comprehensively reported. Other researchers will be able to use these results to compute other reliability measures, such us minimal detectable change. In addition, this information can be important for studies having different research focus, not limited to ACL injury, such as ankle instability (Delahunt et al., 2006) and patellofemoral pain (Boling et al., 2009).

The approaches of quantifying and interpreting reliability in **Paper I and II** are traditional and well accepted. ICC is a common method for assessing the reliability of a measurement (Fleiss et al., 1986; Weir, 2005; Hayen et al., 2007). Previous studies used similar approaches to investigate the reliability of a task, and hence, to judge the suitability for assessing injury risk (Ford et al., 2007; Malfait et al., 2014; Sankey et al., 2015).

In the two tasks, both between-session ICC values and Spearman's rank correlations observed for the various measures of biomechanical variables were generally classified to be good to excellent or strongly positive, according to the generally accepted interpretations (Fleiss, 1986; Zou et al., 2003). However, in a screening perspective, it is still unclear how accurately we can identify athletes with increased risk of ACL injury. To better illustrate the consequences for screening, we did a sub-analysis from calculating the classification agreement of a target variable between session one and two. We assumed the top 30% athletes were high risk in session one, before subsequently computing the portion of the high-risk athletes being reclassified as high-risk in session two.

Based on our data from the VDJ reliability study (**Paper I**), we evaluated the number of athletes that would have been picked out among the top 30% (n=12) athletes of having high peak knee abduction moment in both sessions. Although the peak knee abduction moments achieved good between-session ICC (0.69) and moderately positive Spearman's rank correlation (0.72), 25% of the high-risk athletes (four out of 12 athletes) in session one failed to be reclassified as high-risk in session two.

The same calculation was imposed on the data from the sidestep cutting reliability study (**Paper II**). When using peak knee abduction moments (good between-session ICC (0.72) and moderately positive Spearman's rank correlation (0.59)) as the target variable, 50% of the high-risk athletes (six out of 12 athletes) in session one failed to be reclassified in the top 30% in session two.

This sub-analysis also provides insight to the evaluation of results from a prospective cohort study, because the between-session misclassification limits the ability to determine whether a measured variable is indeed a risk factor for injury. The between-session misclassification also implies the deteriorated sensitivity of identifying the high-risk athletes in a retest. After investigating the reliability by a continuous variable approach (ICC and Spearman's rank correlation), the reliability of a screening task should also be examined by a categorical variable approach (Kappa scores). To carry out reliability of a categorical screening results, the true risk classification (high/low risk) or injury record (injury/non-injury) are needed. However, this information are unavailable for this thesis. Obviously, it is necessary to conduct studies where the injury risk is known, in order to report the classification agreement within- and between-session.

Methodological concerns

The effect of thigh marker placement on knee valgus angles in vertical drop jumps and sidestep cutting (Paper III)

Substantial differences were observed between the different marker clusters (Figure 20). For sidestep cutting, the valgus angle at initial contact ranged from 0.7° ± 3.3° (mean ± SD) to 6.2° ± 3.4°, and peak valgus angle ranged from 9.0° ± 4.0° to 13.7° ± 5.4° with the different thigh marker clusters ($p \leq 0.05$) (Table 7). Twenty-five out of 30 marker cluster comparisons were significantly different ($p \leq 0.05$) (Table 8). For VDJ, the valgus angle at initial contact ranged from -5.0° ± 3.5° to 2.1° ± 3.0°, and peak valgus angle ranged from 1.9° ± 2.8° to 15.8° ± 5.8° with the different thigh marker clusters (Table 7). Twenty-three out of 30 marker cluster comparisons were significantly different ($p \leq 0.05$) (Table 8). The duration of the contact phase was 0.32 s ± 0.07 s for sidestep cutting and 0.61 s ± 0.11 s for VDJ.

In both sidestep cutting and VDJ, the ranking of trials based on knee valgus angles were consistent between marker clusters II, III and IV, as well as between marker clusters V and VI (Table 9). Marker cluster I was distinctly different from all the other marker clusters in term of both the magnitude and the trial ranking by knee valgus measurements. For the indirect cross-talk measure, the correlation between the knee flexion-extension and varus-valgus angles varied from 0.28 to 0.64 in sidestep cutting and from 0.27 to 0.72 in VDJ (Table 7).

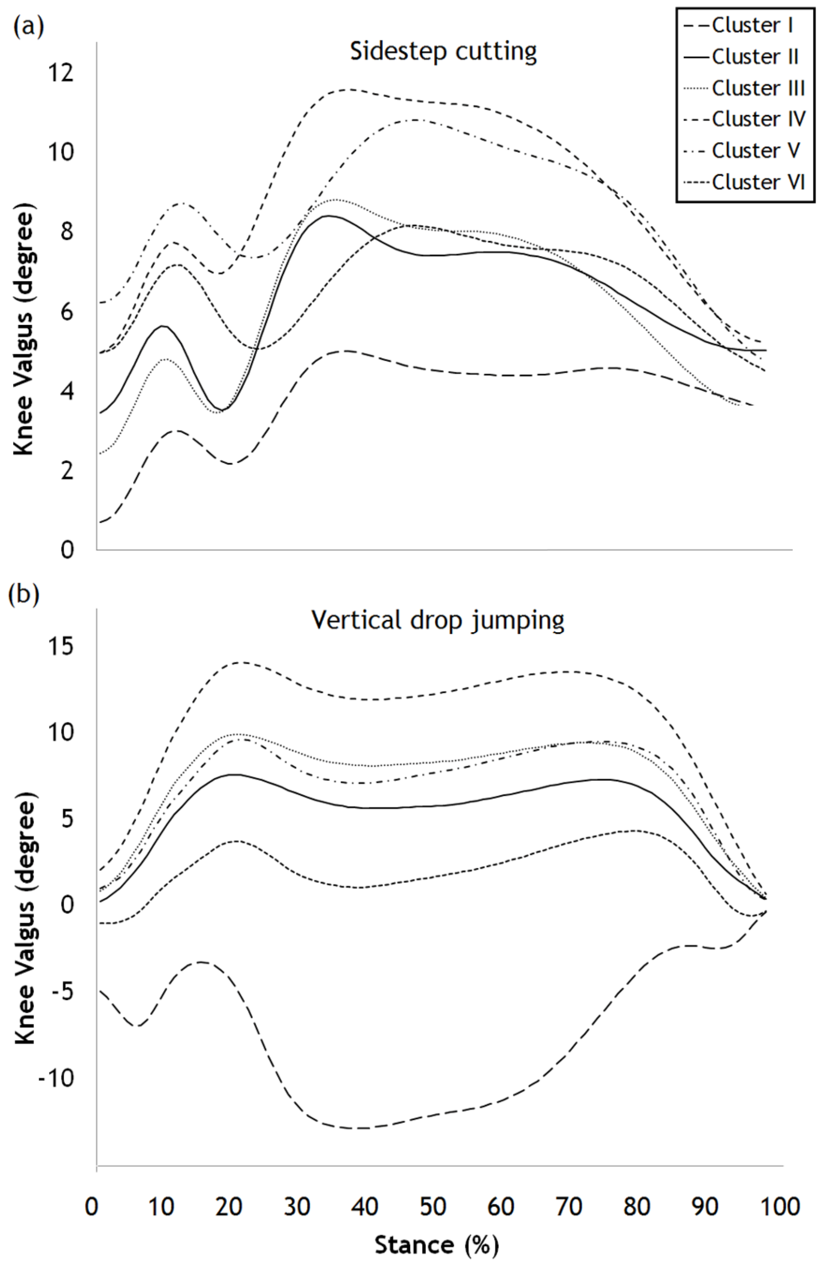


Figure 20. Mean profile of knee valgus angles during stance phase in (a) sidestep cutting and (b) vertical drop jumping

Table 7. The mean (SD) of the outcome variables in the six different thigh marker clusters, and the correlation between the knee flexion-extension and varus-valgus angle profiles

Outcome variables	Marker cluster						Peak flexion
	I	II	III	IV	V	VI	
Sidestep cutting							62.0 (8.4)
Valgus at IC (deg)	0.7 (3.3)	3.5 (2.6)	2.4 (3.0)	5.0 (3.1)	6.2 (3.4)	5.0 (3.4)	
Peak valgus (deg)	9.0 (4.0)	10.5 (4.1)	10.6 (4.5)	13.5 (4.6)	13.7 (5.4)	11.4 (5.5)	
Correlation	0.31	0.38	0.57	0.64	0.43	0.28	
Vertical drop jumping							107.4 (15.2)
Valgus at IC (deg)	-5.0 (3.5)	0.24 (3.0)	0.9 (3.2)	2.1 (3.0)	1.0 (3.5)	-1.0 (3.6)	
Peak valgus (deg)	1.9 (2.8)	9.5 (5.0)	11.6 (5.4)	15.8 (5.8)	12.0 (5.3)	6.9 (5.3)	
Correlation	0.63	0.53	0.59	0.72	0.51	0.27	

Table 8. P-value for the post-hoc t-test with Bonferroni correction for different combinations of the six thigh skin marker clusters in sidestep cutting and vertical drop jumping (IC: the valgus angle at initial contact; Peak: peak valgus angle)

Marker cluster	II		III		IV		V		VI	
	IC	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**
Sidestep cutting										
I	<0.01	.038	<0.01	.013	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01
II			.032	1.00*	<0.01	<0.01	<0.01	<0.01	<0.01	.387*
III					<0.01	<0.01	<0.01	<0.01	<0.01	.853*
IV							.002	1.00*	1.00*	.001
V									.002	<0.01
Vertical drop jumping										
I	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01	<0.01
II			1.00*	.021	.001	<0.01	1.00*	.002	.068*	.001
III					.110*	<0.01	1.00*	1.00*	<0.01	<0.01
IV							.320*	<0.01	<0.01	<0.01
V									<0.01	<0.01

** No significant difference on the outcome variable of two respective marker clusters in post-hoc t-tests with Bonferroni correction ($p \leq 0.05$)*

Table 9. Spearman's rank correlation coefficients for the six different combinations of thigh skin marker clusters in sidestep cutting and vertical drop jumping (IC: the valgus angle at initial contact; Peak: peak valgus angle)

Marker cluster	II		III		IV		V		VI	
	IC	Peak	IC	Peak	IC	Peak	IC	Peak	IC	Peak
Sidestep cutting										
I	.735	.661	.631	.625	.630	.585	.744	.732	.817*	.753
II			.930*	.974*	.944*	.930*	.711	.671	.678	.589
III					.918*	.957*	.681	.683	.613	.574
IV							.756	.710	.666	.588
V									.965*	.966*
Vertical drop jumping										
I	.781	.543	.762	.505	.691	.419	.684	.699	.726	.778
II			.970*	.989*	.960*	.943*	.778	.783	.739	.673
III					.947*	.966*	.787	.782	.739	.656
IV							.810*	.738	.735	.586
V									.972*	.947*

**Coefficient > 0.8 corresponds to strongly positive correlated and consistent.*

Notes: All the Spearman's rank correlation coefficients achieved significant level (p≤0.05).

The results indicate that the choice of thigh marker cluster can have a substantial impact on the magnitude of knee valgus angle, as well as the trial ranking. A standardized thigh marker cluster, including clear guidelines for placing non-anatomical markers, is needed to minimize the variation of the measurement, and to enable direct comparisons of motion data between different studies.

For the VDJ task, there was a discrepancy in the calculated magnitude of peak knee valgus angles between the six marker clusters up to 13.9°. For certain pairs of marker clusters, such as marker cluster I and III, the rank correlation coefficient was as low as 0.505 in the peak valgus angle in vertical drop jumping. Hence, different trials were identified as representing poor knee control. We observed large differences between marker clusters in the middle stance phase of the vertical drop jump (Figure 20b). The average difference between cluster I and IV at this point was as large as 25°. However, we observed smaller differences at initial contact when the knee was more extended. Furthermore, the knee valgus measurement at initial contact was more consistent (Table 9 and Figure 20b). The valgus angle at initial contact would be the most reliable variable at which to compare frontal-plane knee angles across different marker sets.

Marker cluster I was substantially different from the other marker clusters in both sidestep cutting and vertical drop jumping. Surprisingly, the only difference between marker cluster I and II was the non-anatomical marker on the anterior thigh, which was approximately five cm apart. The observed difference in valgus angles can possibly be explained by previous studies reporting that markers along the proximal-distal direction on anterior thigh showed different magnitude of medio-lateral STA (Akbarshahi et al., 2010; Camomilla et al., 2013), in particular for a jump-landing task (Monnet et al., 2012). Akbarshahi and co-workers (2010) showed that a distal anterior thigh marker has less medio-lateral STA than a more proximal anterior thigh marker in hip axial rotation and step-up motion. It should be noticed that some methods will use asymmetrical placement of the mid-thigh markers, such as Plug-in-Gait (Vicon®, 2002), in order to optimize the automatic marker tracking process. In such cases, bilateral comparison on lower limb kinematics should be carefully interpreted.

Marker clusters V and VI, which both included the HJC, gave consistent magnitude and trial ranking result with each other, but were less consistent with the other marker clusters. The inclusion of the HJC may introduce a systematic difference to the thigh technical coordinate system. In this study, the HJC was estimated based on a regression model using anatomical landmarks of the pelvis (Bell et al., 1990). The average error in this HJC regression model is about 15 mm (Sangeux et al., 2011). This is still likely to be a better proximal anatomical marker than the Greater Trochanter where a STA close to 35 mm has been reported (Camomilla et al., 2013; Cappozzo et al., 1996). The accuracy of HJC estimation may even be improved by using the functional method for the estimation of HJC (Leardini et al., 1999; Della Croce et al., 2005).

The cross-talk effect is the erroneous coupling between knee varus-valgus and flexion-extension because of the misalignment of knee joint flexion-extension axis (Schache et al., 2006). In our study, the flexion-extension axis definition was equal for all marker clusters so we would assume the same degree of cross-talk for all the clusters. However, the correlation coefficient varied from 0.28 to 0.64 in sidestep cutting and from 0.27 to 0.72 in vertical drop jumping (Table 7). This implies that the cross-talk effect is not only sensitive to the definition of knee joint flexion-extension axis, but also the marker clusters. For instance, lateral movement of a thigh marker away from the midline of the body will result in an external rotation of the thigh coordinate system along the longitudinal axis, which will again result in increasing measured knee valgus angles when the knee is flexed. Hence, the placement of all markers is crucial, not only had those placed on bony landmarks.

A true gold standard was not present in this study. Therefore, a robust recommendation on which marker cluster to choose cannot be made. Biplanar videoradiography techniques have previously been utilized to provide a non-invasive gold standard in measuring lower-extremity limb motion during jump landings, but due to the limited recording volume, it is currently not possible to measure sport-specific sidestep cutting similar to those in our test protocols, nor the whole landing phase of vertical drop jumps (Myers et al., 2011; Torry et al., 2013; Miranda et al., 2013). Myers et al. (2011) reported peak knee valgus in the first 0.1 s after initial contact for female during jump-landings of 1.6° ± 0.9° using biplanar videoradiography technique. Using the same kinematics calculation approach (Grood & Suntay, 1983), the peak knee valgus angle was observably smaller in value and standard deviation compared to the present study (Table 7) as well as previous studies using skin marker-based motion analysis, which have reported knee valgus angles close to 30° (Ford et al., 2003; Chappell & Limpisvasti, 2008). Nevertheless, it is important to realize that although the measured peak knee valgus angles are likely to be overestimated, the measurements can still be valuable in a risk factor context provided that the trial ranking is consistent.

The effect of overhead target on the lower limb biomechanics during a vertical drop jump test in elite female athletes (Paper IV)

Jump height increased by 5.8% when athletes reached for an overhead target, corresponding to a medium effect size (Table 10). Of all variables studied (Table 10), a significant interaction between leg dominance and task was only observed for medial knee displacement ($p=0.02$). We therefore analyzed the dominant and non-dominant leg results separately for this variable, whereas data for the dominant and non-dominant leg were combined for the remaining variables. The addition of an overhead target resulted in a significant change in the majority of the biomechanical variables examined; however, the effect sizes were small (Table 10). We observed a strong rank correlation between the two tasks in 23 out of the 32 variables, and a moderate correlation in the remaining nine variables (Table 10). The medial knee displacement variable had the lowest rank correlation.

Table 10. Comparison of biomechanical variables during vertical drop jumps in target and non-target conditions (n=523). Data are shown as means, standard deviations (SD), mean differences (percent change in parentheses) with 95% confidence interval (CI), effect sizes (Cohen's d) and Spearman's rank correlation coefficients. Positive values denote an increased score with the addition of an overhead target; statistically significant differences are shown in bold.

	Non-target condition	Target condition	Comparison		
	Mean±SD	**Mean±SD**	**Mean difference (95% CI)**	**Effect size**	**Spearman's rank correlation coefficient**
Jump height (cm)	41.6±4.8	44.0±5.0	**2.4 (2.2 to 2.7)**	0.51	0.84
Kinematics					
Hip flexion angle at IC (°)	48.3±8.5	49.9±8.4	**1.6 (1.3 to 2.0)**	0.20	0.79
Peak hip flexion angle (°)	83.0±15.6	78.2±15.5	**-4.8 (-5.4 to -4.4)**	0.31	0.85
Hip abduction angle at IC (°)	5.3±3.6	5.0±3.8	-0.3 (-0.4 to -0.1)	0.07	0.86
Peak hip abduction angle (°)	7.6±3.9	7.2±3.9	**-0.4 (-0.6 to -0.4)**	0.12	0.86
Hip internal rotation angle at IC (°)	-0.6±5.3	-0.5±5.4	0.1 (-0.1 to 0.2)	0.01	0.90
Peak hip internal rotation angle (°)	4.6±5.6	4.4±5.6	-0.2 (-0.3 to 0.0)	0.03	0.90
Knee flexion angle at IC (°)	32.3±7.9	34.9±8.1	**2.6 (2.3 to 2.9)**	0.33	0.81
Peak knee flexion angle (°)	93.5±14.8	89.1±13.5	**-4.4 (-5.0 to -3.9)**	0.31	0.80
Range of knee flexion (°)	61.2±16.8	54.2±15.5	**-7.0(-8.4 to -5.6)**	0.43	0.79
Knee valgus angle at IC (°)	-2.0±4.2	-1.8±4.4	0.2 (0.18 to 0.4)	0.07	0.92
Peak knee valgus angle (°)	5.1±5.2	5.6±5.2	**0.5 (0.4 to 0.6)**	0.10	0.95
Knee internal rotation angle at IC (°)	-0.5±5.4	0.3±5.7	**0.8 (0.6 to 1.0)**	0.15	0.83
Peak knee internal rotation angle (°)	10.4±6.1	9.9±5.9	-0.5 (-0.6 to -0.3)	0.07	0.94
Ankle plantarflexion at IC (°)	10.9±9.7	11.3±9.1	0.4 (0.1 to 0.8)	0.04	0.77
Peak ankle plantarflexion angle (°)	25.1±7.3	25.1±7.1	0.02 (-0.2 to 0.2)	<0.01	0.89
Ankle inversion angle at IC (°)	9.3±5.2	8.5±5.2	**-0.8 (-1.0 to -0.6)**	0.16	0.83
Peak ankle inversion angle (°)	10.8±4.6	10.2±4.7	**-0.6 (-0.7 to -0.4)**	0.11	0.85
Ankle internal rotation angle at IC (°)	0.5±6.6	-0.1±6.7	**-0.6 (-0.7 to -0.5)**	0.09	0.87
Peak ankle internal rotation angle (°)	6.1±6.1	5.6±6.2	-0.5 (-0.6 to -0.3)	0.08	0.80
Medial knee displacement on the dominant side (cm) [a]	2.2±1.3	2.4±1.6	**0.2 (0.1 to 0.3)**	0.15	0.58
Medial knee displacement on the non-dominant side (cm) [a]	2.1±1.4	2.0±1.6	-0.08 (-0.1 to 0.0)	0.06	0.63
Kinetics					
Peak vertical GRF (N)	1294.6±391.6	1359.5±400.7	**64.9 (48.9 to 80.7)**	0.16	0.77
Peak hip flexion moment (Nm)	203.3±46.3	208.2±52.4	**4.9 (2.9 to 6.8)**	0.10	0.80
Peak hip abduction moment (Nm)	40.0±17.1	41.4±17.8	1.4 (0.8 to 2.1)	0.08	0.77
Peak hip internal rotation moment (Nm)	20.1±9.5	20.8±10.1	0.7 (0.3 to 1.0)	0.07	0.81
Peak knee flexion moment (Nm)	141.3±37.3	150.1±40.3	**8.8 (7.5 to 10.1)**	0.23	0.87
Peak knee abduction moment (Nm)	20.5±10.7	21.7±12.1	**1.2 (0.7 to 1.6)**	0.10	0.78
Peak knee internal rotation moment (Nm)	10.2±6.9	11.1±7.7	**0.9 (0.6 to 1.1)**	0.12	0.83
Peak ankle dorsiflexion moment (Nm)	112.7±31.0	120.4±4.8	**7.7 (6.5 to 8.8)**	0.23	0.85
Peak ankle inversion moment (Nm)	1.8±4.0	2.0±4.6	0.2 (0.0 to 0.3)	0.04	0.78
Peak ankle internal rotation moment (Nm)	16.5±8.7	17.5±9.7	**1.0 (0.7 to 1.3)**	0.11	0.85

[a] *Separated by leg because of the significant interaction of leg dominance and condition.*

Our results are generally consistent with previous studies, including the increase in jump height and joint loading when introducing an overhead target (Ford et al., 2005a; Wulf & Dufek, 2009). However, the changes in kinematics and kinetics are likely to be clinically insignificant, as indicated by the small effect sizes.

Comparing the two conditions, we recorded a 5.0% increase in peak vertical ground reaction force (Table 10), which corresponds well to the increase in jump height. This demonstrates that an external motivation enhance the effort even if we provided strong verbal encouragement for the non-target task and also asked the athletes to repeat the trial if a sub-maximal effort was suspected.

According to the definition of frontal plane knee control of Nilstad and co-workers (2014b), the target condition seemed to slightly decrease frontal plane knee control as reflected by the increase in the peak knee valgus angle from 5.1° to 5.6°, and dominant leg medial knee displacement from 2.2 to 2.4 cm (Table 10). The change in frontal plane knee control is of particular interest as such variables have been suggested to be associated with the risk of ACL injury in a mixed-sport cohort of female athletes (Hewett et al., 2005). The target condition resulted in changes in biomechanical variables of particular interest in ACL injury risk. However, the small effect sizes suggest that the kinematic differences have little clinical relevance.

The overhead target generated a stiffer landing, as we recorded a 4.4° decrease in peak knee flexion angle, a 7.0° decrease in the range of knee flexion and a 65 N increase in peak vertical ground reaction force (Table 10). It is known that higher vertical ground reaction force and extended knee position in stiff landings will increase ACL strain due to the increased anterior tibial translation (Markolf et al., 1995; Myers et al., 2011). In contrast, soft landings will allow the vertical ground reaction force to be dissipated over a larger range of knee flexion than stiff landings (Devita & Skelly, 1992; Myers et al., 2011; Pollard et al., 2010), and are therefore advocated as a preventive strategy for reducing ACL injuries in successful injury prevention programs (Mandelbaum et al., 2005; Myer et al., 2004; Myklebust et al., 2013; Taylor et al., 2015). Our results are consistent with previous studies assessing differences between soft and stiff landings (Devita & Skelly, 1992; Myers et al., 2011; Pollard et al., 2010). However, again the effect sizes were small, suggesting that these differences are unlikely to be clinically relevant.

The rank correlations observed for the various measures of lower limb biomechanics between the two conditions were generally interpreted as strong to moderate. Even so, when applying these tasks as screening tests for ACL injury risk, they must be interpreted with caution. Hewett et al. (2005) have suggested that a high knee abduction (valgus) moment during a drop jump task without an overhead target is a strong predictor for ACL injury risk. The rank correlation for knee abduction moment between the two tasks in our study was 0.78, which is generally considered as moderate to strong. However, if we consider the 20% of athletes (n=105) with the highest magnitude of peak knee abduction moment in the non-target condition as being at risk, only 67% of these were also included in the top quintile in the target condition. Although biomechanical variables were highly correlated between two tasks, the association between lower limb biomechanics and the risk of ACL injury could be different between the two conditions.

There are some limitations to consider when interpreting the results of the current study. Marker placement and soft tissue artifacts are well-known sources of error in a skin marker-based motion analysis (Leardini et al., 2005; Miranda et al., 2013). However, since marker placement was standardized, these errors are expected to be similar for the two tasks and unlikely to have affected our findings. The fact that we did not randomize the task order means that fatigue may have been induced in the target condition, which was always performed after the non-target condition. However, the fact that they were able to jump 2.4 cm higher in the target condition, implies that fatigue effects were less likely. Another potential limitation is that reaching a horizontal bar by the head may pose less of a challenge for neuromuscular control than e.g. grasping an overhead ball. Lastly, it is still unknown how the other demographic factors, such as sex and age, could have affected the results. Therefore, the generalizability of these results to athletes of a different skill level, sport or even injury history is unknown.

Implications

Paper III identified significant knee valgus measurement discrepancies between different thigh marker clusters. Moreover, the inconsistency of trial rankings challenges results from previous studies investigating knee valgus angles and its implication for ACL injury risk. Our results clearly demonstrate the importance of standardizing marker clusters for marker-based 3D motion analysis. A standardized marker clusters is therefore necessary to procedure consistent screening results.

Ideally, an optimal thigh marker cluster should be developed based on a gold standard. The validity of knee valgus measurements could be investigated with true bony motion as a gold standard. Invasive bone-pin markers can be used to measure the bony motion (Reinschmidt et al, 1997a; Benoit et al., 2006), however it is an invasive procedure, which may cause discomfort and the anesthetics may alter the movement. A non-invasive bi-plane fluoroscopy could be an alternative. However, the feasibility of performing repeated vigorous sport motions is currently limited, because of small capture volume, time-consuming bone tracking procedures and radiation exposure.

The findings in **Paper IV** add to the knowledge of the specific considerations when using a VDJ task. The VDJ can potentially be used for mass screening of ACL injury risk because it is easy to setup and time-efficient. An overhead target will enhance performance, but does not lead to clinically relevant changes in biomechanical variables assumed to be associated with injury. Therefore, it seems to be less important to include an overhead target in the VDJ test. However the ability of the test to identify players of increased risk for future ACL injury must ultimately be tested in a prospective cohort study.

Conclusions

Reliability

- **(Paper I)** The knee biomechanical variables of the vertical drop jump task are reliable in both within- and between-session analyses for elite female handball and football athletes. The moderate to strong positive between-session consistency of athlete measurements and rankings imply that the ranking of athletes based on knee biomechanics in the vertical drop jump task can be reliably reproduced, which is critical for injury risk screening purposes. The knee flexion angle varied considerably within-session in some athletes, however the task can reliably measure knee valgus angles and internal rotation angles, which are believed to be related to the ACL injury mechanism.

- **(Paper II)** The lower limb biomechanical variables of sport-specific sidestep cutting tasks are reliable in both within- and between-session analyses for elite female handball and football athletes. In addition, the between-session consistency of athlete rankings achieved moderate to strong positive rank correlation, implying that the measurements could reliably reproduce the ranking of individuals in multiple-session studies. Adequate reliability could be attained from three trials only, with only minor increase in reliability when adding more trials. Sport-specific sidestep cutting tasks are likely to be important in future ACL injury research as a high proportion of injuries takes place in such movements.

Methodological concerns

- **(Paper III)** The choice of thigh marker clusters affected the magnitude of the knee valgus angle in VDJ and sidestep cutting, as well as the trial ranking. A standardized thigh marker cluster, including clear guidelines for placing non-anatomical markers, is needed to minimize the variation of the measurement, and to enable direct comparisons of motion data between different studies.

- **(Paper IV)** Adding an overhead target to the VDJ test improves jump height, but generates only minor changes in lower limb biomechanics. In addition, athlete ranking based on the lower limb biomechanics were consistent between two conditions. Therefore, an overhead target is unlikely to change responses in biomechanical variables in elite female handball and football athletes.

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Paper I-IV

Paper I

Reliability of knee biomechanics during a vertical drop jump in elite female athletes

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3D motion analysis, knee kinematics and kinetics, ACL injury, risk screening, rank of individuals, repeatability.

Abstract (250 words)

The purpose of the study was to assess the within-session and between-session reliability of knee kinematics and kinetics in a vertical drop jump task among elite female handball and football athletes. Specifically, we aimed to quantify the within-session waveform consistency and between-session consistency of the subject ranking for a variety of knee kinematics and kinetics. Forty-one elite female handball and football (soccer) athletes were tested in two sessions. The reliability of three-dimensional knee biomechanical measurements was quantified by the intra-class correlation, Spearman's rank correlation, and typical error. All the selected discrete variables achieved excellent within-session reliability (ICC > 0.87). The typical error of valgus angles, internal rotation angles, and internal rotation moment was constant throughout the whole stance phase. For between-session reliability, the selected discrete variables achieved good to excellent reliability (ICC > 0.69), except peak internal rotation moment (ICC = 0.40). All between-session rank correlation coefficients ranged from 0.56 to 0.90. Most of the discrete variables achieved good to excellent reliability in both within-session and between-session analysis. Moreover, moderate to strong between-session consistency of subject rankings was found, implying that the measurements assessed during the vertical drop jump demonstrate sufficient reliability to be used in both single-session and multiple-session studies.

1. Introduction

Vertical drop jump tasks have been widely used in anterior cruciate ligament (ACL) injury-related research in the last decade. Three-dimensional knee kinematics and kinetics, quantified using marker-based motion analysis systems, have been used to identify potential risk factors for ACL injuries [1,2]. Furthermore, knee kinematics and kinetics in vertical drop jumps are utilized for ACL injury risk assessment [2,3] and evaluation of training interventions [4,5].

Previous research has investigated both within-session and between-session reliability of various knee biomechanical variables in vertical drop jump tasks [6,7]. Ford et al. [6] utilized the intra-class correlation coefficient (ICC) and typical error of various discrete biomechanical variables to quantify the between-session reliability. The majority of the knee kinematic and kinetic variables were shown to have fair to excellent reliability within- (ICC from 0.67 to 0.99) and between-sessions (ICC from 0.59 to 0.92) [6] in young female high school athletes. Malfait et al. [7] assessed the within-session reliability and showed that the knee valgus angle displayed a small variability of only 1.1°, while the variability of knee flexion angle was larger at 3.8° [7].

However, the low number of participants in the previous reliability studies is a major concern. Methodology studies of reliability in sports medicine suggest that such studies should contain a minimum of 40 subjects [8]. The reliability of vertical drop jump tasks have, up until now, only been investigated in very limited populations, i.e. one study on 8 recreational athletes [7] and one on 11 high school athletes [6]. Likewise, the reliability of medial knee displacement was only reported from a study with five subjects [9].

Furthermore, previous studies have not investigated the reliability of vertical drop jump task in homogenous elite populations. Elite female handball and football cohorts are of particular interest, knowing that the risk of sustaining ACL injuries is higher, compared with other groups of athletes [10].

The aim of the present study was to assess the within-session and between-session reliability of knee kinematics and kinetics in a vertical drop jump task among elite female handball and football athletes. Specifically, we aimed to quantify the within-session waveform error of measurements and between-session consistency of the subject measurements and rankings.

2. Methods

2.1 Subjects. Nineteen elite female handball and 22 elite female football (soccer) athletes (mean ± SD: 22 ± 4 yrs old, 168 ± 5 cm, 66 ± 8 kg) performed vertical drop jumps in our biomechanics laboratory. The Regional Ethics Committee approved the study and all subjects provided signed informed consent forms.

2.2 Sample size calculation. Sample size calculation was performed using the formula of Shoukri et al. [11]. The formula is specifically designed for reliability studies by setting the limit of the confidence interval width of the reliability coefficient. The width of the confidence interval was set to be 0.2 based on the reliability coefficient reported by Ford and colleagues [6]. Based on this, with three repeated trials and mean reliability coefficient value of 0.8, the formula gave a minimum sample size requirement of 37 subjects.

2.3 Design and protocol. Subjects were tested in two separate sessions, on average separated by two weeks. We instructed subjects to drop off a 30 cm box and perform a maximal jump upon landing with their feet on separate force platforms (AMTI LG6-4-1, Watertown, Massachusetts, USA). They were allowed to have three practice trials and at least three valid trials were collected for each player. At least two test operators observed the execution of the jump. If sub-maximal effort was suspected, or when jumping instead of dropping off the box (i.e.

increasing the vertical center of mass position at take-off from the box), we asked the subject to repeat the jump. Players were encouraged to jump with maximal effort for every jump.

Subjects wore indoor sport shoes, shorts and a sports bra. Thirty-seven reflective markers were attached over anatomical landmarks on the legs, arms and torso [12]. All marker positions were defined uniquely, also those not defined by anatomical landmarks. One experienced physiotherapist, with several years practice for marker placement, was employed for skin marker placement in both sessions.

We used a 480 Hz 16-camera system (Oqus 4, Qualisys, Gothenburg, Sweden) to capture motion, while we recorded ground reaction forces using two force platforms collecting at 960 Hz (AMTI LG6-4-1, Watertown, Massachusetts, USA). We calibrated the motion analysis system according to guidelines from the manufacturer, and calculated and tracked marker trajectories using the Qualisys Track Manager (Qualisys, Gothenburg, Sweden).

We defined the contact phase as the period where the unfiltered vertical ground reaction force exceeded 20 N. Marker trajectories and force data were filtered and interpolated using Woltring's smoothing spline in the cubic mode [13], using a 15 Hz cut-off [12].We calculated the hip joint center using the method proposed by Bell et al. [14], with the anterior-posterior position of the hip joint decided by the anterior-posterior position of the marker over the greater trochanter. Furthermore, we defined the knee joint center according to Davis [15], and the ankle joint center according to Eng & Winter [16]. Anatomical coordinate systems of the thigh and shank were determined from the static calibration trials. We defined the vertical axis in the direction from the distal to the proximal joint center, while the anterio-posterior axis was defined perpendicular to the vertical axis with no mediolateral component. The third axis was the cross product of the vertical and antero-posterior axes. Consequently, all segments had neutral internal/external rotation in the static calibration trial. We obtained technical, dynamic thigh and shank segment coordinate systems using an optimization procedure involving singular value decomposition [17].

We estimated inertia parameters based on 46 measures of segment heights, perimeters and widths using a modified Yeadon's method [18], with hand and foot parameters calculated with the method of Zatsiorsky & Seluyanov [19]. We calculated hip and knee joint moments with inverse dynamics using recursive Newton-Euler equations of motion as described by Davis et al. [15] and projected onto the three rotational axes of the joint according to the joint coordinate system standard [20].

We used the Grood & Suntay [20] convention for calculating joint angles from the marker-based motion analysis. We calculated medial knee position as the perpendicular distance between the knee joint center and the line joining the ankle and hip joint centers, projected on the frontal plane. The difference between the perpendicular position at the initial foot contact and the peak value was defined as the medial knee displacement. An advantage of this convention compared with a pure knee separation measure is that we can assess knee control individually for the left and right leg. We ran all calculations using custom Matlab scripts (MathWorks Inc., Natick, Massachusetts, USA).

2.4 Statistical analysis. For simplicity, only the measurements from the right leg were used for analyses. Each trial was time-normalized from 0% to 100% of the stance phase. For every time point, we calculated the typical error based on three trials from each subject. The typical error was calculated from the standard deviation of inter-trial differences divided by the square root of 2 [21]. The typical error represented the 52% of test-retest differences of a subject in the sample group [21]. Moreover, the mean curves and standard deviation were computed to represent the motion and between-subject variability.

We report the mean, standard deviation, and typical error of both within and between-session measurements. The ICC values for within-session (ICC(3,k)) and between-session (ICC(3,1)) were both computed. The ICC classifications of Fleiss [22] (less than 0.4, poor; between 0.4 and 0.75, fair to good; and greater than 0.75, excellent) were used to describe the range of ICC values. To assess the consistency of subject ranking between-sessions, Spearman's rank correlation coefficients were calculated based on the mean measurement from sessions 1 and 2. The classifications of Zou et al. [23] (greater than 0.5, moderately positive; greater than 0.8, strongly positive) were used to interpret the rank correlation coefficients. Paired t-tests were employed to assess the significant difference of the mean measurement between sessions. Cohen's *d* was computed to assess the effect size of the mean differences. Statistical significance was set at $p \leq 0.05$. Statistical analyses were performed using SPSS 18 (SPSS Inc., Chicago, IL, USA) and the statistics toolbox of Matlab (MathWorks Inc., Natick, MA, USA).

3. Results

All the selected discrete variables achieved excellent within-session reliability with all ICC values greater than 0.87 (Table 1). The typical errors were generally small. The errors related to the valgus angles were < 1.0°.

The within-session typical error of the knee valgus, internal rotation angle and internal rotation moment was relatively constant throughout the whole stance phase (Fig 1 and 2). We observed an increase in typical error during the mid-stance for the knee flexion angle, medial knee displacement, knee flexion moment, and knee valgus moment. The maximal typical error for the vertical ground reaction force was found in the first 15% of the stance phase.

The between-session ICC values for most of the selected discrete variables achieved good to excellent between-session reliability (Table 1). However, peak internal rotation moment displayed only fair between-session reliability with an ICC value of 0.40. All the rank correlation coefficients demonstrated positive correlation on the between-session subject ranking. Peak flexion moment, peak valgus angle, peak internal rotation, medial knee displacement and jump height showed a strong between-session consistency with rank correlation coefficients greater than 0.8 [23]. The flexion angle at initial contact, peak flexion angle, knee flexion range of motion, peak internal rotation and the medial knee displacement were significantly different between sessions ($p<0.05$). However, all the Cohen's *d* values were smaller than 0.31, indicating that the effect sizes of the mean differences were small.

Differences in typical error waveforms were observed between sessions (Figure 1 and 2). The knee flexion angle, medial knee displacement, flexion moment and valgus moment had a higher typical error in the mid-stance phase in session 2.

4. Discussion

In the current study, we assessed the reliability of knee kinematics and kinetics in a vertical drop jump task in elite female handball and football athletes. Most of the discrete knee biomechanical variables achieved good to excellent reliability in both within-session and between-session analyses. Moreover, we found a strong between-session consistency of subject measurements and rankings, implying that the estimates could reliably reproduce the testing results in both single-session and multiple-session studies. Valgus and internal rotation angles were highly reliable while the knee flexion angle can vary considerably within-session in some subjects.

The within-session reliability (ICC 0.87 to 0.98) was, in general, better than between-session reliability (ICC 0.40 to 0.90). The slightly lower reliability between sessions is likely a result of variability in skin marker placement and changes in subject movement. We were surprised to see a relatively large typical error of knee flexion angle and medial knee displacement during mid-stance in session 2 (figure 1). The observed variation was likely a result of variability of subject movement. A further analysis showed that the large typical error was generated by two subjects who performed jumps with both high and low knee flexion within the same session. When removing the two subjects from the analysis, the typical error waveform in session two was similar to session one. Landing technique standardization or instruction may likely attenuate the variation in knee flexion angles [24,25]. In our testing protocol, the knee flexion angle was not standardized because we did not want to impose a specific jump landing technique unnatural to the subjects. Although it can be speculated that some jumps amongst the players with high variability should be excluded due to submaximal jumping performance, the jump heights were nearly identical. Hence, the jumps were correctly assessed to be valid trials according to our definition. Importantly, the knee valgus angles and internal rotation angles had a small and constant typical error throughout the whole stance phase in both sessions despite the variability in knee flexion.

The results obtained in this study are strikingly similar to those of Ford et al. [6]. With the exception of the internal rotation moments, the within and between session reliability for the two studies was close to identical (average ICC difference less than 0.1 for all common variables). In other words, good to excellent reliability of biomechanical measures have been obtained across two different biomechanical laboratories/protocols, including different cohorts of elite and reactional female high school athletes.

An increase in typical error appeared in all the kinetic variables between 0 to 20% of the stance phase in all three directions (Figure 2). This is consistent to previous findings from Malfait and co-workers [7] who also concluded that kinetic measurement shortly after initial contact are more variable. In contrast, they did not find a similar variation of the kinetic measurement in the mid-stance as found in our study. The larger sample size (n =41) in our study may potentially explain the greater kinetic variability which was absent in the aforementioned study (n = 8).

The ICC is commonly used to describe reliability, however, there is considerable confusion concerning both the calculation and interpretation of the ICC [26]. The ICC will give high reliability when the subject range is large, even if trial-to-trial variability is large [26,27]. Spearman's rank correlation will be unaffected by the range in the variable as it transforms the measurements to the ranking domain for the correlation calculation thus is less sensitive to between-subject variability. However, although Spearman's rank correlation coefficient can theoretically provide additional information on reliability, there was a good correspondence between the two measures in the current study.

Both ICC and Spearman's rank correlations observed for the various measures of lower limb biomechanics between the two sessions are generally satisfactory. This implies that the subject rankings could be reliably reproduced in multiple sessions. But from a screening perspective, the question is if repeated tests identify the same athletes as being at risk, i.e. as having outer-range test results. Hewett et al. [1] have suggested that a high knee abduction (valgus) moment during a vertical drop jump task is a strong predictor for ACL injury risk. When using peak knee abduction moments (Spearman's rank correlation: 0.72), only 75% of athletes (nine out of 12) included in the top 30% in session one were also included in the top 30% in session two. However, the accuracy of lower limb biomechanics for identifying the risk of ACL injury needs to be established from a prospective cohort study using injuries as the outcome.

The waveform reliability can be quantified by the typical error of every time point. The coefficient of multiple correlations (CMC) has been used to assess the waveform reliability [6]. However, CMC coefficient measures are sensitive to the data range, indicating lower reliability for smaller range of motion such as knee valgus motion [28]. Moreover, CMC is generally insensitive to systematic error [29]. An alternative to the CMC, the waveform reliability, can be quantified as the typical error of every time point. With this temporal presentation, the movement variability can be further described in a specific region such as initial contact or mid-stance. Using this approach, we could detect landing technique variation between session and attribute them to a specific phase of the movement. Furthermore, since the typical error has the same unit as the measurement, the variability can be directly related the measurement itself [30]. Future reliability studies may benefit from using this approach to present waveform reliability.

5. Conclusion

Our results suggest that biomechanical variables of the vertical drop jump task are reliable and can be used for research purposes. Additionally, the strong and positive between-session consistency of subject measurements and rankings imply that the ranking of athletes based on knee biomechanics in the vertical drop jump task can be reproduced reliably, which is critical for injury risk screening purposes. The vertical drop jump task can reliably measure knee valgus angles and internal rotation angles, which are believed to be related to the ACL injury mechanism. Furthermore, the new approach for calculating medial knee displacement was found to be reliable.

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Table 1. Reliability values for selected biomechanical measurements. SD: Standard deviation; TE: Typical Error; ICC: Intra-class correlation; CI: Confidence interval; IC: Initial Contact; ROM: Range of motion, GRF: Ground reaction force.
* $p<0.05$, significant difference between the mean measurements of session 1 and 2.

	Session 1					Session 2					Between Session					
Variables	Mean	SD	TE	ICC(3,k)	95% CI	Mean	SD	TE	ICC(3,k)	95% CI	TE	ICC(3,1)	95% CI	Spearman's rho	95% CI	Cohen's d
Kinematics																
Knee flexion angle at IC (°)	29.66	8.45	3.05	0.90	0.83-0.94	31.50	9.11	2.88	0.93	0.88-0.96	2.40	0.79	0.64-0.88	0.78	0.58-0.89	0.21*
Peak knee flexion angle (°)	108.32	14.66	3.19	0.95	0.92-0.97	112.89	15.12	5.47	0.92	0.87-0.96	4.78	0.74	0.57-0.85	0.72	0.47-0.86	0.31*
ROM of knee flexion angle (°)	100.09	14.64	3.69	0.94	0.90-0.97	103.61	15.16	5.54	0.91	0.85-0.95	4.00	0.76	0.60-0.87	0.76	0.55-0.88	0.24*
Knee valgus angle at IC (°)	-1.07	3.54	0.70	0.97	0.95-0.98	-1.09	4.09	0.61	0.98	0.97-0.99	1.11	0.75	0.58-0.86	0.69	0.44-0.84	<0.01
Peak knee valgus angle (°)	6.92	5.21	0.77	0.98	0.97-0.99	6.60	5.60	0.94	0.98	0.96-0.99	1.51	0.81	0.67-0.89	0.81	0.63-0.91	0.06
ROM of knee valgus angle (°)	10.69	3.40	0.92	0.92	0.86-0.95	10.94	3.65	0.98	0.94	0.91-0.97	1.06	0.81	0.66-0.89	0.79	0.59-0.89	0.07
Internal knee rotation angle at IC (°)	-0.25	4.86	1.26	0.94	0.91-0.97	0.003	5.83	1.27	0.95	0.92-0.97	1.58	0.75	0.58-0.86	0.73	0.49-0.86	0.05
Peak internal knee rotation angle (°)	9.57	5.69	1.04	0.98	0.73-0.99	10.78	7.23	1.26	0.98	0.71-0.99	1.80	0.84	0.73-0.91	0.85	0.70-0.93	0.19*
ROM of internal knee rotation angle (°)	16.02	5.17	1.79	0.91	0.86-0.95	16.78	6.99	1.91	0.94	0.90-0.97	2.28	0.73	0.55-0.85	0.77	0.56-0.88	0.12
Medial knee displacement (cm)	3.54	1.86	0.59	0.91	0.86-0.95	4.22	2.46	0.86	0.92	0.87-0.96	0.87	0.75	0.58-0.86	0.80	0.62-0.90	0.31*
Jump height (m)	0.39	0.05	0.01	0.98	0.96-0.99	0.39	0.06	0.01	0.98	0.97-0.99	0.01	0.90	0.82-0.95	0.89	0.79-0.95	0.08
Kinetics																
Peak knee flexion moment (Nm)	122.37	33.60	8.26	0.95	0.92-0.97	128.08	40.57	12.31	0.95	0.91-0.97	9.96	0.85	0.74-0.92	0.90	0.80-0.95	0.15
Peak knee abduction moment (Nm)	21.44	9.82	4.21	0.87	0.78-0.93	23.64	15.12	7.06	0.88	0.80-0.93	5.63	0.69	0.50-0.83	0.72	0.49-0.86	0.17
Peak knee internal rotation moment (Nm)	12.50	4.41	1.51	0.90	0.83-0.94	12.51	4.82	1.95	0.87	0.78-0.93	2.59	0.40	0.11-0.63	0.56	0.25-0.76	<0.01
Vertical GRF (N)	930	153	53	0.91	0.85-0.95	935	157	54	0.91	0.85-0.95	56	0.69	0.48-0.82	0.73	0.50-0.86	0.03

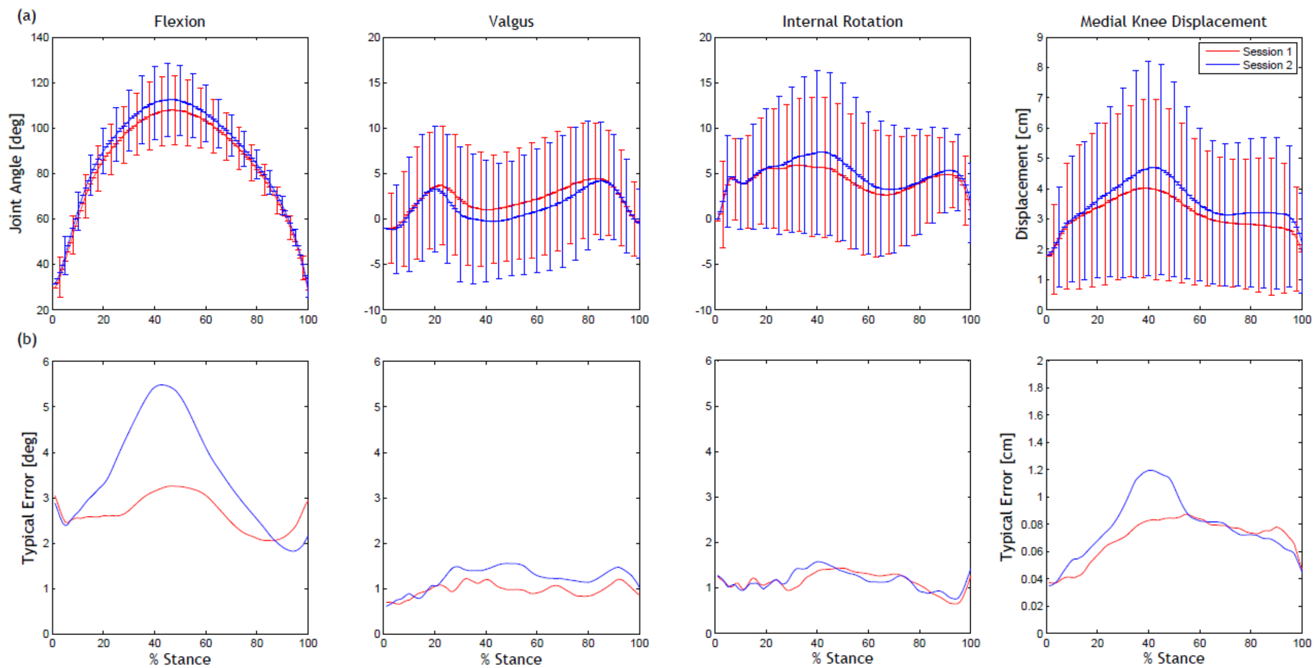

Figure 1. The kinematics mean curves and the standard deviation of the subjects from the session 1 and 2 in the part (a) of the figure. The mean typical error of the subjects in the part (b) of the figure. Red: Session 1; Blue: Session 2.

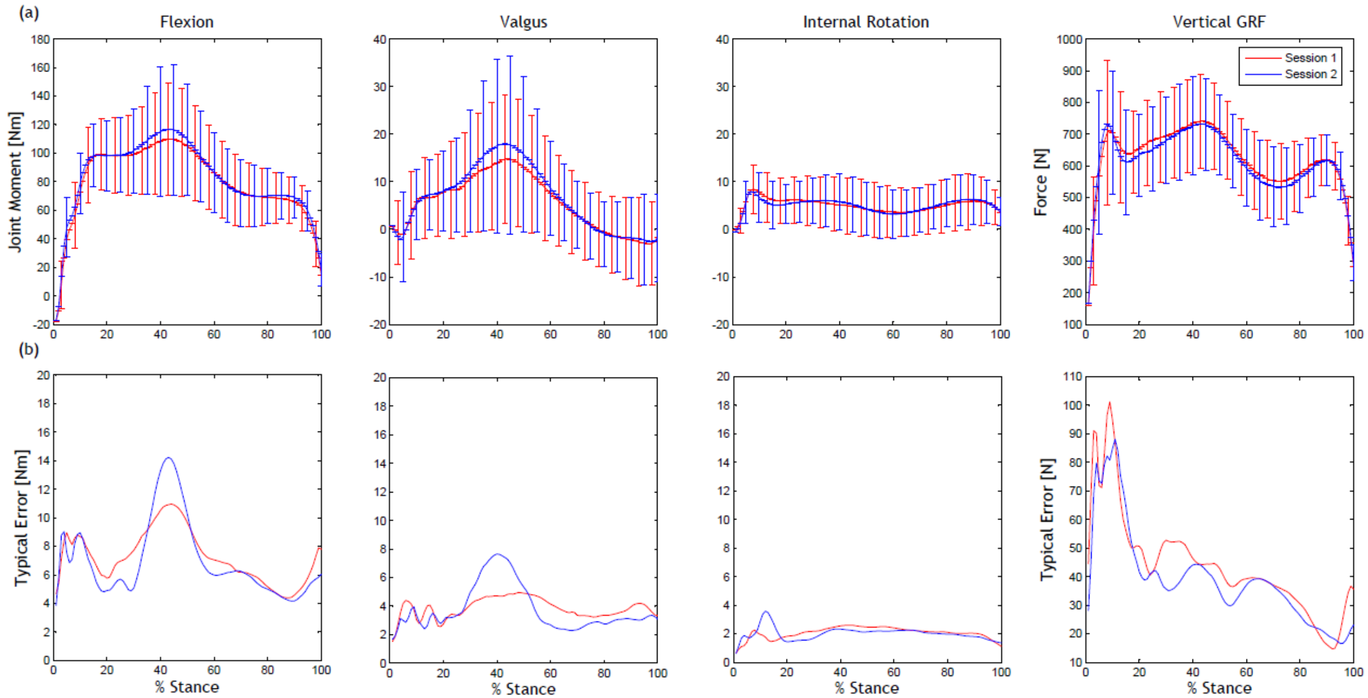

Figure 2. The kinetics and vertical GRF mean curves and the standard deviation of the subjects from the session 1 and 2 in the part (a) of the figure. The mean typical error of the subjects in the part (b) of the figure. Red: Session 1; Blue: Session 2.

Paper II

Reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks

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3D motion analysis, kinematics and kinetics, elite female athletes, handball, football, screening task.

Abstract (250 words)

The purpose of this study was to assess the within- and between-session reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks performed by elite female handball and football athletes. Moreover, we aimed at determining the minimum number of sidestep cutting trials necessary to obtain a reliable measure. Nineteen elite female handball and 22 elite female football (soccer) athletes (mean ± SD: 22 ± 4 yrs old, 168 ± 5 cm, 66 ± 8 kg) were tested in two sessions, on average separated by two weeks. The reliability of two sport-specific sidestep cutting tasks was quantified by intra-class correlations (ICC), typical error and Spearman's rank correlation. ICC values were computed based on trials 1-3, 1-4 and 1-5. Only minor improvements in ICC values were seen when increasing the number of trials from three to five. Based on trials 1-3, all variables showed excellent within-session reliability (mean ICC: 0.91, 95% CI: 0.89-0.93), and good to excellent between-session reliability (mean ICC: 0.73, 95% CI: 0.70-0.76). All between-session rank correlation coefficients demonstrated moderately positive correlation (mean: 0.72, 95% CI: 0.69-0.76). A few frontal plane biomechanical variables displayed lower between-session reliability in the football task compared with the handball task. All the discrete biomechanical variables achieved good to excellent within- and between-session reliability. The between-session ranking showed moderate to strong positive correlations, implying that the measurements could reliably reproduce the ranking of individuals in multiple-session studies. Adequate reliability could be attained from three trials, with only minor improvements when adding more trials.

1. Introduction

Sidestep cutting tasks have been studied extensively in ACL injury related research [1-5]. Marker-based three-dimensional kinematics and kinetics have been used to investigate, for instance, the influence of gender on knee biomechanics [5] or the effect of cutting technique on knee abduction loading [6]. In some protocols, the task has been made more sport-specific by involving a static defender [4,7], passing/receiving a ball [8] or performing unanticipated cuts [3].

Surprisingly, the reliability of lower limb biomechanical measurements during sidestep cutting has not been adequately evaluated. Recently, Sankey et al. [9] investigated the reliability of knee loading variables during sidestep cutting. However, the generalizability of results is limited because the sample size was restricted to four males and four females. Sigward & Power [5] reported the between-session waveform reliability to be acceptable, but only five female football (soccer) athletes were included. Ford and co-workers [3] reported the within-session intra-class correlation (ICC) coefficients for knee and ankle frontal plane kinematics to be excellent in 126 adolescent basketball athletes. However, the between-session reliability was not reported. Lastly, Kaila et al. [10] reported the reliability of lower limb biomechanics in 15 male football athletes using Pearson's correlation coefficients. However, Pearson's correlation is insensitive to systematic differences, and is therefore inappropriate for reliability measurements [11].

A sport-specific sidestep cutting task is more complex and varied than a simple change-of-direction cut, because the sport-specific setting such as ball-passing and static defender could induce movement variations. Therefore, it may result in a lower reliability. It is necessary to determine the minimum number of trials necessary to obtain a reliable measure for such as a task.

Thus, the purpose of this study was to assess the within-session and between-session reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks among elite female handball and football athletes. Moreover, we aimed at determining the minimum number of sidestep cutting trials necessary to obtain a reliable measure.

2. Methods

2.1 Subjects. Nineteen elite female handball and 22 elite female football athletes (mean ± SD: 22 ± 4 yrs old, 168 ± 5 cm, 66 ± 8 kg) performed a sport-specific sidestep cutting task in our biomechanics laboratory. The study was approved by the Regional Ethics Committee and all subjects provided signed informed consent forms.

2.2 Design and protocol. The athletes were tested in two separate sessions, on average separated by two weeks. The athletes wore indoor shoes, shorts and a sports bra. Thirty-seven reflective markers were attached over anatomical landmarks on the legs, arms and torso [1]. All marker positions were defined uniquely, also those not defined by anatomical landmarks. One experienced physiotherapist, with several years practice for marker placement, was employed for skin marker placement in both sessions.

The handball athletes performed a handball-specific faking maneuver involving a static human defender [12], whereas the football athletes performed a sidestep cutting with a football through pass.

For the handball-specific protocol, the athlete used an approach run of close to 6 m, allowing match-like approach speed. The athlete received a lateral pass from a teammate before executing a match-like faking maneuver to pass a 170 cm tall static defender (Figure 1a). The defender adjusted her position during practice trials to ensure that the athlete stepped onto a force platform with her stance foot.

For the football-specific protocol, the athlete also used an approach run of close to 6 m. A teammate passed a football in a direction that forced the athlete to perform a sharp sidestep cutting maneuver in order to catch up (Figure 1b).

For both tasks, athletes were allowed to have at least three practice cuts to familiarize themselves with the situation, and at least five successful trials from each side (left-right and right-left) were completed. Two test operators ensured that these trials were performed with match-like intensity with the stance foot on the force platform and all markers firmly attached to the athlete's skin.

We used a 480 Hz 16-camera system (Oqus 4, Qualisys, Gothenburg, Sweden) to capture the motion, while we recorded ground reaction forces using a force platform (AMTI LG6-4-1, Watertown, Massachusetts, USA) collecting data at 960 Hz. We calibrated the motion analysis system according to guidelines from the manufacturer, and tracked marker trajectories using the Qualisys Track Manager (Version 2.8, Qualisys, Gothenburg, Sweden). We defined the contact phase as the period where the unfiltered vertical ground reaction force exceeded 20 N. Marker trajectories and force data were filtered and interpolated using Woltring's smoothing spline in the cubic mode with 15 Hz low-pass cut-off [1]. We calculated the hip joint center using the method proposed by Bell et al. [13], with the anterior-posterior position of the hip joint decided by the anterior-posterior position of the marker over the greater trochanter. Furthermore, we defined the knee joint center according to Davis [14], and the ankle joint center according to Eng and Winter [15]. Anatomical coordinate systems of the thigh and shank were determined from the static calibration trials. We defined the vertical axis in the direction from the distal to the proximal joint center, while the anteroposterior axis was defined perpendicular to the vertical axis with no mediolateral component. The third axis was the cross product of the vertical and anteroposterior axes. Consequently, all segments had neutral internal/external rotation in the static calibration trial.

We obtained technical, dynamic thigh and shank segment coordinate systems using an optimization procedure involving singular value decomposition [16].

We used the Grood and Suntay [17] convention for calculating joint kinematics from the marker-based motion analysis. Furthermore, we calculated medial knee position as the perpendicular distance between the knee joint center and the line joining the ankle and hip joint centers projected onto the plane, which is defined by the knee joint flexion axis and the line joining the ankle and hip joint centers. The difference between the position at the initial foot contact and the peak value was defined as the medial knee displacement for analysis.

We estimated inertia parameters based on 46 measures of segment heights, perimeters and widths using a modified Yeadon's method [18], with hand and foot parameters calculated with the method of Zatsiorsky and Seluyanov [19]. We calculated joint kinetics with inverse dynamics using recursive Newton-Euler equations of motion as described by Davis et al. [14] and projected onto the three rotational axes of the joint according to the joint coordinate system standard [17]. The horizontal ground reaction force was projected on the shank coordinate system, resulted as the anterior shear force and medial shear force components. We ran all calculations using custom Matlab scripts (MathWorks Inc., Natick, Massachusetts, USA).

2.3 Statistical analysis. Thirty-three discrete variables were extracted from the joint kinematics, joint kinetics and forces waveform for the reliability analysis. We defined the dominant leg as the preferred leg when kicking a ball. We used paired t-tests and Pearson's correlations to evaluate the symmetry between dominant and non-dominant leg [20]. We report the mean, standard deviation, and typical error of the within- and between-session measurements. The typical error was calculated from the standard deviation of inter-trial differences divided by the square root of two [21]. We computed ICC values for within sessions (ICC(3,k)) and between sessions (ICC(3,1)) reliability measures. The ICC classification of Fleiss [22] (less than 0.4, poor; between 0.4 and 0.75, fair to good; and greater than 0.75, excellent) was used to interpret the ICC values. The ICC values were computed based on trials 1-3, 1-4 and 1-5, separately. We used a Z-test to test the significance of difference between the ICC values of handball and football athletes. The critical value for the Z-score was set at 1.96.

To assess the consistency of subject ranking between-sessions, Spearman's rank correlation coefficients and between-session ICC were calculated based on the mean measurements from sessions 1 and 2. The classifications of Zou et al. [23] (greater than 0.5, moderately positive; greater than 0.8, strongly positive) were used to interpret the rank correlation coefficients. Paired t-tests were employed to assess the mean difference between sessions. Cohen's *d* was computed to assess the effect size of the mean differences (less than 0.2, none; between 0.2 and 0.5, small to medium; 0.5 and 0.8, medium to large; and greater than 0.8, very large) [24]. Statistical significance was set at $p \leq 0.05$. Statistical analyses were performed using SPSS 18 (SPSS Inc., Chicago, IL, USA) and the statistics toolbox of Matlab (MathWorks Inc., Natick, Massachusetts, USA). Moreover, the mean curves were time-normalized from 0% to 100% of the stance phase to represent the motion.

3. Results

The time courses of mean joint kinematics (Figure 2a), joint kinetics (Figure 2b) and ground reaction forces (Figure 2c) were consistent between sessions. However, we could observe a noticeable difference in the waveforms between the handball- and football-specific sidestep cutting tasks.

The mean ICC values for dominant and non-dominant sides were calculated with respect to trials 1-3, 1-4 and 1-5 (Table 1). Only small changes were seen increasing the number of trials from three to five. For the dominant side, the mean between-session ICC value improved from 0.73 (good) to 0.75 (excellent), and the mean within-session ICC values improved from 0.91 (excellent) to 0.95 (excellent). For the non-dominant side, the mean between-session ICC value improved from 0.75 (excellent) to 0.78 (excellent), and the mean within-session ICC values improved from 0.91 (excellent) to 0.94 (excellent). Since there were only slight improvements in reliability from increasing the number of trials, we have reported reliability measures based on trials 1-3 in the following.

All 33 variables showed a statistically significant correlation between sides (mean: 0.59, 95% CI: 0.53-0.65). Significant differences between limbs were only observed for four of the variables (paired t-test; peak hip abduction angle, peak knee internal rotation angle, peak knee valgus moment and peak knee flexion moment). This implies that the majority of the variables were dependent and symmetrical between dominant and non-dominant side in the sport-specific sidestep cutting task. For simplicity, we will therefore report reliability measures based on the dominant leg only.

There were no significant differences in the within-session ICC values of handball and football athletes. For the between-session ICC values, two variables on the dominant side and five variables on the non-dominant side showed a significant difference (Table 2). Since the vast majority of the ICC values showed no significant difference between handball and football, we have pooled the results of handball and football athletes (Table 3). Six variables showed a significant difference between sessions. All the Cohen's d values were less than 0.21, which implies a small effect size. All the variables showed excellent within-session reliability (mean ICC: 0.91, 95% CI: 0.89-0.93), and good to excellent between-session reliability (mean ICC: 0.73, 95%

CI: 0.70-0.76). In addition, all the between-session rank correlation coefficients demonstrated moderate to strong positive correlation (mean: 0.72, 95% CI: 0.69-0.76).

4. Discussion

All the discrete biomechanical variables achieved good to excellent reliability in both within- and between-session analyses. The between-session rank correlations were moderate to strong, implying that the measurements could reliably reproduce the ranking of individuals in multiple-session studies. We furthermore found adequate reliability to be attained from three trials. Handball- and football-specific sidestep cutting tasks showed good to excellent reliability level in most of the variables, except the between-session reliability of a few frontal plane biomechanical variables in the football-specific sidestep cutting task.

Importantly, we found that increasing the number of trials from three to five trials only slightly improved the reliability of the measurements, even if these tasks can be considered technically demanding. Thus, adequate reliability can be attained using only three trials. However, it should be noted that we required the athlete to have at least three practice trials before the official trials. In some cases, the athlete needed up to five practice trials to become familiarized with the task.

The between-session ICCs were, as expected, lower than the within-session reliability. Similar findings have been reported in 3D motion analysis studies of gait [20,25], running [20,26], and vertical drop jumps [20,27]. Although several explanations exist, such as the differences in movement execution, this phenomenon can likely be explained predominantly by differences in marker placement between sessions [25,28]. Nevertheless, the between-session ICC values attain good reliability level for all discrete variables (Table 3). Moreover, the moderate to strong between-session rank correlations implies that all the discrete variables can provide a consistent ranking of athletes, which is essential for reliable screening of athletes. Coupled with the considerably small between-session typical error and effect size, the two sport-specific sidestep cutting tasks can generally provide reliable within- and between-session biomechanical measurements.

Handball- and football-specific sidestep cutting tasks were found to have comparable reliability in most of the variables. A few variables showed lower between-session reliability for the football athletes, especially on the non-dominant side (Table 2). These were mainly frontal plane biomechanical variables of the hip and knee. A possible reason for this difference is that the handball faking maneuver is a fundamental motion and performed repeatedly in every training session, which in turn may enhance movement consistency. McLean et al. [4] reported that a static defender could provoke the athlete to change direction more rapidly during cutting, and thereby affect frontal plane biomechanics of the lower limb. In line with the findings of McLean et al. [4], we found that the protocol including a static defender induced larger medial ground reaction force, larger hip abduction angle and knee valgus angle (Figure 2a and 2c). The static defender could thus potentially limit the possibility for movement variability, and thereby enhance the reliability.

The sport-specific sidestep cutting tasks investigated in the current study showed similar reliability characteristics to the other motion tasks, except the between-session peak knee flexion angle and peak knee internal rotation angle. When comparing our results with gait [29], running [26] and vertical drop jumps [27], the between-session peak flexion angle is more reliable in gait (ICC: 0.96) and running (ICC: 0.93), than vertical drop jumping (ICC: 0.62) and sidestep cutting (ICC: 0.63). In contrast to sidestep cutting and vertical drop jumps, gait and running are daily life motions which have been executed and developed from early childhood [30]. However, sidestep cutting maneuver (ICC: 0.82) has substantially higher between-session knee internal rotation reliability compared to gait (ICC: 0.54), and similar reliability compared to running (ICC: 0.83) and vertical drop jumping (ICC: 0.87).

From an injury-risk screening perspective, the question is if the same individuals are identified as having outer-range scores (i.e. being at risk) with repeated tests. If using peak knee valgus angle (Spearman's rank correlation: 0.73) as the variable of interest, 10 out of 12 athletes that were classified as being at risk (identified among the top 30%) in session one were also classified among the top 30% in session two. However, if using peak knee abduction moment (Spearman's rank correlation: 0.59), only half of the high-risk athletes (six out of 12 athletes) in session one were also included in the top 30% in session two. Nevertheless, the accuracy of lower limb biomechanics for identifying the risk of ACL injury needs to be established from a prospective cohort study using injuries as the outcome.

5. Conclusion

All the discrete biomechanical variables examined achieved good to excellent within- and between-session reliability in both within- and between-session analyses. In addition, the between-session ranking of subjects was moderately to strongly correlated, implying that the measurements can reliably reproduce the ranking of individuals if tested repeatedly. Adequate reliability could be attained from three trials only, with only minor increase in reliability when adding more trials.

6. References

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Table 1. The average ICC values (interpretation from Fleiss, 1986) of all 33 variables on dominant and non-dominant side from different sum of trials

	Side	Trial 1-3	Trial 1-4	Trial 1-5
Within-session	Dominant	0.91 (Excellent)	0.94 (Excellent)	0.95 (Excellent)
	Non-dominant	0.91 (Excellent)	0.93 (Excellent)	0.94 (Excellent)
Between-session	Dominant	0.73 (Good)	0.74 (Good)	0.75 (Excellent)
	Non-dominant	0.75 (Excellent)	0.77 (Excellent)	0.78 (Excellent)

Table 2. The list of variables having significant difference on the between-session ICC values between handball and football athletes, significant level set at Z-score > 1.96

	Between-session ICC		
Variables	Football	Handball	Z-score
Dominant side			
Peak hip adduction moment	0.37	0.90	2.71
Peak knee valgus angle	0.22	0.72	2.12
Non-dominant side			
Peak medial shear force	0.59	0.93	2.30
Peak ankle eversion moment	0.31	0.85	2.55
Peak knee abduction moment	0.51	0.95	2.57
Peak ankle plantarflexion angle	0.88	0.52	2.07
Peak hip adduction angle	0.32	0.91	3.05

Table 3. Within- and between-session mean, typical error and reliability measurements for dominant side from trials 1-3. SD: standard deviation; IC: initial contact; GRF: ground reaction force

	Within-session					Between-session					
	Mean	SD	Typical Error	ICC(3,k)	95% CI	Typical Error	ICC(3,1)	95% CI	Spearman's rho	95% CI	Cohen's d
Kinematics											
Hip internal rotation angle at IC (°)	-1.7	8.4	2.1	0.94	0.90-0.97	3.3	0.64	0.42-0.79	0.64	0.37-0.82	0.11
Peak hip internal rotation angle (°)	10.9	7.7	1.8	0.95	0.92-0.97	2.6	0.73	0.55-0.85	0.74	0.52-0.87	0.07
Hip abduction angle at IC (°)	18.6	7.5	1.7	0.96	0.93-0.97	2.5	0.76	0.59-0.86	0.84	0.68-0.92	0.14
Peak hip adduction angle (°)	-11.2	7.1	2.4	0.92	0.87-0.96	2.7	0.64	0.42-0.79	0.72	0.49-0.86	0.13
Peak hip abduction angle (°)	27.9	6.9	1.8	0.94	0.89-0.96	2.5	0.66	0.44-0.80	0.69	0.44-0.84	0.18*
Hip flexion angle at IC (°)	43.1	11.9	1.5	0.99	0.98-0.99	4.5	0.72	0.53-0.84	0.83	0.68-0.92	0.01
Peak hip flexion angle (°)	51.3	13.6	2.9	0.96	0.94-0.98	4.4	0.74	0.56-0.85	0.78	0.58-0.89	0.06
Knee internal rotation angle at IC (°)	4.0	8.5	2.9	0.93	0.88-0.96	3.1	0.70	0.50-0.83	0.70	0.45-0.85	0.02
Peak knee internal rotation angle (°)	12.2	5.8	1.1	0.97	0.95-0.98	1.6	0.82	0.69-0.90	0.85	0.70-0.93	0.05
Knee valgus angle at IC (°)	5.5	3.5	1.2	0.92	0.87-0.96	1.7	0.55	0.29-0.73	0.55	0.24-0.76	0.06
Peak knee valgus angle (°)	11.8	5.3	1.4	0.95	0.92-0.97	2.5	0.64	0.42-0.79	0.73	0.50-0.86	0.06
Knee flexion angle at IC (°)	21.8	7.5	2.7	0.90	0.83-0.94	2.2	0.68	0.48-0.82	0.60	0.31-0.79	0.06
Peak knee flexion angle (°)	60.5	6.9	4.1	0.75	0.56-0.85	2.6	0.63	0.41-0.79	0.65	0.38-0.82	0.09
Ankle internal rotation angle at IC (°)	-5.2	5.4	1.7	0.93	0.88-0.96	1.3	0.83	0.70-0.90	0.83	0.68-0.92	0.04
Peak ankle internal rotation angle (°)	-2.4	5.9	2.5	0.88	0.79-0.93	1.6	0.83	0.70-0.90	0.84	0.69-0.92	0.05
Ankle inversion angle at IC (°)	4.8	6.6	2.0	0.94	0.90-0.97	2.7	0.61	0.37-0.77	0.60	0.31-0.79	0.09
Peak ankle inversion angle (°)	28.4	5.9	1.4	0.95	0.92-0.97	2.1	0.71	0.49-0.84	0.74	0.51-0.87	0.21*
Ankle plantarflexion at IC (°)	8.0	12.1	5.6	0.90	0.83-0.94	4.0	0.77	0.60-0.87	0.77	0.57-0.89	0.15*
Peak ankle plantarflexion angle (°)	26.3	9.3	2.3	0.95	0.91-0.97	2.7	0.80	0.66-0.89	0.70	0.46-0.85	0.01
Peak ankle dorsiflexion angle (°)	23.2	6.8	1.8	0.94	0.89-0.96	2.5	0.69	0.48-0.82	0.54	0.22-0.75	0.09
Medial knee displacement (cm)	3.7	2.0	0.6	0.93	0.88-0.96	0.7	0.65	0.44-0.80	0.59	0.28-0.78	0.11
Kinetics											
Peak hip internal rotation moment (Nm)	43.8	14.6	5.3	0.89	0.82-0.94	5.7	0.68	0.47-0.81	0.69	0.43-0.84	0.04
Peak hip adduction moment (Nm)	50.4	82.5	15.8	0.97	0.96-0.98	24.0	0.85	0.74-0.92	0.81	0.64-0.91	0.08
Peak hip flexion moment (Nm)	243.3	95.3	19.6	0.96	0.93-0.98	35.8	0.77	0.60-0.87	0.83	0.68-0.92	0.05
Peak knee external rotation moment (Nm)	27.4	13.5	3.3	0.95	0.91-0.97	4.6	0.74	0.56-0.85	0.63	0.36-0.81	0.06
Peak knee abduction moment (Nm)	120.3	39.5	13.0	0.90	0.82-0.94	13.6	0.72	0.52-0.84	0.59	0.29-0.78	0.17*
Peak knee flexion moment (Nm)	167.3	45.2	12.3	0.94	0.89-0.96	13.0	0.80	0.66-0.89	0.74	0.52-0.87	0.00
Peak ankle internal rotation moment (Nm)	63.9	18.2	6.1	0.91	0.85-0.95	6.7	0.71	0.51-0.84	0.72	0.49-0.86	0.19*
Peak ankle eversion moment (Nm)	36.5	15.4	4.6	0.93	0.88-0.96	4.4	0.76	0.59-0.86	0.70	0.45-0.84	0.05
Peak ankle dorsiflexion moment (Nm)	164.3	30.3	6.6	0.96	0.94-0.98	7.5	0.86	0.75-0.92	0.82	0.65-0.91	0.10
Ground Reaction Forces											
Peak vertical GRF (N)	1543.2	441.8	93.3	0.97	0.94-0.98	88.7	0.92	0.85-0.96	0.89	0.78-0.95	0.10
Peak posterior shear force (N)	439.1	159.2	47.2	0.94	0.90-0.97	53.1	0.77	0.60-0.87	0.76	0.54-0.88	0.15*
Peak medial shear force (N)	738.5	154.3	42.7	0.94	0.90-0.96	37.5	0.85	0.73-0.92	0.81	0.63-0.90	0.10

* $p<0.05$, significant difference between the mean measurements of session 1 and 2.

Figure 1. The testing situation of (a) the handball-specific sidestep cutting task and (b) the football-specific sidestep cutting task

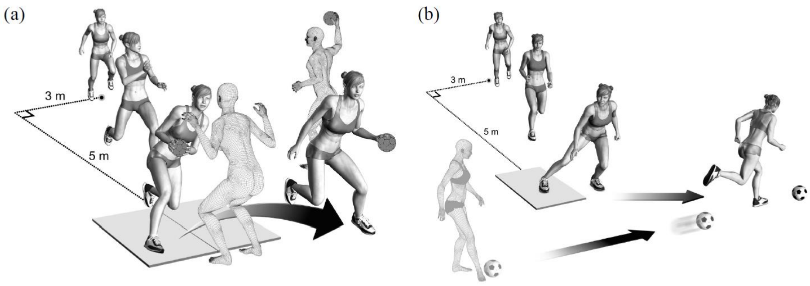
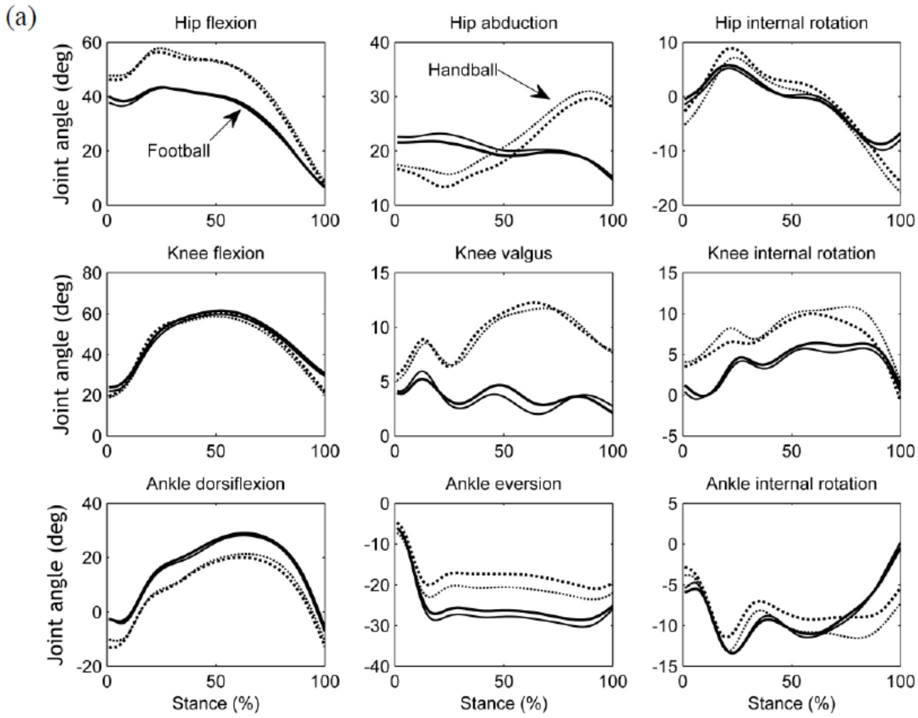
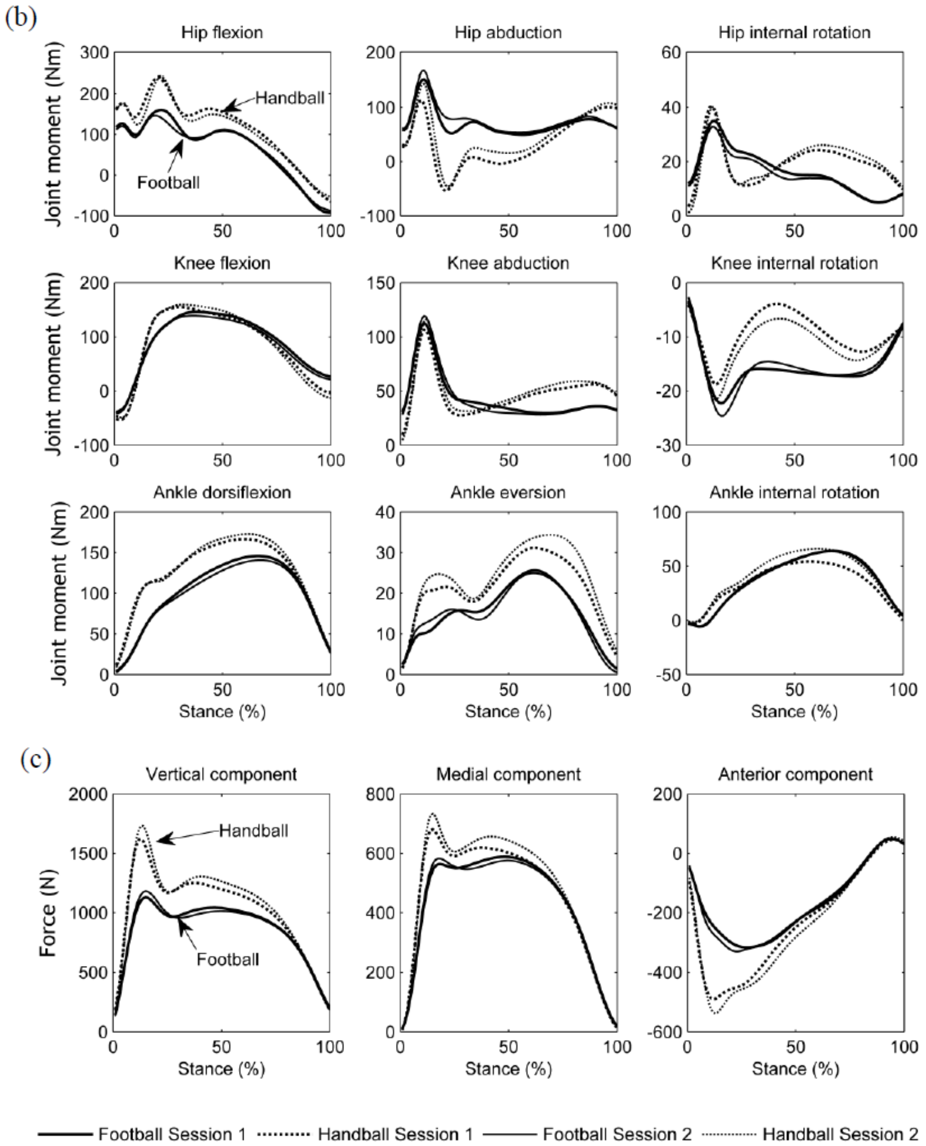


Figure 2. The time courses of mean measurements of handball and football athletes from session 1 and 2, for (a) joint kinematics, (b) joint kinetics and (c) ground reaction forces




(b)
Hip flexion
Hip abduction
Hip internal rotation
Joint moment (Nm)
Handball
Football
Knee flexion
Knee abduction
Knee internal rotation
Ankle dorsiflexion
Ankle eversion
Ankle internal rotation
Stance (%)
(c)
Vertical component
Medial component
Anterior component
Force (N)
Football Session 1
Handball Session 1
Football Session 2
Handball Session 2

Paper III

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TECHNICAL NOTE

The Effect of Thigh Marker Placement on Knee Valgus Angles in Vertical Drop Jumps and Sidestep Cutting

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Knee valgus angles measured in sidestep cutting and vertical drop jumps are key variables in research on anterior cruciate ligament (ACL) injury causation. These variables are also used to quantify knee neuromuscular control and ACL injury risk. The aims of the current study were to (1) quantify the differences in the calculated knee valgus angles between 6 different thigh marker clusters, (2) investigate the trial ranking based on their knee valgus angles, and (3) investigate the influence of marker clusters on the cross-talk effect. Elite female handball and football players (n = 41) performed sidestep cutting and vertical drop jumping motions. We found systematic differences up to almost 15° of peak valgus between the marker sets in the drop jump test. The Spearman's rank correlation coefficient varied from .505 to .974 among the 6 marker sets. In addition, the cross-talk effect varied considerably between the marker clusters. The results of the current study indicate that the choice of thigh marker cluster can have a substantial impact on the magnitude of knee valgus angle, as well as the trial ranking. A standardized thigh marker cluster, including nonanatomical landmark, is needed to minimize the variation of the measurement.

Keywords: motion analysis, knee biomechanics, kinematics, marker cluster

Measurements of knee valgus angles in sidestep cutting and vertical drop jumping are commonly used to quantify knee neuromuscular control[1–4] and predict anterior cruciate ligament (ACL) injury risk.[5] Knee kinematic measures, typically calculated via a marker-based motion analysis system, use a minimum of 3 markers per segment to generate a coordinate system and predict movement of underlying bone. The coordinate systems and calculations of joint angles are usually standardized;[6,7] however, the marker cluster configurations differ between studies.[8,9]

In gait analysis, substantial differences in knee valgus measurements have been reported between 5 marker cluster protocols, such as plug-in-gait and total 3D gait.[10] The average difference in knee valgus angles for the entire contact phase was found to be 9.7°, while the maximum difference could be as high as 25° between 2 protocols.[10] This study, as well as many others, suggests that the differences originate predominantly from the location of the thigh markers, with the largest errors attributed to markers placed at the proximal thigh.[9–15] Therefore, certain thigh marker cluster configurations, such as the plug-in-gait,[16] include the hip joint center (HJC) as a nonsurface virtual marker. The HJC is normally estimated based on pelvic markers,[17] which are less prone to soft tissue artifacts. Therefore, the HJC may be a better proximal anatomical marker than the greater trochanter, where a soft tissue artifact (STA) of close to 30 mm has been reported.[18]

Furthermore, if the calculated knee flexion–extension axis is not aligned with the true knee flexion–extension axis, so-called cross-talk error in the calculated knee valgus–varus angle will be induced as a function of knee flexion–extension.[19] This phenomenon has been shown in both model simulation and in-vivo sport motion.[19–22] Cross-talk can be minimized by correcting the femoral coordinate system with reference to the functional knee flexion–extension axis, which is believed to be closest to the true knee flexion–extension.[19,23] However, although methods to reduce crosstalk exist, the calculation involves the knee flexion–extension axis of the thigh marker cluster.[6] No previous study has investigated cross-talk using a constant knee flexion–extension axis definition while changing the thigh marker cluster.

Unfortunately, there is limited information on differences between various marker clusters in high-impact movements such as sidestep cuts and vertical drop jumps. The aims of the current study were to (1) quantify the differences in the calculated knee valgus angles between 6 different thigh marker clusters, (2) investigate the trial ranking based on their peak knee valgus angles, and (3) investigate the influence of marker clusters on the cross-talk effect.

Methods

Nineteen elite female handball players and 22 elite female football (soccer) athletes (21.6 ± 4.0 y, 168 ± 5 cm, 66 ± 8 kg, BMI 23.3 ± 1.8, mean ± SD) performed sidestep cutting and vertical drop jumps. The subjects were recruited from the top Norwegian handball and football divisions for the current methodological study. The regional ethics committee approved the study and all subjects signed the informed consent form.

Thigh markers were put on the following landmarks: lateral femoral epicondyle, greater trochanter, lateral thigh, anterior thigh, and distal anterior thigh (Figure 1 and Table 1). We defined 6 different thigh marker clusters for the current study (Table 2). Marker cluster I, II, and VI are based on previous studies that investigated knee biomechanics in sidestep cutting and vertical drop jumping.[1–5] Marker cluster III has no marker on proximal thigh, where we will typically find high STA.[11,14,24] Marker cluster IV and V are composed of markers on the distal thigh and only 1 marker on the

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proximal thigh. In marker cluster V and VI, we included the HJC as a virtual marker. The HJC was determined by the regression method of Bell et al.[17] Since all 6 marker clusters were attached to the subject, the current thigh marker set-up allowed all 6 marker clusters to be calculated simultaneously. The rest of the full-body marker set-up followed the standard protocol of the Oslo Sports Trauma Research Center, using 35 retro-reflective markers.[3,25–29] An experienced physiotherapist was responsible for marker placement.

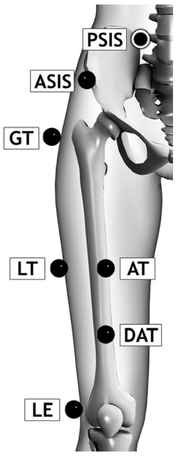

Figure 1 — Illustration of skin markers on thigh segment. PSIS = posterior superior iliac spine; ASIS = anterior superior iliac spine; GT = greater trochanter; LT = lateral thigh; AT = anterior thigh; DAT = distal anterior thigh; LE = lateral femoral epicondyle.

Sixteen 480 Hz infrared cameras (Oqus, Qualisys, Gothenburg, Sweden) recorded the movement of the retro-reflective skin markers. We recorded ground reaction forces using 2 force plates (AMTI, Watertown, MA) sampling at 960 Hz. A standing static calibration was performed to define the anatomical coordinate systems of each segment. The midpoint of the condyles defined the knee joint center and the ankle joint center.[30,31] The longitudinal axis of the shank and thigh anatomical coordinate system was pointing from the distal to the proximal joint center, while the anteroposterior axis was defined perpendicular to the vertical axis with no mediolateral component. The third axis was the cross product of the vertical and anteroposterior axis.

The subjects completed 5 valid sidestep cuts and then 3 valid vertical drop jumps.[3] They were told to perform their regular sidestep cut, attempting to fake a static defender into going one way while cutting to the other side. In addition, the subjects were instructed to drop off a 30-cm box and perform a maximal jump after landing. The contact phase was defined by having unfiltered vertical ground reaction force exceeding 20 N. Marker trajectories were low-pass filtered using a smoothing spline in the cubic mode with a 15 Hz cut-off frequency.[32] We obtained technical thigh and shank segment coordinate systems from an optimization procedure involving singular value decomposition.[33] Then we employed the joint coordinate system convention to resolve the three-dimensional knee joint angles including flexion–extension, valgus–varus, and internal–external rotation.[6] Finally, we calculated knee joint angles with the 6 different thigh marker clusters using otherwise identical methods. The calculations were done using custom Matlab scripts (MathWorks Inc., Natick, MA).

Statistical analysis was performed using SPSS 18 (IBM, Inc., Chicago, IL). Peak knee valgus during the contact phase and knee

Table 1 Definition of skin markers on the thigh segment

Skin Markers	Descriptions
Lateral femoral epicondyle	Lateral epicondyle on femur; it was palpated in the standing position with the subject flexing and extending the knee
Greater trochanter	The most superior point of the greater trochanter in standing position
Lateral thigh	2 cm anterior of the middle point of greater trochanter and lateral femoral epicondyle in standing position
Anterior thigh	The most anterior point of the thigh on the same height of lateral thigh in standing position
Distal anterior thigh	The most anterior point of the thigh at the level of the midpoint between lateral thigh and lateral femoral epicondyle in standing position

Table 2 Thigh skin marker cluster description

Marker Cluster	Skin Marker Component	Number of Markers	Design
I	LE, AT, GT	3	Ford et al[1]; Hewett et al[5]; Imwalle et al[2]
II	LE, DAT, GT	3	McLean et al[4]
III	LE, DAT, LT	3	Distal thigh
IV	LE, DAT, LT, GT	4	Distal thigh + greater trochanter
V	LE, DAT, LT, HJC	4	Distal thigh + hip joint center
VI	LE, DAT, LT, HJC, GT	5	Kristianslund et al[25]

Abbreviations: LE = lateral femoral epicondyle; AT = anterior thigh; GT = greater trochanter; DAT = distal anterior thigh; LT = lateral thigh; HJC = hip joint center.

valgus at initial contact were compared across the 6 different thigh skin marker clusters using one-way repeated-measures ANOVA. Each trial (N = 123 for vertical drop jumps; N = 205 for sidestep cuts) was considered as a single data point and used to construct the distribution for the statistical analyses. Post hoc *t* tests with Bonferroni correction for repeated tests were then conducted to investigate which marker clusters differed by pairwise comparison (30 pairs in total for 6 marker clusters and 2 outcome measurements). Spearman's rank correlation coefficient was calculated to assess the consistency of trial ranking across conditions. Statistical significance was set at $P \leq .05$. As an indirect measure of the cross-talk, the correlation between the knee flexion–extension and varus–valgus angle profiles was used.[23]

Results

For the magnitude of the measurement, substantial differences were observed between the different marker clusters (Figure 2). For sidestep cutting, the valgus angle at initial contact ranged from 0.7° ± 3.3° (mean ± SD) to 6.2° ± 3.4°, and peak valgus angle ranged from 9.0° ± 4.0° to 13.7° ± 5.4° with the different thigh marker clusters ($P \leq .05$) (Table 3). Twenty-five out of 30 marker cluster comparisons were significantly different ($P \leq .05$) (Table 4). For vertical drop jumping, the valgus angle at initial contact ranged from –5.0° ± 3.5° to 2.1° ± 3.0°, and peak valgus angle ranged from 1.9° ± 2.8° to 15.8° ± 5.8° with the different thigh marker clusters (Table 3). Twenty-three out of 30 marker cluster comparisons were significantly different ($P \leq .05$) (Table 4). The duration of the contact phase was 0.32 s ± 0.07 s for sidestep cutting and 0.61 s ± 0.11 s for vertical drop jumping.

The consistency of the trial rankings between marker cluster pairs was assessed. Rank correlation coefficients larger than .8 were defined as strongly positive,[34,35] thereby consistent. All rank correlations were significant ($P \leq .05$). For the ranking of trials based on their knee valgus angles, in both sidestep cutting and vertical drop jumping, the rankings were consistent between marker clusters II, III, and IV, as well as between marker clusters V and VI (Table 5). Marker cluster I was distinctly different to all the other marker clusters in terms of both the magnitude and the trial ranking by knee valgus measurements. For the indirect cross-talk measure, the correlation between the knee flexion–extension and varus–valgus angles varied from .28 to .64 in sidestep cutting and from .27 to .72 in vertical drop jumping (Table 3).

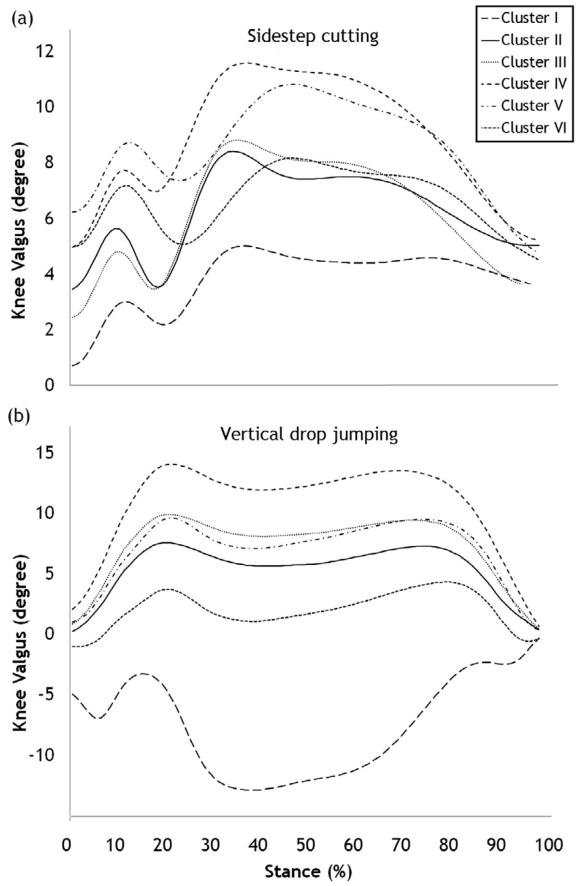

Figure 2 — Mean profile of knee valgus angles during stance phase in (a) sidestep cutting and (b) vertical drop jumping.

Discussion

This is the first study to quantify the differences in knee valgus angles between different thigh marker clusters in sidestep cutting and vertical drop jumping. The results of the current study indicate that the choice of thigh marker cluster can have a substantial impact on the magnitude of knee valgus angle, as well as the trial ranking. A standardized thigh marker cluster, including clear guidelines for placing nonanatomical markers, is needed to minimize the variation of the measurement, and to enable direct comparisons of motion data between different studies.

For the vertical drop jumping task, there was a discrepancy in the calculated magnitude of peak knee valgus angles between the 6 marker clusters up to 13.9°. For certain pairs of marker clusters, such as marker cluster I and III, the rank correlation coefficient was as low as .505 in the peak valgus angle in vertical drop jumping. Hence, different trials are identified as representing poor knee control. We observed large differences between marker clusters in the middle stance phase of the vertical drop jump (Figure 2b). The average difference between cluster I and IV at this point was as large as 25°. However, we observed smaller differences at initial contact when the knee was more extended. Furthermore, the knee valgus measurement at initial contact was more consistent (Table 5 and Figure 2b). The valgus angle at initial contact would be the most reliable point at which to compare frontal-plane knee angles across different marker sets.

Marker cluster I was substantially different from the other marker clusters in both sidestep cutting and vertical drop jumping. Surprisingly, the only difference between marker clusters was the nonanatomical marker on the anterior thigh, which was approximately 5 cm apart. The observed difference in valgus angles can possibly be explained by previous studies reporting that markers along the proximal–distal direction on anterior thigh showed different magnitude of mediolateral STA,[18,36] in particular for a jump landing task.[37] Akbarshahi et al showed that the distal anterior thigh

Table 3 The mean (SD) of the outcome variables in the 6 different thigh marker clusters and the correlation between the knee flexion–extension and varus–valgus angle profiles

	Marker Cluster						
Outcome Variables	**I**	**II**	**III**	**IV**	**V**	**VI**	**Peak Flexion**
Sidestep cutting							62.0 (8.4)
Valgus at IC (°)	0.7 (3.3)	3.5 (2.6)	2.4 (3.0)	5.0 (3.1)	6.2 (3.4)	5.0 (3.4)	
Peak valgus (°)	9.0 (4.0)	10.5 (4.1)	10.6 (4.5)	13.5 (4.6)	13.7 (5.4)	11.4 (5.5)	
Correlation	0.31	0.38	0.57	0.64	0.43	0.28	
Vertical drop jumping							107.4 (15.2)
Valgus at IC (°)	–5.0 (3.5)	0.24 (3.0)	0.9 (3.2)	2.1 (3.0)	1.0 (3.5)	–1.0 (3.6)	
Peak valgus (°)	1.9 (2.8)	9.5 (5.0)	11.6 (5.4)	15.8 (5.8)	12.0 (5.3)	6.9 (5.3)	
Correlation	0.63	0.53	0.59	0.72	0.51	0.27	

Abbreviation: IC = initial contact.

Table 4 *P*-value for the post hoc *t* test with Bonferroni correction for different combinations of the 6 thigh skin marker clusters in sidestep cutting and vertical drop jumping

	II		III		IV		V		VI	
Marker Cluster	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**
Sidestep cutting										
I	< .01	.038	< .01	.013	< .01	< .01	< .01	< .01	< .01	< .01
II			.032	1.00*	< .01	< .01	< .01	< .01	< .01	.387*
III					< .01	< .01	< .01	< .01	< .01	.853*
IV							.002	1.00*	1.00*	.001
V									.002	< .01
Vertical drop jumping										
I	< .01	< .01	< .01	< .01	< .01	< .01	< .01	< .01	< .01	< .01
II			1.00*	.021	.001	< .01	1.00*	.002	.068*	.001
III					.110*	< .01	1.00*	1.00*	< .01	< .01
IV							.320*	< .01	< .01	< .01
V									< .01	< .01

Note. IC = the valgus angle at initial contact; peak = peak valgus angle.

*No significant difference on the outcome variable of 2 respective marker clusters in post hoc *t* tests with Bonferroni correction ($P \le .05$).

marker has less mediolateral STA than the anterior thigh marker in hip axial rotation and step-up motion.[36] Some methods will use asymmetrical placement of the midthigh markers, such as plug-in-gait,[16] to optimize the automatic marker tracking process. In such cases, bilateral comparison on lower limb kinematics should be carefully interpreted.

The cross-talk effect is the erroneous coupling between knee varus–valgus and flexion–extension because of the misalignment of knee joint flexion–extension axis.[23] In our study, the flexion–extension axis definition was equal for all marker clusters so we would assume the same degree of cross-talk for all the clusters. However, the correlation coefficient varied from .28 to .64 in sidestep cutting and from .27 to .72 in vertical drop jumping (Table 3). This implies that the cross-talk effect is not only sensitive to the definition of knee joint flexion–extension axis, but also the marker clusters. For instance, lateral movement of a thigh marker away from the midline of the body will result in an external rotation of the thigh coordinate system along the longitudinal axis, which will again result in increasing measured knee valgus angles when the knee is flexed. Hence, the placement of all markers is crucial, not only those placed on bony landmarks.

Marker clusters V and VI, which included the HJC, gave consistent magnitude and trial ranking results with each other, but were less consistent with the other marker clusters. The inclusion of the HJC may introduce a systematic difference to the thigh technical coordinate system. In this study, the HJC was estimated based on a regression model using anatomical landmarks of the pelvis.[17] The average error in this HJC regression model is about 15 mm.[38] It is

Table 5 Spearman's rank correlation coefficients for the 6 different combinations of thigh skin marker clusters in sidestep cutting and vertical drop jumping

	II		III		IV		V		VI	
Marker Cluster	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**	**IC**	**Peak**
Sidestep cutting										
I	.735	.661	.631	.625	.630	.585	.744	.732	.817*	.753
II			.930*	.974*	.944*	.930*	.711	.671	.678	.589
III					.918*	.957*	.681	.683	.613	.574
IV							.756	.710	.666	.588
V									.965*	.966*
Vertical drop jumping										
I	.781	.543	.762	.505	.691	.419	.684	.699	.726	.778
II			.970*	.989*	.960*	.943*	.778	.783	.739	.673
III					.947*	.966*	.787	.782	.739	.656
IV							.810*	.738	.735	.586
V									.972*	.947*

Note. All the Spearman's rank correlation coefficients achieved significant levels ($P \leq .05$). IC = the valgus angle at initial contact; peak = peak valgus angle.

*Coefficient > 0.8 corresponds to strongly positive correlated and consistent.

likely to be a better proximal anatomical marker than the greater trochanter, where a STA close to 30 mm has been reported.[18] The accuracy of HJC estimation may even be improved by using the functional method for the estimation of HJC.[20,39]

A true gold standard was not present in this study. Therefore a robust recommendation on which marker cluster to choose cannot be made. Biplanar videoradiography techniques have previously been used to provide a noninvasive gold standard in measuring lower extremity limb motion during jump landings, but due to the limited recording volume it is currently not possible to measure sidestep cutting, nor the whole landing phase of vertical drop jumps.[40–42] Myers et al[40] reported peak knee valgus in the first 0.1 seconds after initial contact for females during jump landings of 1.6° ± 0.9° using biplanar videoradiography techniques. Using the same kinematics calculation approach,[6] the peak knee valgus angle was observably smaller in value and standard deviation compared with the current study (Table 3), as well as previous studies using skin marker-based motion analysis, which have reported knee valgus angles close to 30°.[1,43]

In conclusion, the choice of thigh marker clusters affected the magnitude of the knee valgus angle in drop jumps and sidestep cutting, as well as the trial ranking. A standardized thigh marker cluster, including clear guidelines for placing nonanatomical markers, is needed to minimize the variation of the measurement, and to enable direct comparisons of motion data between different studies.

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Paper IV

The effect of overhead target on the lower limb biomechanics during a vertical drop jump test in elite female athletes

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Abstract

The purpose of the study was to investigate the effect of an overhead target on the jump height and lower limb biomechanics in all three planes of motion in a vertical drop jump (VDJ) task among elite female handball and football (soccer) players. The hypothesis was that adding an overhead target to the VDJ task improves jump height, increases joint loading and decreases frontal plane knee control. Five hundred twenty-three female handball and football players (mean ± SD: 21 ± 4 yrs, 168 ± 6 cm, 65 ± 8 kg) completed the test. The overhead target increased jumping height by 5.8%. Furthermore, the overhead target lead to statistically significant changes in many of the lower limb biomechanical variables examined. However, all the changes in kinematics and kinetics were clinically insignificant, as indicated by the small effect sizes. Strong to moderate positive Spearman's rank correlations were found between the two conditions. Therefore, an overhead target is unlikely to increase the range of responses in biomechanical variables in elite female handball and football athletes.

Introduction

Anterior cruciate ligament (ACL) injuries often occur in jump-landing situations (Boden et al., 2000; Krosshaug et al., 2007; Koga et al., 2010). The aim of the vertical drop jump (VDJ) test is to assess lower extremity neuromuscular control in a jump-landing situation (Hewett et al., 2005; Smith et al., 2012; Nilstad et al., 2014). Setting up the VDJ test is simple, whether in the field; using visual scoring, or in the laboratory; using more advanced motion analysis systems (Padua et al., 2009; Myer et al., 2011; Kristianslund & Krosshaug, 2013; Nilstad et al., 2014). In a small prospective study, Hewett and coworkers (2005) reported a significant association between frontal plane knee motion during a VDJ and the risk for ACL injuries. The VDJ test has therefore been advocated as a screening task in sports with a high risk of ACL injury (Padua et al., 2009; Myer et al., 2011; Smith et al., 2012), although the predictive ability of the test has yet to be replicated by others (Goetschius et al., 2012).

To mimic a jump-landing situation with match-like joint loading, several external stimuli have been tested. Verbal encouragement can enhance performance in exercise testing (Mcnair et al., 1996; Campenella et al., 2000). Introducing a target as an external focus could also improve exercise performance (Porter et al., 2010). Moreover, an external focus of attention could serve to distract the athlete and increase the automaticity in movement control (Wulf et al., 2001; Wulf & Dufek, 2007). Therefore, an overhead target can potentially increase joint loading and distract the athlete from focusing on frontal plane knee control when tested. An overhead target may therefore increase the range of response in biomechanical variables and thereby increase the sensitivity of the VDJ task to assess ACL injury risk.

Other than ACL injury risk, other common sports injuries, such as ankle instability (Delahunt et al., 2006) and patellofemoral pain (Boling et al., 2009), have been suggested to be associated with an inappropriate jump-landing technique. Moreover, the effect of an overhead target on jumping height and lower limb posture is of interest for therapist (Butler et al., 2003). Therefore, the current study reported the differences for a range of lower limb biomechanical variables.

Previous studies have confirmed that an overhead target increases jump height (Ford et al., 2005; Wulf & Dufek, 2007; Wulf & Dufek, 2009). Nonetheless, its effect on lower extremity biomechanics has not been investigated adequately. Wulf & Dufek (2009) concluded the overhead target increases lower extremity joint loading, but only in a limited sample of four male and six female university students. Ford and co-workers (2005) found significant increases in knee flexion angle and joint extensor moment in a sample of 18 collegiate female athletes. However, the effect of an overhead target on frontal and transverse plane biomechanics has never been investigated. Moreover, the correlation of biomechanical measures between the two conditions is not known.

The objective of the current study was therefore to investigate the effect of an overhead target on jump height and lower limb biomechanics in all three planes of motion in a VDJ task in a sample of 523 elite female handball and football players. We hypothesized that adding an overhead target to the VDJ task would improve jump height, increase joint loading and decrease frontal plane knee control.

Methods

Participants. The current study was part of a prospective cohort study aimed at investigating risk factors for noncontact ACL injuries in female elite handball and football players. Three hundreds and sixty-three elite female football (soccer) and 160 elite female handball athletes (mean ± SD: 21 ± 4 yrs old, 168 ± 6 cm, 65 ± 8 kg) completed a VDJ task in target and non-target condition in our biomechanics laboratory between 2009 and 2014. The Regional Committee for Medical and Health Research Ethics South East approved the study and all athletes provided signed informed consent forms.

Data collection. The athletes wore indoor shoes, shorts and a sports bra. The same, experienced physiotherapist attached 37 reflective markers over anatomical landmarks on the legs, arms and torso (Kristianslund et al., 2012). All marker positions were uniquely defined, also those not defined by anatomical landmarks (Mok et al., 2015).

The athletes first performed the VDJ task in a non-target condition, and subsequently the VDJ task in a target condition. In the non-target condition, we instructed athletes to drop off a 30 cm stand and perform a maximal jump upon landing, with their feet on separate force platforms (AMTI LG6-4-1, Watertown, Massachusetts, USA). They were allowed to have three practice trials and data from at least three valid trials were collected for each athlete. At least two test operators observed the execution of the jump and monitored 30 seconds to 45 seconds rest between trials. If a sub-maximal effort was suspected, or when jumping instead of dropping off the stand (i.e. increasing the vertical center of mass position at take-off from the stand), we asked the athlete to repeat the jump. Athletes received verbal encouragement to give a maximal jump effort.

In the target condition, we set up a horizontal bar as an overhead target, about 30 cm in front of the athlete (Figure 1). The height of the bar was set based on jump height in the VDJ task in the non-target condition. We asked the athletes to perform a VDJ and reach the bar by the head. If the athlete managed to reach the bar, it was raised in increments of five cm or less. The task ended when the athlete failed to reach the same height twice and at least three valid trials had been collected. The final three valid trials of each athlete were used for the analysis.

We used a 480 Hz 16-camera system (Oqus 4, Qualisys, Gothenburg, Sweden) to capture motion, while we recorded ground reaction forces using two force platforms collecting at 960 Hz (AMTI LG6-4-1, Watertown, Massachusetts, USA). We calibrated the motion analysis system according to guidelines from the manufacturer, and calculated and tracked marker trajectories using the Qualisys Track Manager (Qualisys, Gothenburg, Sweden).

Data processing. We defined the contact phase as the period where the unfiltered vertical ground reaction force exceeded 20 N. Marker trajectories and force data were filtered and interpolated using Woltring's smoothing spline in the cubic mode (Woltring, 1986), using a 15 Hz cut-off (Kristianslund et al., 2012). We calculated the hip joint center using the method proposed by Bell et al. (1990), with the anterior-posterior position of the hip joint decided by the anterior-posterior position of the marker over the greater trochanter. Furthermore, we defined the knee joint center according to Davis (1991), and the ankle joint center according to Eng & Winter (1995). Anatomical coordinate systems of each segment were determined from the static calibration trials. We defined the vertical axis in the direction from the distal to the proximal joint center, while the anterio-posterior axis was defined perpendicular to the vertical axis with no mediolateral component. The third axis was the cross product of the vertical and antero-posterior axes. Consequently, all segments had neutral internal/external rotation in the static calibration trial. We obtained technical and dynamic segment coordinate systems using an optimization procedure involving singular value decomposition (Soderkvist & Wedin, 1993).

We estimated inertia parameters based on 46 measures of segment heights, perimeters and widths using a modified Yeadon’s method (Yeadon, 1990), with hand and foot parameters calculated with the method of Zatsiorsky & Seluyanov (1983). We calculated joint moments with inverse dynamics using recursive Newton-Euler equations of motion as described by Davis et al. (1991) and projected onto the three rotational axes of the joint according to the joint coordinate system standard (Grood & Suntay, 1983; Kristianslund et al., 2014).

We used the Grood & Suntay (1983) convention for calculating joint angles from the marker-based motion analysis. We calculated medial knee position as the perpendicular distance between the knee joint center and the line joining the ankle and hip joint centers, projected on the frontal plane. The difference between the medial knee position at the initial foot contact and the peak value was defined as the medial knee displacement. An advantage of this convention compared with a pure knee separation measure is that we can assess knee control individually for the left and right leg. We ran all calculations using custom Matlab scripts (MathWorks Inc., Natick, Massachusetts, USA).

Statistical analysis. Jump height was defined by the difference of the vertical center of mass position between the static anatomical position and the maximal height position during the jump. Thirty selected biomechanical variables were extracted from the joint kinematics, joint kinetics and force time course for the analyses. We extracted variables for both legs and organized them into dominant and non-dominant side. The mean of three trials for each athlete was used for the analysis of each variable. A two-way ANOVA was used to determine the main effects and interaction of side dominance and condition on each variable. If a significant interaction was found, post-hoc paired t-tests were done separately for the dominant and non-dominant side. We report the mean with standard deviation for each variable from the two conditions. Moreover, we computed effect size as the mean difference divided by the pooled standard deviation to assess the clinical significance of the mean difference between conditions (Cohen, 1992). We interpreted effect size as follows: <0.2, no effect; 0.2-0.5, small effect; 0.5-0.8, medium effect; >0.8, large effect (Cohen, 1992). To assess the consistency of athlete ranking between two tasks, we calculated Spearman’s rank correlation coefficients based on the measurements from the non-target and target condition and the rank correlation coefficients >0.8 as strongly positive and 0.5-0.8 as moderately positive (Zou et al., 2003). Statistical significance was set at $p<0.05$. Statistical analyses were performed using SPSS 18 (SPSS Inc., Chicago, IL, USA).

Results

Jump height increased by 5.8% when athletes reached for an overhead target, corresponding to a medium effect size (Table 1). Of all variables studied (Table 1), a significant interaction between side dominance and task was only observed for medial knee displacement ($p=0.02$). We therefore analyzed the dominant and non-dominant side results separately for this variable, whereas data for the dominant and non-dominant side were combined for the remaining variables. The addition of an overhead target resulted in a significant change in the majority of the biomechanical variables examined; however, the effect sizes were small (Table 1). We observed a strong rank correlation between the two tasks in 23 out of the 32 variables, and a moderate correlation in the remaining nine (Table 1). The medial knee displacement variable had the lowest rank correlation.

Discussion

The main finding of this study was that introducing an overhead target increased jump performance by 5.8% and led to statistically significant changes in many lower limb biomechanical variables. Our results are generally consistent with previous studies, including the increase in jump height and joint loading when introducing an overhead target (Ford et al., 2005; Wulf & Dufek, 2009). However, the changes in kinematics and kinetics are likely to be clinically insignificant, as indicated by the small effect sizes. The explanation for this apparent discrepancy is the unprecedented sample size of 523 players.

Comparing the two conditions, we recorded a 5.0% increase in peak vertical ground reaction force (Table 1), which corresponds well to the increase in jump height. This demonstrates that an external motivation enhance the effort even if we provided strong verbal encouragement for the non-target task and also asked the athletes to repeat the trial if a sub-maximal effort was suspected.

According to the definition of frontal plane knee control used (Nilstad et al., 2014), the target condition seemed to decrease frontal plane knee control as reflected by the increase in the peak knee valgus angle from 5.1° to 5.6°, and dominant side medial knee displacement from 2.2 to 2.4 cm (Table 1). The change in frontal plane knee control is of particular interest as such variables have been suggested to be associated with the risk of ACL injury in a mixed-sport cohort of female athletes (Hewett et al., 2005). The target condition resulted in changes in biomechanical variables of particular interest in ACL injury risk. However, the small effect sizes suggest that the kinematic differences have no clinical relevance.

The overhead target generated a stiffer landing, which may in turn increase the load on the ACL. We recorded a 4.4° decrease in peak knee flexion angle, a 7.0° decrease in the range of knee flexion and a 65 N increase in peak vertical ground reaction force (Table 1). The results are consistent with previous studies assessing differences between soft and stiff landings (Devita & Skelly, 1992; Pollard et al., 2010; Myers et al., 2011). It is known that higher vertical ground reaction force and extended knee position in stiff landings will increase ACL strain due to the increased anterior tibial translation (Markolf et al., 1995; Myers et al., 2011). In contrast, soft landings will allow the vertical ground reaction force to be dissipated over a larger range of knee flexion than stiff landings (Devita & Skelly, 1992; Pollard et al., 2010; Myers et al., 2011), and are therefore advocated as a preventive strategy for reducing ACL injuries in successful injury prevention programs (Boden et al., 2000; Myer et al., 2004; Mandelbaum et al., 2005; Myklebust et al., 2013; Taylor et al., 2015). However, again the effect sizes were small, suggesting that these differences are unlikely to be clinically relevant.

The rank correlations observed for the various measures of lower limb biomechanics between the two conditions are generally interpreted as strong to moderate. Even so, when applying these tasks as screening tests for ACL injury risk, they must be interpreted with caution. Hewett et al. (2005) have suggested that a high knee abduction (valgus) moment during a drop jump task without an overhead target is a strong predictor for ACL injury risk. The rank correlation for knee abduction moment in our study was 0.78, which is generally considered as moderate to strong. However, if we consider the 20% of players (n=105) with the highest magnitude of peak knee abduction moment in the non-target condition as being at risk, only 67% of these were also included in the top quintile in the target condition. Therefore, the association between lower limb biomechanics and the risk of ACL injury needs to be examined separately for both tasks in prospective studies.

There are some limitations that should be borne in mind when interpreting the results of the current study. Marker placement and soft tissue artifacts are well-known sources of error in a skin marker-based motion analysis (Leardini et al., 2005; Miranda et al., 2013). However, since marker placement was standardized, these errors are expected to be similar for the two tasks and unlikely to have affected our findings. The fact that we did not randomize the task order means that fatigue may have been induced in the target condition, which was always performed last. However, considering that they were able to jump 2.4 cm higher in the target condition, fatigue effects were likely small. Another potential limitation is that reaching a horizontal bar by the head may pose less of a challenge for neuromuscular control than e.g. grasping an overhead ball. Lastly, it is still unknown how the other demographic factors, such as gender and age, could have affected the results. Therefore, the generalizability of these results to players of a different skill level, sport or even injury history is unknown.

Perspectives

Adding an overhead target to the VDJ test improves jump height, but generates only minor changes in lower limb biomechanics. In addition, athlete ranking based on the lower limb biomechanics were consistent between two conditions. Therefore, an overhead target is unlikely to increase the range of responses in biomechanical variables in elite female handball and football athletes.

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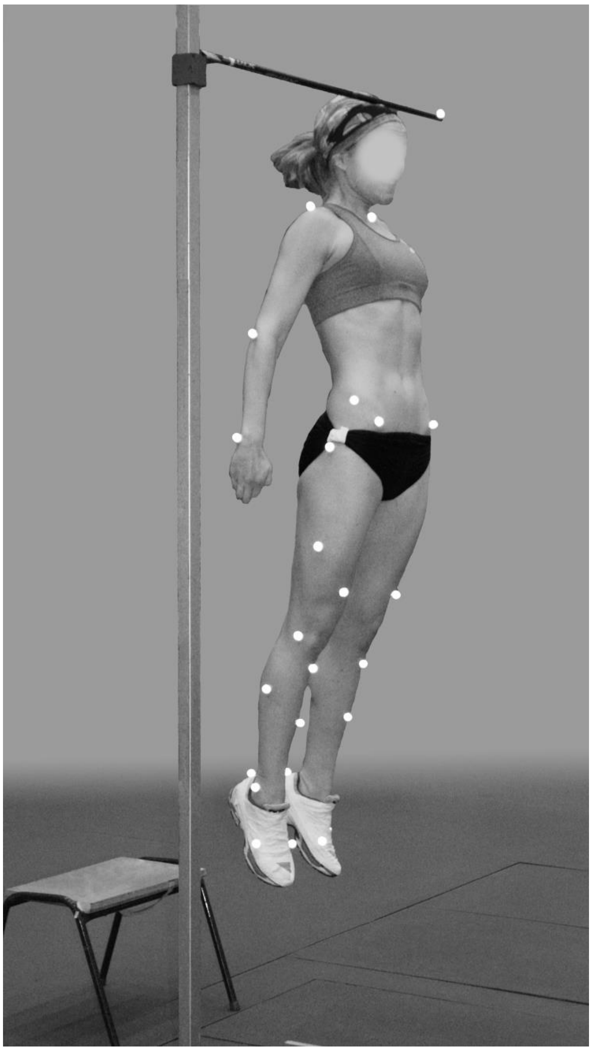
Table 1. Comparison of biomechanical variables during vertical drop jumps in target and non-target conditions (n=523). Data are shown as means, standard deviations (SD), mean differences (percent change in parentheses) with 95% confidence interval (CI), effect sizes (Cohen's d) and Spearman's rank correlation coefficients. Positive values denote an increased score with the addition of an overhead target; statistically significant differences are shown in bold.

	Non-target condition	Target condition	Comparison		
	Mean±SD	Mean±SD	Mean difference (95% CI)	Effect size	Spearman's rank correlation coefficient
Jump height (cm)	41.6±4.8	44.0±5.0	**2.4 (2.2 to 2.7)**	0.51	0.84
Kinematics					
Hip flexion angle at IC (°)	48.3±8.5	49.9±8.4	**1.6 (1.3 to 2.0)**	0.20	0.79
Peak hip flexion angle (°)	83.0±15.6	78.2±15.5	**-4.8 (-5.4 to -4.4)**	0.31	0.85
Hip abduction angle at IC (°)	5.3±3.6	5.0±3.8	-0.3 (-0.4 to -0.1)	0.07	0.86
Peak hip abduction angle (°)	7.6±3.9	7.2±3.9	**-0.4 (-0.6 to -0.4)**	0.12	0.86
Hip internal rotation angle at IC (°)	-0.6±5.3	-0.5±5.4	0.1 (-0.1 to 0.2)	0.01	0.90
Peak hip internal rotation angle (°)	4.6±5.6	4.4±5.6	-0.2 (-0.3 to 0.0)	0.03	0.90
Knee flexion angle at IC (°)	32.3±7.9	34.9±8.1	**2.6 (2.3 to 2.9)**	0.33	0.81
Peak knee flexion angle (°)	93.5±14.8	89.1±13.5	**-4.4 (-5.0 to -3.9)**	0.31	0.80
Range of knee flexion (°)	61.2±16.8	54.2±15.5	**-7.0(-8.4 to -5.6)**	0.43	0.79
Knee valgus angle at IC (°)	-2.0±4.2	-1.8±4.4	0.2 (0.18 to 0.4)	0.07	0.92
Peak knee valgus angle (°)	5.1±5.2	5.6±5.2	**0.5 (0.4 to 0.6)**	0.10	0.95
Knee internal rotation angle at IC (°)	-0.5±5.4	0.3±5.7	**0.8 (0.6 to 1.0)**	0.15	0.83
Peak knee internal rotation angle (°)	10.4±6.1	9.9±5.9	-0.5 (-0.6 to -0.3)	0.07	0.94
Ankle plantarflexion at IC (°)	10.9±9.7	11.3±9.1	0.4 (0.1 to 0.8)	0.04	0.77
Peak ankle plantarflexion angle (°)	25.1±7.3	25.1±7.1	0.02 (-0.2 to 0.2)	<0.01	0.89
Ankle inversion angle at IC (°)	9.3±5.2	8.5±5.2	**-0.8 (-1.0 to -0.6)**	0.16	0.83
Peak ankle inversion angle (°)	10.8±4.6	10.2±4.7	**-0.6 (-0.7 to -0.4)**	0.11	0.85
Ankle internal rotation angle at IC (°)	0.5±6.6	-0.1±6.7	**-0.6 (-0.7 to -0.5)**	0.09	0.87
Peak ankle internal rotation angle (°)	6.1±6.1	5.6±6.2	-0.5 (-0.6 to -0.3)	0.08	0.80
Medial knee displacement on the dominant side (cm) [a]	2.2±1.3	2.4±1.6	**0.2 (0.1 to 0.3)**	0.15	0.58
Medial knee displacement on the non-dominant side (cm) [a]	2.1±1.4	2.0±1.6	-0.08 (-0.1 to 0.0)	0.06	0.63
Kinetics					
Peak vertical GRF (N)	1294.6±391.6	1359.5±400.7	**64.9 (48.9 to 80.7)**	0.16	0.77
Peak hip flexion moment (Nm)	203.3±46.3	208.2±52.4	**4.9 (2.9 to 6.8)**	0.10	0.80
Peak hip abduction moment (Nm)	40.0±17.1	41.4±17.8	1.4 (0.8 to 2.1)	0.08	0.77
Peak hip internal rotation moment (Nm)	20.1±9.5	20.8±10.1	0.7 (0.3 to 1.0)	0.07	0.81
Peak knee flexion moment (Nm)	141.3±37.3	150.1±40.3	**8.8 (7.5 to 10.1)**	0.23	0.87
Peak knee abduction moment (Nm)	20.5±10.7	21.7±12.1	**1.2 (0.7 to 1.6)**	0.10	0.78
Peak knee internal rotation moment (Nm)	10.2±6.9	11.1±7.7	**0.9 (0.6 to 1.1)**	0.12	0.83
Peak ankle dorsiflexion moment (Nm)	112.7±31.0	120.4±4.8	**7.7 (6.5 to 8.8)**	0.23	0.85
Peak ankle inversion moment (Nm)	1.8±4.0	2.0±4.6	0.2 (0.0 to 0.3)	0.04	0.78
Peak ankle internal rotation moment (Nm)	16.5±8.7	17.5±9.7	**1.0 (0.7 to 1.3)**	0.11	0.85

[a] Separated by side because of the significant interaction of side dominance and condition.

Figure 1. The target condition of vertical drop jump test



Appendix 1

Approval from the Regional Committee for Medical Research Ethics

UNIVERSITETET I OSLO
DET MEDISINSKE FAKULTET

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

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

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

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

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

Komiteen tar svar på merknader til etterretning.

Komiteen har ingen merknader til revidert informasjonsskriv med samtykkeerklæring.

Komiteen tilrår at prosjektet gjennomføres.

Vi ønsker lykke til med prosjektet.

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

Jørgen Hardang
Jørgen Hardang
Sekretær

UNIVERSITETET I OSLO
DET MEDISINSKE FAKULTET

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

Regional komité for medisinsk og helsefaglig forskningsetikk Sør-Øst A (REK Sør-Øst A)
Postboks 1130 Blindern
NO-0318 Oslo

Telefon: 22 84 46 66
Telefaks: 22 85 05 90
E-post: jorgen.hardang@medisin.uio.no

Dato: 15.12.08
Deres ref.:
Vår ref.: S-07078a

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

Vi viser til skjema for protokolltillegg og endringer datert 3.12.08 vedlagt revidert informasjonsskriv.

Prosjektleder ønsker å utvide prosjektpopulasjonen til kvinnelige elitefotballspillere fra Toppserien i Norge (ca 240 spillere).

Komiteen godkjenner endringen slik den er beskrevet i skjema for protokolltillegg og endringer og videresender kopi av informasjonsskriv, endringsskjema samt komiteens vedtak til Helsedirektoratet for behandling av endring av biobanken.

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

Jørgen Hardang
Jørgen Hardang
Sekretær

Kopi: Helsedirektoratet, Postboks 7000, St. Olavs plass, 0130 Oslo

Appendix 2

Approval from the Norwegian Social Science Data Services

Norsk samfunnsvitenskapelig datatjeneste AS
NORWEGIAN SOCIAL SCIENCE DATA SERVICES

NSD

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

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

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

TILRÅDING AV BEHANDLING AV PERSONOPPLYSNINGER

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

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

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

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

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

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

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

Vennlig hilsen

Bjørn Henrichsen Katrine Utaaker Segadal

Kontaktperson: Katrine Utaaker Segadal tlf: 55 58 35 42

Vedlegg: Prosjektvurdering

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

Norsk samfunnsvitenskapelig datatjeneste AS
NORWEGIAN SOCIAL SCIENCE DATA SERVICES

NSD

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

Tron Krosshaug
Senter for idrettsskadeforskning
Norges idrettshøgskole
Pb 4014 Ullevål Stadion
0806 OSLO

Dato: 16.02 2009	Vår ref: 16639 PB/LR	Deres dato:	Deres ref:

ENDRING AV FORSKNINGSPROSJEKT

Vi viser til endringsmelding mottatt 28.12.2008, samt påfølgende e-postkorrespondanse med daglig ansvarlig (senest 13.02.2009), gjeldende prosjektet

16639 Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndball- og elitefotballspillere – en prospektiv kohortstudie

I endringsskjema opplyses det om at man ønsker å utvide prosjektpopulasjonen til å også omfatte kvinnelige elitefotballspillere fra toppserien i Norge (ca. 240 individer). Tittelen på prosjektet endres dermed fra *Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndballspillere – en prospektiv kohortstudie* til *Risikofaktorer for fremre korsbåndskader hos kvinnelige elitehåndball- og elitefotballspillere – en prospektiv kohortstudie.*

For hele utvalget ønsker man videre å se på genetiske faktorer som risikofaktorer for fremre korsbåndskader. Man skal ta blodprøve (5 ml. venøs prøve) av deltagerne for å studere genvarianter som kan bidra til å lage et svakere ligament. Kollagen er en viktig substans i ligamenter, og man vil i første omgang se på gener som er ansvarlige for kvaliteten mht. kollagenfibre. Prøven sendes til avidentifisert (med kobling til navneliste som oppbevares ved NIH) til Ullevål Universitetssykehus for å ekstrahere DNA. Ekstrahert DNA vil bli sendt til samarbeidspartner i Sør-Afrika, Exercise Science and Sports Medicine Research Unit (ESSM) for videre analyse. Det vil på grunnlag av analysene gjøres sammenligninger mellom skadede og ikke skadede spillere. Resultatene av testene vil kun være tilgjengelig for dette forskningsformålet. Biobanken opprettes ved Ullevål Universitetssykehus.

En ytterligere endring av prosjektet består i at ombudet etter avtale med daglig ansvarlig Tron Krosshaug, registrerer prosjektet som forskerprosjekt i stedet for som studentprosjekt. Studenten ved NIH Eirik Kristianslund er fortsatt å regne som medarbeider i prosjektet, men registreringsendringen foretas på bakgrunn av at prosjektets tidsperspektiv (planlagt avslutning i 2017) gjør det lite hensiktsmessig å la studenten bli stående som kontaktperson for ombudet. Videre registreres stipendiaten ved NIH Agnethe Nilstad som medarbeider i prosjektet sammen med Dr. Scient. Kathrin Steffen og Dr. Med. Thor Einar Andersen.

Ombudet mottok 13.02.2009 reviderte informasjonsskriv for rekruttering av deltagere til prosjektet og finner begge skrivene meget tilfredsstillende.

Ombudet legger til grunn at endringen, inkludert opprettelsen av forskningsbiobank, godkjennes av REK. Det bes om at kopi av tilråding ettersendes.

Ombudet anbefaler at det opprettes en databehandleravtale med Ullevål og med ESSM, jf. personopplysningsloven § 15.

Endringene medfører ingen endring av ombudets opprinnelige vurdering og tilråding av prosjektet (se brev datert 03.05.2007) mht. behandlings- eller hjemmelsgrunnlag.

Ombudet minner om at bruk av videoopptak i undervisnings- eller formidlingsøyemed kan medføre meldeplikt overfor Datatilsynet. Dette bør avklares direkte med tilsynet.

Ta gjerne kontakt dersom noe er uklart.

Vennlig hilsen

Bjørn Henrichsen

Pernilla Bollman

Kontaktperson: Pernilla Bollman 55 58 24 10

Appendix 3

Informed consent

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

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

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

Gjennomføring av undersøkelsen
Vi ønsker at du som elitespiller deltar i denne studien, og deltakelsen er frivillig. Testingen vil finne sted på Norges idrettshøgskole. I løpet av en dag vil vi gjennomføre ulike styrke-, balanse- og bevegelighetstester, anatomiske målinger, samt gjennomføre en bevegelsesanalyse av hvordan du finter, vender, hopper og lander. Undersøkelsen starter med en kort oppvarming, deretter får du festet små refleksmarkører på kroppen (35 stk totalt). Du vil så bli bedt om å gjennomføre tre finter/vendinger og tre fallhopp. Under disse øvelsene vil det være 8 infrarøde kamera som filmer markørene, samtidig som kreftene fra underlaget blir målt. Dataene fra markører, kraftplattform og anatomiske mål benyttes i en matematisk modell som gir ut leddkrefter og momenter. Disse kreftene/momentene gir oss informasjon om hvordan muskler og passive strukturer som leddbånd belastes.

Bevegelsesanalysen vil ta ca. 1,5 time, inkludert anatomiske målinger og påsetting av markører. De andre testene gjennomføres resten av tiden laget er på NIH, og totalt vil testene ta om lag åtte timer. I tillegg til disse testene vil du få utdelt et skjema, der vi spør om treningserfaring, tidligere skader, skade i familien, treningsmengde, menstruasjonsstatus og knefunksjon. Spørreskjemaet besvares i løpet av testdagen, og det vil ta ca. 30 min.

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

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

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

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

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

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

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

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

SAMTYKKEERKLÆRING

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

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

Sted ……………………….. Dato ………………………..

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Underskrift

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Navn med blokkbokstaver

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Adresse

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Mobiltelefon

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E-postadresse

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

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

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

Hva får du ut av det?

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

Angrer du?

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

Spørsmål?

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

NORGES IDRETTSHØGSKOLE

SENTER FOR Idrettsskadeforskning KLOKE AV SKADE

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

SAMTYKKEERKLÆRING

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

☐ Jeg ønsker ikke å bli kontaktet etter endt karriere med tanke på oppfølgingsstudier

☐ Jeg ønsker ikke at video av meg skal brukes i undervisningssammenheng

Sted ……………………….. Dato ………………………..

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Underskrift spiller Underskrift foresatt

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Navn med blokkbokstaver

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Adresse

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Mobiltelefon

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E-postadresse