Kinematic and kinetic analyses of human movement with respect to health, injury prevention and rehabilitation aspects

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1 Introduction

Biomechanics is the application of mechanical principles to biological systems. In sports biomechanics the focus is set on the human body with the aim to improve sportive performance but also to reduce injury (Bartlett, 1999). The forces acting on the human muskuloskeletal system are a major field of interest as they are the reason for locomotion, for deformation of soft and semi-rigid structures, for growth and development of biological tissue, but also for acute and chronic injuries (Nigg, 2000).

In order to realize a movement a force needs to be applied to the specific segment. Forces acting from outside on a system, such as ground reaction forces in human gait or a person's weight, are called external forces. In reaction to the external forces muscles contract and transmit force over the tendons onto bones and onto joint structures. The forces acting within the human body are referred to as internal forces (Nigg, 2000). The direction of the applied external force might not directly act on the joint center of interest, creating a lever arm to the respective joint center. The resulting load is referred to as joint moments and is defined in a (quasi-) static movement as $force \times lever\ arm$. In dynamic situations enhanced calculation methods (inverse dynamics) including moment of inertia, and segment center accelerations can be applied (Nigg & Herzog, 2007). Calculating the internal joint moment is the first step in understanding the forces in the respective joints. For example, when an external knee flexor moment is applied during a squat the athlete needs to produce an internal quadriceps extensor moment to guarantee force equilibrium. Hence, a quadriceps force will be needed to generate this moment, which has further effects on single sub-components of the knee such as tibiofemoral and patellofemoral compression forces or on the forces on the anterior and posterior cruciate ligament (Escamilla, 2001).

External impact forces during activities such as running have frequently been associated with the development of degenerative joint diseases, headaches, prosthetic joint loosening, plantar fasciitis, Achilles tendinitis, muscle tears and stress fractures (Radin et al., 1982; Folman et al., 1986; Pratt et al., 1989; Radin et al., 1991; Milgrom et al., 1992; Whittle, 1999; Nigg, 2000; Mosley, 2000). This paradigm is currently being discussed as more scientific evidence arouses that mechanical stimuli in form of impact forces may not only have detrimental effects but also lead to biopositive adaptation processes in the bone architecture and bone mass (Jones et al., 1977; Rubin et al., 1995; Kersting & Brüggemann, 1999; Mosley, 2000; Kersting et al., 2006; Potthast, 2005). The mechanism that matches bone mass and architecture to functional demand is known as functional adaptation. This is a lifelong process, as the skeleton adapts to changes in mechanical use and as the bone is continually optimized for its load-bearing role. The optimization is based on the process of functionally adaptive remodeling to maintain the inherent safety factor that keeps fracture risk at an acceptable biological level (Alexander, 1981; Biewener, 1993; Mosely, 2000). Repetitive coordinated bone loading, such as occurring at habitual daily life activities, may not have a biopositive effect on osteogenic modeling, but research showed that high-magnitude, high-rate strains, presented in unusual distribution promote this process (Mosley, 2000). The effects of forces acting on the bone have frequently been the focus of research, but little research is available on the mechanisms of the behavior of soft-tissue such as articular cartilage, ligaments and tendons, when exposed to strain (Hudelmaier et al., 2006; Nigg & Herzog, 2007; Arampatzis et al., 2007; Legerlotz et al., 2007; Petrigliano et al., 2007, Eckstein et al., 2008; Mademli & Arampatzis, 2008; Arampatzis et al., 2010; Trattnig et al., 2009). It is assumed that moderate levels of strain enhance the characteristics of the soft tissue, but higher intensities of strain might lead to its damage (Nigg et al., 1995; Kersting, 1997). Yet it remains unclear which intensity of stress on the structure leads to either biopositive or bionegative adaptation processes. A major aim of biomechanical research is to prevent injury. So one is eager to find the balance between the right set of stimuli for biopositive adaptations as needed in rehabilitation as well as in performance enhancement while not exposing the athlete time to inproper loading stimuli in order to prevent overloading of the system at the same time (Potthast, 2005).

This thesis adresses the issue of loading occuring during human movement with respect to different subject groups and settings. Therefore three independent studies with different currently discussed questions of research and methods have been conducted to give insight over the broad spectrum of possible applications of sports biomechanics focusing on prevention and rehabilitation. As subject groups “healthy subjects”, “subjects at risk for overuse injury” and already “injured subjects” have been chosen. Clearly this thesis can only give a small insight of application of sports biomechanics, and there is no claim for completeness.

Chapter 3 investigates the kinematics and kinetics of healthy subjects performing squats with three different techniques. Even though the squat is a frequently used exercise in rehabilitation and sports programs, little research has been established of the effects of the technique on kinematics and kinematics in terms of joint loading. 16 male, sportive active subjects conducted three squat variations: “standard squat”, “knee shifted squat” and “squat on a declined surface”. Kinematic and kinetic data was recorded via an infrared system (VICON, 200 Hz) and two force plates (AMTI, 1000 Hz). Results provide information about the impact kinematic changes induce on joint loading.

Chapter 4 investigates the influence of body weight on joint loading of children climbing stairs. Even though a high prevalence between obesity and degenerative diseases of the musculo-skeletal system is verified, little is known about the joint loading of overweight and adipose persons and especially of adipose children performing daily life activities. Regarding early detection and prevention of degenerative joint diseases, it is of great importance to already gain information on joint loading in children. Especially for this subject group research has yet to be established. Therefore 18 obese children and 17 normal-weight children were recruited to ascend and descend stairs. A Vicon system (VICON, 200 Hz) and two force plates (AMTI, 1000 Hz) collected kinematic and kinetic data. The results give insight where differences occur and help to set up adequate rehabilitation programs to reduce joint overloading in obese children.

Chapter 5 examines the effect of two different functional braces on laxity and functional achievements in patients with ACL ruptures. Functional braces are frequently incorporated during the rehabilitation process of ACL injured knee patients. The discussion whether functional braces have a beneficial effect on the stabilization of the injured knee is argued very controversially. The fact that different types of braces might also have differing effects on joint loading and functional tests adds to the uncertainties. Therefore the aim of this study was to investigate the effects of two different functional braces - a sleeve brace (SofTec Genu, Bauerfeind Germany Inc, Zeulenroda) and a rigid shell brace (4Titude Donjoy, ORMED GmbH, Freiburg) - on joint laxity, proprioception, postural control, lower limb force and gait patterns. 28 subjects with an ACL deficient knee were tested in a sleeve braced, a rigid shell braced and a non-braced condition. Results provide information to which extent different brace types show comparable effects and confirm the supporting requests set on in the rehabilitation of ACL deficient knees used orthoses.

The ability to observe and interpret human movement is limited by the available analysis tools and methods. The progress in instrumentation and computer technologies has provided new possibilities for the advancement of studies in human locomotion and made it feasible to extend the application of kinetic analysis from the sole focus on sportive aspects to the analysis of clinical problems (Andriacchi & Alexander, 2000; Gollhofer & Müller, 2009). Some aspects of human locomotion can be measured directly, e.g. the time a person spends in the stance phase, the maximum isometric or dynamic force a person can apply, or the ground reaction force that occurs at a certain movement. Other aspects withdraw themselves from the possibility to be directly observed, for example the loading of the musculo-skeletal system. Therefore, most of the time it is not possible to measure loading at the respective contact points or biological structures. Models and approximations play an important role generating information about these non-observable parameters. The inter-segmental forces and moments are approximated by modeling the body as a system of rigid segments linked to each other and measuring the three-dimensional position of the segments and external ground reaction forces (Andriacchi & Alexander, 2000). The accuracy of the approximations highly depends on the measurement instruments, the quality of the input data, the assumptions taken in the model and the model itself (Cappozzo et al., 2005). Currently, a frequently used method to analyze the three-dimensional position of segments is to track markers placed on the skin. This approach allows conclusion on the underlying movement of the skeletal structure. Clearly, the uncertainties of skin movement relative to the underlying bone represent a limitation of this approach (Cappozzo et al., 1997; Sati et al., 1996). A variety of models, which differ in marker-position, measured variables, degrees of freedom assigned to the joints, anatomical and technical references, joint rotation conventions and terminology have been introduced to approximate the skeletal movement. Due to the individual specifics differences in results occur which imply difficulties in the comparison between results of different approaches. It is not known to which extent different approaches deviate from each other as it might also depend on the analyzed movement. *Chapter 2* addresses this issue in comparing three currently used models in a sport relevant setting with the study: "The effect of calculating kinematics and kinetics of the squat movement with three different models".

During the last four years on working on this thesis aspects of the specific studies were presented at national and international congresses. Aspects of joint loading of obese children were presented at the "13th Annual Congress of the European College of Sports Science (*ECSS), Estoril 2008"* (Strutzenberger et al., 2008a), the *"Landessymposium Baden Württemberg, 2008"* (Strutzenberger et al., 2008b) and at the "*XXVII Conference of International Society of Biomechanics in Sports (ISBS), Limerick 2009"* (Strutzenberger et al., 2009). Aspects of joint loading in different variations of the squat movement were presented at the "*XXVIII Conference of International Society of Biomechanics in Sports (ISBS), Marquette 2010"* (Strutzenberger et al., 2010a). Aspects of the influence of different types of functional braces on subjects with ACL deficient knees were presented at the *"13.*

Kongress der Österreichischen Sportwissenschaftlichen Gesellschaft (ÖSG) Bruck an der Mur 2010", (Strutzenberger et al., 2010b). This presentation was awarded 2nd place of the "Nachwuchsförderpreis der ÖSG". Additionally, the submitted abstract for the "59. Jahrestagung der Vereiningung Süddeutscher Orthopäden und Unfallchirurgen (VSOU), Baden-Baden 2011" (Strutzenberger et al., 2011) was accepted for participation at the young investigator award.

The chapter of joint loading of obese children was submitted to the international journal "Gait & Posture" and is currently in the second review process. The other studies presented in the different chapters will be submitted to international journals within the next months.

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2 Study I: The effect of calculating kinematics and kinetics of the squat movement with three different models

2.1 Introduction

The description of skeletal movement and the prediction of inter-segmental forces and moments play an important role in the understanding of human movement. Hence, inter-segmental forces and moments cannot be observed directly, models have been introduced to converge on them. The models abstract the body to a system of rigid segments and use measured three-dimensional positions of body-segments and external ground reaction forces as input variables (Andriacchi & Alexander, 2000). A widespread method to measure the segment position is the tracking of markers placed on the skin (Benedetti & Cappozzo, 1994; Andriacchi & Alexander, 2000). Various marker-sets and biomechanical models have been developed which differ in marker-position, measured variables, degrees of freedom assigned to the joints, anatomical and technical references, joint rotation conventions, and terminology (Ferrari et al., 2008; Andriacchi & Alexander, 2000). Charlton et al. (2004) classifies currently existing models into three categories:

<u>(1) Landmark specific marker placement:</u> Markers are accurately placed on palpable bony landmarks and describe the underlying bone geometry by standardized algorithms based on vector algebra. The local frame (x,y,z,) is defined upon the existence of three non-collinear markers, m_1, m_2 and m_3, by taking - for example - m_2 as the origin of the local frame, the z-axis set along the line oriented as m_1-m_2, the x-axis set perpendicular to the plane defined by the three markers and directed as vector $\vec{v} = (m_3\text{-}m_2) \times (m_1\text{-}m_2)$, and the y-axis perpendicular to both the x and z that is directed as $(m_1\text{-}m_2) \times \vec{v}$ (Chiari et al., 2005). This method assumes the markers to be rigidly linked with the bone. The most common known model is the Newington Hospital Helen Hayes model (Davis et al., 1991; Kadaba et al., 1990) implemented e.g. in the Vicon Plug-in Gait model (PiG). This model is frequently used although its dependence on the correct placement of markers and sensitivity to skin artifacts is highlighted (Lu & O’Connor, 1999; Rainbow et al., 2003; Chiari et al., 2005). Due to the direct calculation of marker-placement to the underlying bone and the lack of an optimization approach this method is also referred to as “direct method” (Lu & O’Connor, 1999).

<u>(2) Marker clusters with technical-anatomical calibration:</u> These models use a less specific placement of marker clusters and a technical-anatomical calibration process to manually identify the required landmarks and axes directions to reconstruct the underlying bone and joint geometry. Such models include the Cleveland Clinic method (Motion Analysis Corp, Santa Rosa, USA) and technical-anatomical calibration technique (Cappozzo et al., 1995). They are based on a standard-skeletal model, which is then scaled to fit best into the measured markers (Simonidis, 2010; Lu & O’Connor, 1999). Therefore, Lu & O’Connor (1999) introduced a global optimization method, which is based on the search of an optimal position of the multi-link model for each data frame so that the overall differences between the measured and model-determined marker coordinates are minimized in a least squares sense throughout all body segments. For example this technique is used by Visual3D (V3D) software. Due to the underlying optimization method, these models could also be referred to as local or global “optimization-method” (Lu & O’Connor, 1999).

(3) Estimation of functional joint centers and axes: This approach assumes that the joint centers and axes could be more reliably defined by inferring from the segment movement to a center of rotation (Hindiuma et al., 2002; Leardini et al., 1999). These optimized joint centers are then used in models similar to those of (1) and (2) to deduce joint kinetics and kinematics.

One of the biggest factors limiting all these methods is the inference from the skin placed marker to the movement of the underlying bone. Skin-markers displace and rotate relative to the underlying bone, which causes errors in the estimated segment positions, the magnitude of this error is not known for all these methods (Cappozzo et al., 1997; Sati et al., 1996; Reinschmidt et al., 1997; Holden et al., 1997; Andriacchi & Alexander, 2000; Lu & O'Connor, 1999). Furthermore, the skin movement does not underlie the same rules for each subject. Another limitation is the difficulty in repeatability of marker placement, which also causes differences in kinematics and kinematics (Stagni et al., 2000; Holden & Stanhope, 1998; Gorton et al., 2009; Leardini et al., 2007; Della Croce et al., 1999; 2003; 2005; Tabakin & Vaughan, 2000).

Currently, no standard method exists, which makes it difficult to compare results from studies using different approaches. Additionally, sometimes different models might be in use at the same laboratory. At the BioMotion Center of the Karlsruhe Institute of Technology (KIT) two different models are currently implemented: PiG and MKDtools. While the PiG is commercially available, MKDtools is a framework of multi-body routines within MATLAB and allows the numerical creation and computation of equations of motion for any user defined model. Originally it was developed within the Collaborative Research Center 588 (humanoid robots – learning and cooperating multimodal robots) in order to transfer motion capture data to humanoid robots, but was also applied in human motion analyses (Simonidis, 2010). It is a recursive multi-body algorithm using a model based on Zatsiorsky / Seluyanov parameters (DeLeva, 1996) with a standard skeletal model of 1.77 m, which can be up- or downscaled to individual body height. The joint coordinate trajectories are obtained by solving a non-linear least squares fit in minimizing the distance between the model markers and measured markers (Simonidis & Seemann, 2008) and can therefore be classified as global optimization method.

The objective of this paper is to compare the three different modeling methods "PiG" (direct method), "MKDtools" (optimization method) and "V3D" (optimization method) to get a better understanding of the differences and similarities between the three methods. Previous studies comparing different models mainly focused on clinical gait. For this paper a squat movement was chosen for the analysis to combine the following issues: (1) the movement should be slow motion in order to reduce the effect of skin movement artifacts, which have been proven to affect the accuracy of calculated joint kinematics especially in the frontal and transversal planes (Cappozzo et al., 1996); (2) it should be possible to obtain information about the interaction between the methods and the position over a wide range of motion in a sport-relevant setting.

2.2 Methods

2.2.1 Subjects

Fifteen healthy physically active male students (24.3 ± 2.8 yrs, 1.83 ± 0.06 m, 80.4 ± 7.8 kg) participated in this study. The subjects had no history of lower extremity injuries and were able to perform the exercise pain free and with proper technique for four consecutive repetitions with an additional mass of 20 kg.

2.2.2 Exercise description

Each subject was instructed to perform a high-bar parallel squat with the feet at a hip-wide position and with a barbell of 20 kg placed on the superior aspect of the trapezius. The start and end positions were both the same: standing in upright position with the knees extended. Squats were performed until a knee angle of 90°. Tactile feedback for reaching the 90° angle was given by a horizontal bar, which was prepositioned for each subject at individual height. Each foot was positioned on a separate force platform. Four repetitions were performed at a velocity of four seconds/squat, resulting in an angular velocity of 45°/s at the knee.

2.2.3 Data collection

Kinematic and kinetic recordings were collected simultaneously by a 10 camera, three-dimensional motion analysis system (VICON, MX camera system, Oxford Metrics Ltd, UK; 200 Hz)[a] and two force platforms (AMTI, model BP600900, Advanced Mechanical Technology, Watertown, MA, 1000 Hz) embedded in the floor. Reflective markers were placed according to a modified Helen Hayes marker set to fulfill the requirements of all the different methods: For the PiG approach the markers of the standard Helen Hayes marker set were used, for the MKDtools and V3D approaches additional markers were placed on the olecranon, the medial epicondyles of the knee, the lateral side of the metatarsal V, the medial side of metatarsal I, the medial malleolus, and the barrel were also included in the calculation.

2.2.4 Data reduction

Data was reconstructed and labeled using Vicon Nexus[a] software and exported as c3d-files. Based on these c3d-files kinematic and kinetic data for the hip, knee and ankle were analyzed using PiG, MKDtools and V3D. Joint angles (normalized to time) and joint moments (normalized to body weight & time) of the hip, knee and ankle were calculated in the sagittal, frontal and transversal plane with each approach. Due to joint restrictions of the MKDtools kinematic and kinetic data at the knee and ankle joint in transversal plane do not exist for this approach. All joint angles were filtered with a fourth order Butterworth low-pass filter with a cut-off frequency of 15 Hz. Joint moments were filtered with a fourth order Butterworth low-pass filter with a cut-off frequency of 4 Hz. Due to different joint and moment definitions for the three approaches all parameters were transformed to conform to the same definition (Figure 2.1). Moments were reported as external moments.

The analysis was similar between the limbs for adults without lesion and movement experience (Escamilla et al., 1998). Therefore, only one limb (left) of the subject was considered for further analysis. For each method the joint angles and joint moments of the fourth squat were analyzed at 10°, 30°, 50°, 70°, 90° knee flexion angle (eccentric phase of the squat) and 70°, 50°, 30° and 10° knee flexion angle of the knee (concentric phase of the squat) respectively.

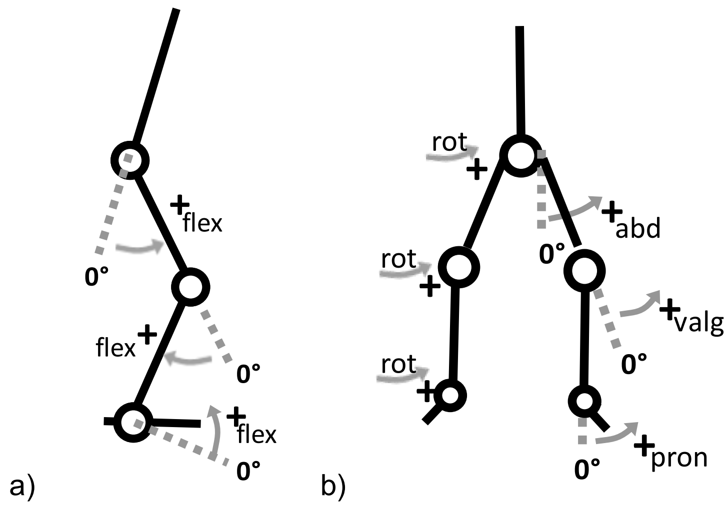

Figure 2.1: Angle and moment definition of the hip, knee and ankle joint.

2.2.6 Data analysis

Joint angles and joint moments of the left side were statistically analyzed to determine characteristics between the three approaches using (a) an Intra-Class-Correlation (ICC) analysis and (b) a two factor (method x angle) repeated measure ANOVA with Bonferroni correction. The pair-wise comparisons were assessed using t-tests as post-hoc tests. Effect sizes were calculated via partial eta² (η^2_p) for overall effect sizes and via Cohens d (*d*) with pooled standard deviation for each pair-wise comparison (Cohen, 1992). An ICC-value between 0.50 and 0.79 was considered as acceptable and an ICC-value ≥ 0.80 was considered to be good. The level of significance was set at $p \leq 0.05$. Borders for partial effect size η^2_p were set to be η^2_p=0.01 for small, η^2_p=0.06 for medium η^2_p=0.14 and for high effect sizes. The effect size *d* was quantified to be medium between 0.40 and 0.79 and high with effect size $d \geq 0.80$. Since the results of the concentric and eccentric phase were identical, only the data for the eccentric squat phase are presented in this paper.

2.3 Results

2.3.1 Kinematics

The specific calculation methods had distinct differing effects on the kinematics. However, results varied between methods and joint angles and no systematic identification of the effects could be made. ICC-values vary from -0.99 to 0.91 (Table 2.1). The ANOVA with repeated measurements showed overall significant differences in the sagittal plane for the hip (p: 0.004, η^2_p: 0.949), in the frontal plane for the knee (p: 0.001, η^2_p: 0.678) and ankle (p: 0.003, η^2_p: 0.838) and in the transversal plane for the hip (p: 0.005, η^2_p: 0.776), knee (p: 0.012, η^2_p: 825) and ankle (p: 0.0038, η_p ²: 0.363). The subsequent post-hoc tests of these variables revealed significant differences in 47 out of 70 parameters (Figure 2.2). Effect sizes higher than 0.40 in 62 out of 70 parameters (Table 2.1) could be observed.

Sagittal plane: The hip flexed from ~20° to ~ 85° during the eccentric phase of the squat. While the PiG and MKDtools models revealed similar hip flexion angles (ICC values ≥ 0.80: 10°-50°,90°; ICC values ≥ 0.40: 70°), on average the V3D model led to significantly lower hip flexion angles compared to PiG (9°± 3°) as well as compared to MKDtools (7°± 2°). Acceptable and good ICC-values between V3D and the other two models occurred at the more flexed positions of 70° and 90° knee angle and at the comparison MKDtools-V3D for the 10° knee angle. The effect size *d* was high for PiG-V3D (10°) and MKDtools-V3D (10°-50°) and medium for all other parameters.

In the eccentric phase of the squat the knee flexed from fully extended to 90°. Since the data for each model was analyzed as a function of knee flexion angle no statistical difference between models occurred. Low ICC and high effect sizes originate in very small standard deviations.

The dorsiflexion angle of the ankle increased over the squatting movement from ~ 6° (PiG, V3D) respectively 1° (MKDtools) to 30° (PiG) respectively 27° (V3D, MKDtools). Even though no overall significant difference was detected ICC values were higher than 0.50 only for the more flexed positions (ICC ≥ 0.50: 70°: PiG-V3D, MKDtools-V3D; 90°:PiG-MKDtools, PiG-V3D, MKDtools-V3D).

Frontal plane: In all three models hip abduction angles from ~14° to ~20° were calculated during the eccentric phase. No overall significant differences could be identified for this joint. The angular characteristics between the three different methods were quite similar as almost all parameters between the three methods revealed at least acceptable ICC-values (PiG-MKDtools: ICC ≥ 0.50: 10°-90°, *d* ≥ 0.40: 10°-90°; PiG-V3D: ICC ≥ 0.50: 10°-90°, *d*: ≥ 0.40: 50°,90°; MKDtools-V3D: ICC ≥ 0.50: 10°-70°, *d*: ≥ 0.40: ---) including also ICC-values higher than 0.8 (PiG-MKDtools: ICC ≥ 0.80: 10°,30°, 70°; PiG-V3D: ICC ≥ 0.80: 10° - 70°; MKDtools-V3D: ICC ≥ 0.80: 10°).

The squat was initiated with the knee being in an almost neutral position of about -3° varus angle. Depending on the used model this angle got more distinct during the squatting movement: The varus angle calculated with the PiG model stayed very stable at an almost neutral position (1°± 6°), using the V3D model it increased up to 11° ± 0.3° and using the MKDtools model it increased up to 22° ± 0.3°. This led to significant differences at 50° between PiG-V3D, at 70° between PiG-V3D and PiG-MKDtools and at 90° between all three models. Low ICC-values, except for the starting position of 10° between PiG-V3D (ICC ≥ 0.50) and high effect sizes in almost all parameters (PiG-MKDtools: *d* ≥ 0.80 10°,50°-90°; PiG-V3D: *d* ≥ 0.40: 10°,50°; *d* ≥ 0.80: 70°,90°, MKD-V3D: *d* ≥ 0.40: 10°,30°; *d* ≥ 0.80: 70°,90°) could be reported.

Concerning the ankle the model in use had a distinct effect on the angular outcome. While the angles calculated with the PiG model settled around 0°, the angles calculated with the V3D model showed a pronation angle up to 16° and the angles calculated with the MKDtools model revealed a supination angle up to 23.6°. For the three models the range of movement was similar with 6° (PiG), 8° (MKDtools) and 11° (V3D). ICC values did not reach an acceptable value and effect sizes also supported this result in being considerably high with *d* from 2.70 to 4.60 for PiG-MKDtools, 0.92 to 2.41 for PiG-V3D and 3.00 to 4.48 between MKDtools and V3D.

Transversal plane: For the hip the three methods showed distinct differences. At the beginning of the squatting movement the leg was internally rotated at the hip for the PiG and V3D method. When the deepest squatting position was reached this changed into an external rotation. For the MKDtools model angular results showed an external rotation movement throughout the whole squatting movement though. These results showed an average difference of 25° ± 3° between the MKDtools and the PiG method, an average difference of 18° ± 6° between the MKDtools and the V3D method, and an average difference of 7° ± 3° between PiG and V3D. Significant differences appeared at all parameters between the PiG model and MKDtools model, as well as with a knee flexion angle bigger than 50° between PiG and V3D and until a knee flexion angle of 70° between MKDtools and V3D. No acceptable ICC values occurred in this plane and considerable effect sizes appeared between 2.10 and 5.25 for the comparison of PiG-MKDtools, between 0.55 and 1.28 for the comparison of PiG-V3D, and between 0.90 and 3.89 for the comparison of MKDtools-V3D.

During the eccentric phase of the squat the knee rotated internally approximately 32° using the PiG model and approximately 22° using the V3D model. The starting position was neutral for the PiG model and 7° externally rotated for the V3D model. Hence, the comparison between the two models revealed significant differences at all parameters except in the starting position. On average the knee was more internally rotated by 15° ± 5° using the PiG model than using the V3D model. This difference was supported by ICC- values lower than 0.40 and effect sizes higher than 0.80 for all positions.

The starting position of the ankle was internally rotated. During the eccentric phase of the squat this position changed into external rotation. The calculated angles via V3D were generally higher by 15° ± 5.6° with respect to internal rotation than the calculated angles via PiG. This was also expressed by ICC values ≤ 0.40 for all position and effect sizes ≥ 0.80 for all parameters. Additionally, this corresponded with the transversal parameters at the knee, since a more externally rotated knee (V3D) needs to be compensated by a more internal rotated ankle (V3D).

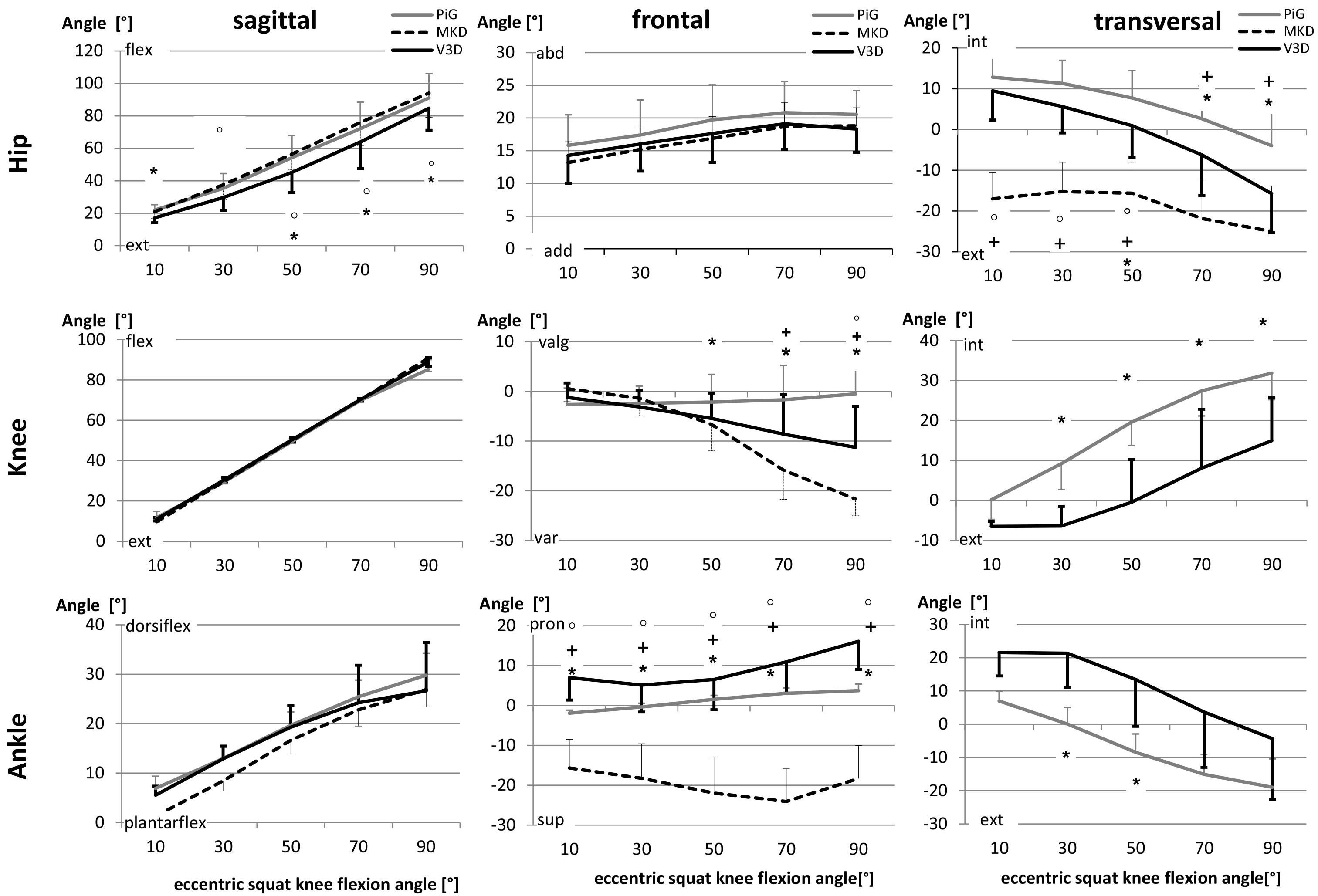

Figure 2.2:: Hip, knee and ankle angles in respect to 10°,30°,50°,70°,90° of the knee flexion angle at the eccentric phase of the squat calculated via three different methods: gray line: PiG; dotted black line: MKDtools; solid black line: V3D. Significant differences were marked as: “+“ PiG – MKDtools, “*” PiG-V3D, “°” MKDtools-V3D

Table 2.1: ICC-values, p-values (ANOVA) and effect sizes (Cohens d) between three methods (MKD, V3d, PiG) and each parameter with respect to 10°,30°,50°,70°,90° of the knee flexion angle at the eccentric phase of the squat. ICC: dark grey: ICC ≥ 0.80; light gray: ICC ≥ 0.50; p-values: light gray: p ≤ 0.017 (Bonferroni corrected significance level); Cohen's d: light gray: *d* ≥ 0.40, dark gray: *d* ≥ 0.8.

ANGLES	Knee Angle	ICC: PiG-MKD	ICC: PiG-V3D	ICC: MKD-V3d	p: PiG-MKD	p: PiG-V3D	p: MKD-V3d	abs Cohens d: PiG-MKD	abs Cohens d: PiG-V3D	abs Cohens d: MKD-V3d
Hip ant.-post.	10°	0.82	0.30	0.57	0.086	0.000	0.029	0.271	1.493	1.054
	30°	0.80	0.24	0.39	0.769	0.000	0.000	0.239	0.676	0.926
	50°	0.85	0.36	0.45	0.318	0.000	0.000	0.183	0.692	1.002
	70°	0.63	0.58	0.84	0.601	0.000	0.000	0.250	0.508	0.785
	90°	0.85	0.81	0.61	0.946	0.007	0.001	0.200	0.440	0.648
Hip med.-lat.	10°	0.85	0.90	0.82	---	---	---	---	---	---
	30°	0.82	0.91	0.74	---	---	---	---	---	---
	50°	0.79	0.83	0.77	---	---	---	---	---	---
	70°	0.81	0.77	0.75	---	---	---	---	---	---
	90°	0.56	0.75	0.38	---	---	---	---	---	---
Hip rot.	10°	-0.77	-0.39	-0.92	0.000	0.235	0.000	5.254	0.551	3.893
	30°	-0.74	-0.36	-0.79	0.000	0.028	0.000	4.110	0.941	3.058
	50°	-0.62	-0.23	-0.62	0.000	0.016	0.004	3.314	0.934	2.180
	70°	-0.61	-0.14	-0.47	0.000	0.006	0.042	2.796	0.980	1.611
	90°	-0.59	-0.10	-0.54	0.008	0.012	0.188	2.096	1.276	0.893
Knee ant.-post.	10°	-0.11	-0.37	-0.19	---	---	---	---	---	---
	30°	0.10	-0.37	-0.37	---	---	---	---	---	---
	50°	0.31	-0.30	0.15	---	---	---	---	---	---
	70°	0.23	0.38	0.20	---	---	---	---	---	---
	90°	-0.34	-0.31	-0.04	---	---	---	---	---	---
Knee med.-lat.	10°	-0.53	0.75	-0.70	0.417	0.117	0.248	1.079	0.455	0.648
	30°	-0.20	0.28	-0.43	0.829	0.071	0.182	0.299	0.204	0.508
	50°	-0.13	-0.16	-0.17	0.248	0.012	0.261	0.827	0.617	0.230
	70°	-0.49	-0.26	-0.10	0.009	0.004	0.787	2.203	0.929	1.037
	90°	-0.82	-0.48	-0.46	0.000	0.010	0.017	4.023	1.433	1.642
Knee rot.	10°		0.28			0.235			1.853	
	30°		-0.31			0.000			2.718	
	50°		-0.34			0.000			2.333	
	70°		-0.37			0.009			1.705	
	90°		-0.70			0.013			1.873	
Ankle ant.-post.	10°	-0.59	0.19	-0.58	---	---	---	---	---	---
	30°	-0.50	0.12	-0.16	---	---	---	---	---	---
	50°	0.14	0.44	0.23	---	---	---	---	---	---
	70°	0.43	0.51	0.58	---	---	---	---	---	---
	90°	0.65	0.51	0.56	---	---	---	---	---	---
Ankle med.-lat.	10°	-0.56	-0.66	-0.93	0.000	0.000	0.000	2.703	2.220	3.520
	30°	-0.69	-0.28	-0.90	0.000	0.000	0.000	2.897	1.132	3.000
	50°	-0.75	-0.14	-0.91	0.000	0.003	0.000	3.669	0.922	3.420
	70°	-0.81	-0.33	-0.94	0.000	0.000	0.000	4.608	1.404	4.361
	90°	-0.81	-0.67	-0.99	0.000	0.004	0.000	3.704	2.410	4.482
Ankle rot.	10°		-0.09			0.065			2.719	
	30°		-0.51			0.000			2.639	
	50°		-0.55			0.003			2.045	
	70°		-0.42			0.034			1.503	
	90°		-0.58			0.170			1.027	

2.3.2 Kinetics

The different calculation methods also had distinct effects on the kinetics of the movement. As in kinematics the results varied between models and respective joint, hence no systematic identification of the effects could be made. ICC-values varied from -0.96 to 0.93 (Table 2.2). The ANOVA with repeated measurements showed overall significant differences in the sagittal plane for the hip (p: 0.002, η^2_p: 0.823) and knee (p: 0.001, η^2_p: 0.884) and in the transversal plane for the hip (p: 0.013, η^2_p: 0.812) and knee (p: 0.001, η^2_p: 0.952). No overall significant differences were identified in the frontal plane. The subsequent post-hoc tests for the resulting parameters revealed significant differences in 39 out of 65 parameters (Figure 2.3) and effect sizes higher than 0.40 occurred in 54 out of 65 parameters (Table 2.2).

Sagittal plane: The hip flexion moment increased from ~ 0.2 Nm/kg to ~ 0.8 Nm/kg during the eccentric phase of the squat. In more detail the method V3D showed on average a 0.11 ± 0.05 Nm/kg lower hip flexion moment than the PiG model and a 0.12 ± 0.04 Nm/kg lower hip flexion moment than the MKDtools model; even though significant differences appeared in most parameters between MKDtools/PiG and V3D effect sizes were considerable high (PiG-V3D: d ≥ 0.80: 10°-70°; MKDtools-V3D: d ≥ 0.40 10°-90°, d ≥ 0.80: 70°). ICC values for PiG-V3D were ≥ 0.5 for all parameters and for MKDtools-V3d ≥ 0.5 from 50°-90° knee flexion angle. A different picture of this variable was presented at the comparison between PiG and MKDtools: The calculated hip flexion moment via PiG and MKDtools was significantly different at the 10° and 30° knee flexion angle, but the values aligned in a more flexed knee position starting with 50°. This was also supported by ICC values higher than 0.80 (50°, 70°) and a Cohen's *d* lower than 0.40 for the position of 50°-90° knee flexion angle.

The squat started with a low knee extension moment (PiG: -0.18 ± 0.18 Nm/kg; MKDtools: -0.19 ± 0.14 Nm/kg, V3D: 0.05 ± -0.10 Nm/kg), which changed into a flexion moment at ~ 30° knee flexion angle and continued up to a maximum knee flexion moment of 0.82 ± 0.14 Nm/kg (PiG), 0.90 ± 0.15 Nm/kg (MKDtools) and 1.07 ± 0.15 Nm/kg (V3D) respectively at 90° knee flexion angle. The V3D model showed the highest knee flexion moments. On average the knee flexion moment calculated via the V3D model was increased by 0.24 ± 0.06 Nm/kg compared to the PiG model and increased by 0.17 ± 0.02 Nm/kg compared to the MKDtools, whilst the PiG and MKDtools models differed on average by 0.07 ± 0.05 Nm/kg. This was supported by the statistical analysis, comparing V3D and PiG, which showed significant differences at three parameters (10°, 70°, 90°).The comparison between V3D and MKDtools models showed significant differences at four parameters (10°, 50°-90 and the comparison between PiG-MKDtools showed differences at two parameters (10°, 70°). Also in terms of ICC values these results were supported: The comparison V3D and PiG revealed - apart from one medium ICC value at 10° knee flexion angle - low ICC values and the comparison V3D to MKDtools revealed two medium ICC values at 10° and 70° knee flexion angle. Comparing PiG to MKDtools medium ICC values were observed for all parameters.

Ankle extension moments were relatively low with peak magnitudes of 0.29 ± 0.21 Nm/kg (PiG), 0.28 ± 0.12 Nm/kg (MKDtools) and 0.29 ± 0.21 Nm/kg (V3D). Until the maximum knee flexion angle was reached magnitudes decreased approximately by 0.12 Nm/kg for each model. ICC values were generally acceptable or higher for all three models. Good ICC values appeared at the comparison of the models PiG-MKDtools (70°), PiG-V3D (10°, 70°, 90°) and MKDtools-V3D (30°-90°).

Frontal plane: The peak hip abduction moment was 0.10 ± 0.16 Nm/kg for the PiG model, it stayed quite neutral for the MKDtools with 0.04 ± 0.26 Nm/kg, and reached 0.22 ± 0.21 Nm/kg for the V3D model. High standard deviations occurred at each parameter and no significant overall differences between methods were identified, but low ICC values indicated no systematic correlation between the three methods.

No overall significant differences were analyzed for frontal plane knee moments between the three methods. For the MKDtools model the moments stayed quite neutral until 30°

(~0.01 ± 0.11 Nm/kg), while for the PiG (-0.08 ± 0.10 Nm/kg) and V3D (-0.03 ± 0.07 Nm/kg) models varus moments were identified. With increasing knee flexion angle the varus moment also increased (90°: PiG: -0.37 ± 0.11 Nm/kg; MKDtools: -0.22 ± 0.39 Nm/kg, V3D: -0.51 ± 0.21 Nm/kg). Even though no overall significance was manifested, generally low ICC values indicated low correlations between the three methods.

At the ankle all three methods showed a supination moment during the squat. At the beginning of the squat the model PiG calculated the lowest magnitudes with 0.03 ± 0.10 Nm/kg, MKDtools the highest with 0.14 ± 0.07 Nm/kg. The V3D model was aligned between the other two models with values of 0.10 ± 0.06 Nm/kg. Even though these supination moments changed little over the eccentric phase of the squat (PiG and MKDtools +0.04 Nm/kg; V3D +0.02 Nm/kg) ICC values were low for 10°-30° and medium for 70° and 90° knee angle.

Transversal plane: While at the hip the V3D and PiG models showed an internal rotation moment, the MKDtools model revealed an external rotation moment with a difference in joint loading of approximately 0.18 ± 0.05 Nm/kg. Significant differences were analyzed between all three models in almost each knee flexion angle except at 10° between PiG and V3D and at 90° between PiG and MKDtools as well as between V3D and MKDtools. Good and acceptable ICC values were analyzed for the comparison between the PiG and V3D models (ICC ≥ 0.50: 30°-90°; ICC ≥ 0.8: 10°), even though the effect size *d* was still high between those two parameters (*d*: ≥ 0.40; 70° *d*: ≥ 0.8; 10°,50°,90°). Between MKDtools and the other two models the effect size *d* was consistently higher than 0.80 due to the different directions of the joint loading. At the beginning of the squat effect sizes reached values up to 5.89 and got smaller as the movement continued (90°: PiG-MKDtools: 1.44, MKDtools-V3D: 2.45).

Both models calculating transversal plane moments revealed a low internal rotation moment acting on the knee. The internal rotation moment at the beginning of the squat was 0.05 ± 0.02 Nm/kg (PiG) and 0.02 ± 0.01 Nm/kg (V3D) respectively. It slightly increased during the eccentric phase of the squat by 0.05 Nm/kg (PiG) and 0.03 Nm/kg (V3D). On average the PiG model showed 0.02 ± 0.03 Nm/kg significantly higher internal rotation moments with medium (30°-70°) and high (10°, 90°) effect sizes *d*. In terms of ICC values, medium (10°, 70°, 90°) and high (30°, 50°) ICC values were calculated.

Ankle internal rotation moments were also very low. At the PiG model an internal rotation moment of 0.05 ± 0.02 Nm/kg acted on the ankle at the beginning of the squat and increased to 0.10 ± 0.08 Nm/kg until the 90° knee flexion angle. Using the V3D model the internal rotation moment stayed quite neutral and decreased from 0.03 ± 0.07 Nm/kg to 0.01 ± 0.07 Nm/kg over the eccentric phase of the squat. ICC values were low except for the 50° knee flexion position at which the ICC value was acceptable.

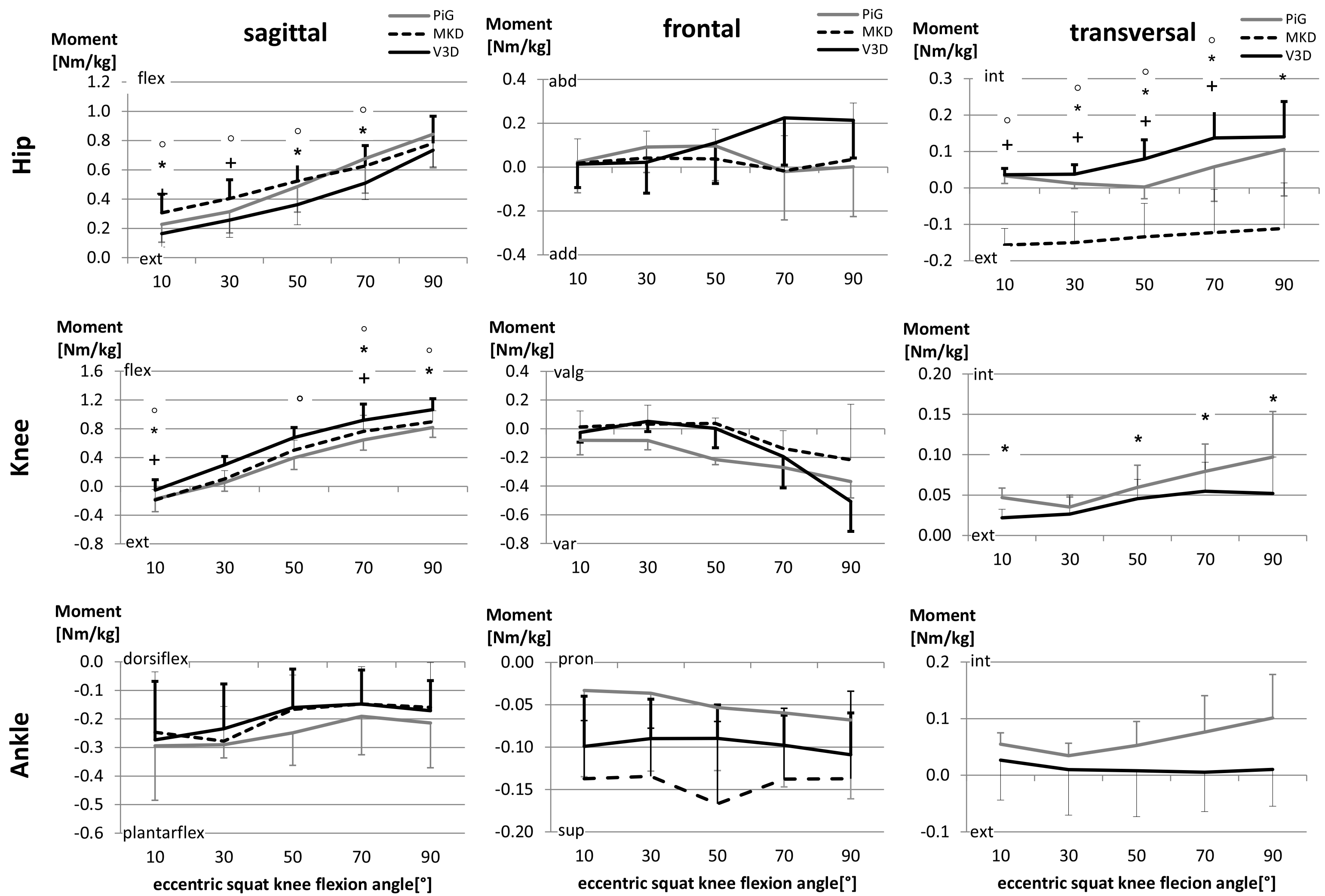

Figure 2.3: Hip, knee and ankle moments in respect to 10°,30°,50°,70°,90° eccentric phase of the knee flexion angle, via three different methods: gray line: PiG; dotted black line: MKDtools; solid black line: V3D. Sign. differences marked as: "+" PiG – MKDtools, "*" PiG-V3D, "°" MKDtools-V3D

Table 2.2: ICC-values, p-values (ANOVA) and effect sizes (Cohen's d) between three methods (MKD, V3d, PiG) and each parameter with respect to 10°,30°,50°,70°,90° of the knee flexion angle at the eccentric phase of the squat. ICC: dark grey: ICC ≥ 0.80; light gray: ICC ≥ 0.50; p-values: light gray: p ≤ 0.017 (Bonferroni corrected significance level); Cohen's d: light gray: *d* ≥ 0.40, dark gray: *d* ≥ 0.8.

MOMENTS	Knee Angle	ICC: PiG-MKD	ICC: PiG-V3D	ICC: MDK-V3d	p: PiG-MKD	p: PiG-V3D	p: MKD-V3d	abs Cohensd: PiG-MKD	abs Cohensd: PiG-V3D	abs Cohensd: MDK-V3d
Hip ant.-post.	10°	0.72	0.72	0.35	0.001	0.002	0.000	0.640	1.292	0.587
	30°	0.51	0.73	0.47	0.002	0.058	0.000	0.682	1.194	0.431
	50°	0.86	0.65	0.55	0.265	0.000	0.000	0.245	1.191	0.788
	70°	0.93	0.65	0.76	0.309	0.000	0.000	0.262	0.919	0.908
	90°	0.71	0.66	0.76	0.123	0.049	0.026	0.296	0.305	0.600
Hip med.-lat.	10°	-0.59	-0.72	0.91	---	---	---	---	---	---
	30°	-0.61	-0.76	0.52	---	---	---	---	---	---
	50°	-0.52	-0.69	0.30	---	---	---	---	---	---
	70°	-0.40	-0.76	0.19	---	---	---	---	---	---
	90°	-0.39	-0.57	-0.62	---	---	---	---	---	---
Hip rot.	10°	-0.78	0.84	-0.73	0.000	0.894	0.000	5.824	0.165	5.896
	30°	-0.59	0.69	-0.63	0.000	0.000	0.000	2.677	0.914	2.901
	50°	-0.57	0.50	-0.71	0.005	0.000	0.000	2.054	1.966	3.025
	70°	-0.90	0.72	-0.93	0.008	0.000	0.001	1.675	0.700	2.252
	90°	-0.91	0.73	-0.95	0.136	0.005	0.022	1.435	0.822	2.446
Knee ant.-post.	10°	0.78	0.67	0.63	0.000	0.008	0.003	0.067	0.877	1.096
	30°	0.68	0.08	0.28	0.023	0.345	0.055	0.467	2.358	1.957
	50°	0.57	-0.10	0.27	0.718	0.140	0.017	0.711	1.926	1.298
	70°	0.73	0.13	0.50	0.006	0.005	0.000	0.620	1.569	0.728
	90°	0.77	-0.25	0.26	0.064	0.000	0.000	0.554	1.760	1.133
Knee med.-lat.	10°	0.19	-0.62	0.24	---	---	---	---	---	---
	30°	-0.52	-0.96	0.41	---	---	---	---	---	---
	50°	-0.90	-0.78	0.54	---	---	---	---	---	---
	70°	-0.66	-0.58	0.55	---	---	---	---	---	---
	90°	-0.46	-0.24	0.01	---	---	---	---	---	---
Knee rot.	10°		0.56			0.008			2.240	
	30°		0.83			0.039			0.462	
	50°		0.89			0.001			0.551	
	70°		0.72			0.001			0.706	
	90°		0.59			0.000			0.883	
Ankle ant.-post.	10°	0.54	0.91	0.54	---	---	---	---	---	---
	30°	0.78	0.72	0.92	---	---	---	---	---	---
	50°	0.74	0.70	0.88	---	---	---	---	---	---
	70°	0.89	0.83	0.91	---	---	---	---	---	---
	90°	0.69	0.82	0.83	---	---	---	---	---	---
Ankle med.-lat.	10°	0.16	-0.43	0.22	---	---	---	---	---	---
	30°	-0.05	-0.14	-0.11	---	---	---	---	---	---
	50°	0.24	-0.02	-0.02	---	---	---	---	---	---
	70°	0.59	0.01	0.14	---	---	---	---	---	---
	90°	0.55	0.29	0.30	---	---	---	---	---	---
Ankle rot.	10°		0.49		---	---	---	---	---	---
	30°		0.49		---	---	---	---	---	---
	50°		0.55		---	---	---	---	---	---
	70°		0.37		---	---	---	---	---	---
	90°		0.05		---	---	---	---	---	---

2.4 Discussion

The aim of this paper was to compare three different analysis models (PiG, MKDtools and V3D) in order to get a better understanding of the differences and similarities between these methods. Due to the underlying calculation and optimization approaches it was expected that MKDtools and V3D as representatives of the optimization method would lead to different results in kinematic and kinetics in contrast to the direct method with its representative, the PiG model. The majority of previous studies, which reported the kinetics and kinematics of movements with different angular conventions showed that the differences in both adduction/abduction (frontal plane) and internal/external rotation (transversal plane) angles were substantial (Ferrari et al., 2008; Lu & O'Connor, 1999; Cappozzo et al., 2005), while only minor differences were observed in the flexion/extension angles (sagittal plane). With respect to inter-session and inter-assessor reliability McGinley et al. (2009) summarized similar results for kinematic data with very high reliability indices (CMC and CMD) for sagittal plane data, and lowest reliability indices for knee frontal plane as well as hip, knee and foot transversal plane data. Excluding the hip in sagittal and frontal plane these general findings could be supported by this study. However, deviant results were reported for the hip, which showed lower values using the V3D model in the sagittal plane and no differences in the frontal plane. This was in line with the findings by Schwartz et al. (2004), who reported the frontal plane being the most reliable for the hip angles overall.

With respect to the different models not one single couple emerged to be consistently the most correlated or the one with the least differences neither between planes nor joints. Four different patterns of the comparison between methods were identified: (1) no differences (e.g. hip, frontal plane), (2) consistent differences over the entire range of motion (e.g. hip, sagittal plane), (3) differences depending on the flexion angle of the knee (e.g. knee varus angle) and (4) differences in the angular characteristics (e.g. ankle, frontal plane, pronation vs. supination).

Regarding the kinematic parameters methods were comparable for knee and ankle flexion angles, but concerning the hip the V3D model revealed significant differences compared to the other two models. For some kinematic parameters, differences between the three methods exceeded 10° over the whole squatting maneuver (e.g. MKDtools to PiG and V3D at hip rotation angles). With respect to clinical biomechanics joint angle differences of 5° can already determine whether the movement is still healthy or already pathological (McGinley et al., 2009). Therefore, it is substantial to identify whether differences occur due to measurement errors and the used model or occur due to the actual movement of the patient. In addition opposing rotational angular directions occurred at the hip. This phenomenon was also reported by Ferrari et al. (2008) for the knee abduction/adduction angle between five different models.

Even though a variety of studies examined the kinematic differences with respect to different models, few studies included joint moments in their analysis. (Ferrari et al., 2008; Leardini et al., 2007; Whatling et al., 2009; Charlton et al., 2004). In general, joint kinetics showed fewer inter-model differences (Ferrari et al., 2008; Leardini et al., 2007; Whatling et al., 2009; Charlton et al., 2004) than in kinematics, which was also supported by this study.

A main factor in calculating joint moments is the position of the ground reaction force vector in relation to the respective joint center. Since the input variables of the ground reaction force were identical for all three approaches the differences in joint moments had to be due to different results in joint center calculations. It also appeared that differences in these calculations had a higher influence at proximal joints than at distal joints. At the ankle joint no significant differences for the moments were identified. At the knee joint as well as at the hip joint significant differences in the moments of the sagittal and the transversal plane appeared. High differences up to 0.15 Nm/kg throughout the motion could be observed at the sagittal (V3D to PiG and to MKDtools at knee), frontal (PiG to MKDtools at the knee) and transversal plane (MKDtools to PiG and to V3D at the hip) between different couples of models.

In previous studies landmark identification, marker placement, and data reduction were identified to affect considerably the calculation of kinematic and kinetic variables (Cappozzo et al., 1996; Gorton et al., 2009; Della Croce et al., 1999, 2003, 2005; Tabakin & Vaughan, 2000; Leardini et al., 2007; Lu & O'Connor, 1999). Yet, uncertainty exists to which extent these differences were caused by the different models, marker-definitions or by the relevant skin artifacts. Lu & O'Connor (1999) and Charlton et al. (2004) propagated a reduction of skin artifacts due to optimization methods. In this study a movement was chosen where only low skin artifacts could be expected due to small soft tissue movement relative to the bony structures. Nevertheless, significant differences appeared between the methods. It is speculated that inter-model differences will increase in more dynamic activities, where higher skin movements will occur.

A second issue is the determination of the joint center location, which is related to marker placement and to the underlying joint model. Additionally, it has been shown that inaccuracies in the hip joint center calculation affect both angles and moments at the hip and the knee (Stagni et al., 2000). This is especially critical for the PiG model as a hierarchical method, which means that errors in hip joint center location will be transferred down to the knee and ankle joint and might be a reason for the kinematic and kinetic differences in sagittal plane at the hip between V3D and PiG.

Based on the findings of the present paper, it is impossible to give an indication of which convention is the best to use. However, this data underlined the fact that a comparison of results among models must be handled very carefully. Lu & O'Connor (1999) concluded that methods based on a concept similar to that of their general optimization method reduce measurement errors and may be useful, for instance, in clinical gait analysis. Based on the in the present study identified differences between V3D and MKDtools, as representatives of the optimization method, it is indicated that a generalization of models using an optimization approach is not possible.

Clearly the three different models serve different needs and should be used adequately. While V3D and MKDtools reduce skin artifacts and offer more precise frontal and transversal kinematics and kinetics in healthy subjects, the PiG-model, with the disadvantage of its dependency on the correct marker placement and its sensitivity to skin artifacts, is still a frequently used model and allows data comparison over a variety of studies. Additionally, for long term studies, where data has been originally obtained by one model (e.g. the PiG at gait laboratories) a change to another method without recalculation the existing data with the new method - if possible - seems to be crucial. For the V3D and MKDtools approach one has to be aware that joint translation are generally treated as artifacts, which might be a limitation for knee motion (Lu & O'Connor, 1999). Also bone deformities of the subjects such as a shorter leg or special joint constraints are not considered in the underlying standard skeletal model in the first place, but have to be programmed separately.

A limitation of this study might be represented by the marker-design. To use the least possible amount of markers it was disclaimed to add additional rigid marker clusters for the V3D approach onto the subjects. Instead, required markers, which are usually represented by the rigid clusters, were identified by either already existing markers of the Helen Hayes model or by additional markers placed on the subject; e.g. the thigh segment was represented by the hip joint center (virtual), the thigh marker (anatomical) and the lateral and medial epicondyle of the knee (anatomical). Since these markers were not removed for dynamical trials, this approach is still able to capture the movement, even though this does not match the ideal cluster placement as propagated by Cappozzo et al. (1997).

2.5 Conclusion

To our knowledge the present study is the first to analyze the differences between the methods V3D, MKDtools and PiG in a sport-relevant setting. It clearly shows that different models lead to distinct differences in kinematic and kinetic outcomes, which implicates that comparing data from different analysis methods has to be handled with great caution. Especially for interpreting and comparing cross sectional studies, specifications of the used model have to be made in order to enable a good judgment of the data.

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3 Study II: Influence of squat technique on lower extremity joint kinematics and kinetics

3.1 Introduction

Squats are commonly incorporated in general fitness training and in rehabilitation programs (Escamilla et al., 1998; Gullett et al., 2008) due to their close biomechanical and neuromuscular similarity to sportive movements such as running and jumping (Flanagan et al., 2003). In addition, the squat and its variations are recommended exercises for the rehabilitation of injured patients and for the elderly, who wish to maintain their functional ability for further physical independence (Flanagan et al., 2003; Salem et al., 2003). Muscle training always has to be considered with respect to functional and adequate joint loading and injury prevention. Although controversy exists concerning the safety of the squat exercise, the majority of research agrees that this exercise is both safe and effective when performed correctly (e.g. Fry et al., 2003; Chandler & Stone, 1991; Escamilla, 2001; Escamilla et al., 2001; 2009a). With improper technique particularly the knees and low back are at risk (Chandler & Stone, 1991; Dunn et al., 1984). Therefore, it is recommended that the lifter maintains a normal lordotic posture and keeps the torso as vertical as possible throughout the entire lift (Chandler & Stone, 1991).

Also, it is recommended that the forward movement of the knees during the squat should not pass the vertical line of the toes in order to reduce the mechanical loading of the knee (Abelbeck, 2002; Fry et al., 2003; Hirata & Duarte, 2007; Escamilla et al., 2009a; 2009b). But limited research has investigated the mechanical loading including hip, knee and ankle moments, specifically when the knee is positioned in front of the toe (Abelbeck, 2002; Fry et al., 2003; Hirata & Duarte, 2007). The authors report that the benefit of decreased mechanical loading around the knee joint is at the expense of an increase in hip moment. However, Abelbeck (2002) and Fry et al. (2003) investigated the hip and knee joint moments with a two-dimensional and static approach. Hirata & Duarte (2007) conducted a three-dimensional approach, but only reported sagittal plane results. Limited research has been performed considering variations of squatting in a three-dimensional setting including hip, knee, and ankle moments.

A traditional squat is performed on an even surface, variations of this squat include squatting on a declined surface. From a clinical perspective the declined squatting version has been proposed to be more effective than squatting on an even surface in the conservative treatment of patella tendinopathy due to a higher load on the knee and consequently the patellar tendon (Purdam et al., 2004; Young et al., 2005; Kongsgaard et al., 2006; Frohm et al., 2007; Richards et al., 2008). The declined squat positions the center of mass further behind the knee joint axes, hence, the knee extensor moment and thereby the load on the patella tendon (Kongsgaard et al., 2006) is increased. Research has yet to establish whether the changes in body position and movement pattern are associated with an increase in muscle activation (Purdam et al., 2004; Frohm et al., 2007) or joint loading (Kongsgaard et al., 2006). The latter would indicate the same mechanisms as for the knee being anteriorly pushed past the virtual vertical line of the toes increasing the displacement between joint center and applied force. Richards et al. (2008) reported an increase in knee joint moments with an increase in decline angle between 0° and 16°. Fortenbaugh et al. (2010) identified a more erect trunk posture with weightlifting shoes, which is comparable to standing on a 3 cm shim (~10° decline). A more erect trunk posture is desired to decrease hip moments (Abelbeck et al., 2002; Fry et al., 2003).

The kinematic changes of squatting on a declined surface indicate similar mechanisms on the joint loading as for the knee being anteriorly pushed past the virtual vertical line of the toes. Therefore, it is hypothesized that the knee-shifted squat has a similar effect on joint loading than squatting on a declined surface. Hence, the aim of this study was to analyze joint moments of three squatting variations representing a standard squat, a squat with the knee being shifted anteriorly over the virtual vertical line of the toe ('knee-shifted squat') and a squat with elevated heels.

3.2 Methods

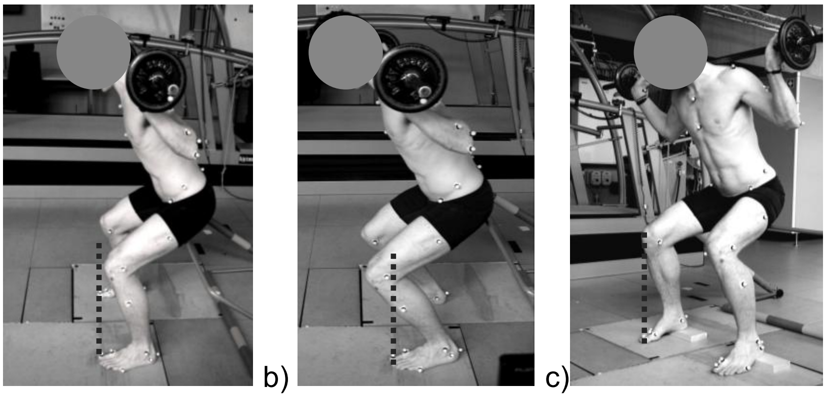
3.2.1 Subjects

Sixteen healthy male physically active students (25.1 ± 2.2 years, 183.0 ± 7.6 cm, 80.3 ± 7.6 kg) participated in this study. The subjects had no history of lower extremity injuries and were required to be able to perform the exercise pain free and with proper form and technique for four consecutive repetitions with an additional mass of 20 kg. The volunteers gave written informed consent before data collection and the study was approved by the institutional ethical review board.

3.2.2 Exercise description

Each subject was instructed to perform a high-bar parallel squat with a barbell of 20 kg placed on the superior aspect of the trapezius. 20 kg were chosen as possible load used in a rehabilitation setting. The starting and ending position was the same for each technique variation, involving standing in an upright position with the knee fully extended and the feet in a hip-wide position. Squats were performed to a knee angle of 90°. Tactile feedback for reaching the 90° angle was given by a horizontal bar, which was prepositioned for each subject at individual height. Each foot was positioned on a separate force platform. Eight repetitions at each variation were performed at a velocity of four seconds/squat, resulting in an angular velocity of 45°/s at the knee. A five min resting period was given between each technique variation. A metronome was used to ensure that the knee flexed and extended at the given velocity. The subjects had to perform the squat in three technique variations (Figure 3.1) in randomized order: the standard squat, the knee-shifted squat and the declined squat.

For the standard squat each subject was instructed to perform a standard half squat with special focus on the knee not being moved over a virtual vertical line of the toes by keeping the weight towards the heels and rather "sitting back" than shifting forward (Figure 3.1a). For the knee-shifted squat subjects were instructed to shift the knees anteriorly across a virtual vertical line of the toes (Figure 3.1b). For the declined squat a wooden shim with the height of three cm was placed under the heels. The instructions given to the subjects were the same as for the standard squat (Figure 3.1c). All three technique variations were controlled by visual observation.



a) b) c)

Figure 3.1: Technique variations: a) standard squat, b) knee-shifted squat, c) declined squat.

3.2.3 Data collection

The three squat variations were performed in randomized order. Kinematic and kinetic recordings were collected simultaneously with a 10 camera, three-dimensional motion analysis system (VICON, MX camera system, Oxford Metrics Ltd, UK; 200 Hz)[a] and two force platforms (AMTI, model BP600900, Advanced Mechanical Technology, Watertown, MA, 1000 Hz) embedded in the floor. Reflective markers were placed according to a modified Helen Hayes marker set (Kadaba et al., 1990) including additional markers on the olecranon, the medial epicondyles of the knee, the lateral side of the metatarsal V, the medial side of metatarsal I, the medial malleolus and on the barrel.

3.2.4 Data analysis

Data was reconstructed and labeled using ViconNexus[a] software and exported as c3d-files. Based on these coordinates kinematic and kinetic data for the hip, knee and ankle were analyzed using the recursive multi-body algorithm MkdTools (Simonidis & Seemann, 2008) and a model based on Zatsiorsky/Seluyanov parameters (DeLeva, 1996). Joint angles and joint moments of the hip, knee and ankle were calculated in the sagittal and frontal plane. Kinematic and kinetic data in transverse plane was only calculated for the hip due to the knee and ankle joint restrictions of the MKDtools method. All data was filtered using a fourth order Butterworth low-pass filter with a cut-off frequency of 15 Hz. In joint angles positive values indicate flexion, adduction and external rotation, while in joint moments positive values indicate an extension moment, abduction moment and external rotation moment. Moments are normalized to body mass and reported as internal moments.

Escamilla et al. (1998) showed that for movement experienced adults without lesion, results are similar between the limbs, therefore only one limb (left) of the subject was considered for analysis. The joint angles and joint moments of each squat repetition were identified at 10°, 30°,50°, 70°, 90° flexion angle of the knee (eccentric phase of the squat) and 70°, 50°, 30° and 10° flexion angle of the knee (concentric phase of the squat). Overall averages were calculated for each subject and taken for further calculation.

In order to consider the high amount of parameters an overall significance over each joint and the respective knee angles was analyzed using a two factor (technique x angles) repeated measurement ANOVA with a Bonferroni adjustment. Necessary requirements for normality and sphericity were given. Partial eta² (η^2_p) was calculated for overall significances.

Borders for effect size η^2_p were set to be: η^2_p=0.01 for small, η^2_p=0.06 for medium and η^2_p=0.14 for high effect sizes. The level of significance was set at $p \leq 0.05$ for overall significance. T-tests were used as post-hoc tests to assess pair-wise comparisons. For post-hoc tests a Bonferroni adjustment was performed by reducing the level of significance to $p \leq 0.016$ (three technique variations).

3.3 Results

3.3.1 Kinematics

Figure 3.2 displays the angular patterns of the hip, knee and ankle for all three squatting techniques. Table 3.1 displays the angular magnitudes for all kinematic parameters and the p-values for statistically significant comparisons. For the knee-shifted squat the knee joint center moved 3.6 (±2.5) cm anteriorly past the vertical line of the toes, while it stayed for the standard squat 4.7 (±3.8) cm and for the declined squat 3.3 (±3.9) cm posteriorly behind the vertical line of the subject's toes. Overall significant differences were identified in the sagittal plane at the hip ($p \leq 0.001$; η^2_p=0.608) and ankle ($p \leq 0.001$; η^2_p=0.967), in the frontal plane at the knee ($p \leq 0.001$; η^2_p=0.790) and ankle ($p \leq 0.001$; η^2_p=0.494), and in the transversal plane at the hip ($p \leq 0.001$; η^2_p=0.652).

For the knee-shifted squat significantly lower hip flexion angles were identified compared to the standard squat (overall: p=0.001; specific: p<0.016: 30° eccentric phase - 30° concentric phase) and to the declined squat (overall: p=0.001; specific: p<0.016: 10°-90° eccentric phase, 50°-10° concentric phase), in additon significantly higher hip internal rotation angles were identified compared to the declined squat (overall: p≤0.001; specific: p<0.016: all knee angles). At the knee joint the knee-shifted squat showed significantly higher varus angles compared to the standard squat (overall: p≤0.001; specific: p<0.016: 30°-70° eccentric phase; 70°-10° concentric phase) and to the declined squat (overall: p≤0.001; specific: p<0.016: 30°-90° eccentric phase, 70°-10° concentric phase). For the knee-shifted squat significantly higher ankle dorsiflexion angles were observed compared to the standard squat (overall: p≤0.001; specific: p<0.016: all knee angles) and to the declined squat (overall: p≤0.001; specific: p<0.016: all knee angles). Additionally, significantly lower foot abduction angles were observed compared to the declined squat (overall: p=0.001; specific: p<0.016: all knee angles).

For the declined squat significantly higher hip flexion angles were identified compared to the knee-shifted squat (overall: p≤0.001; specific: p<0.016: 10°-90° eccentric phase, 50°-10° concentric phase) and significantly lower internal rotation angles were identified compared to the standard squat (overall: p≤0.001; specific: p<0.016: 10°-70° eccentric phase, 70-10° concentric phase). Furthermore, significantly lower varus angles compared to the standard squat (overall: p=0.005; specific: p<0.016: 30°-70° eccentric phase, 70°-10° concentric phase) and compared to the knee-shifted squat (overall: p≤0.001; specific: p<0.016: 30°-90° eccentric phase, 70°-10° concentric phase) were observed. At the beginning of the declined squat the ankle is plantar-flexed and only dorsi-flexes with increasing knee angle, while the ankle is dorsi-flexed during the entire standard and knee-shifted squat, hence significant differences occur compared to the standard squat (overall: p≤0.001; specific: p<0.016: all knee angles), and to the declined squat (overall: p≤0.001; specific: p<0.016: all knee angles).

The declined squat also shows significantly higher ankle adduction angles compared to the knee-shifted squat (overall: $p=0.001$; specific: $p<0.016$: all knee angles).

3.3.2 Kinetics

Figure 3.3 displays the joint moment patterns of the hip, knee and ankle for all three squatting techniques. Table 3.2 displays the joint moment magnitudes for all kinetic parameters and the p-values for statistically significant comparisons.

Overall significant differences were identified in the sagittal plane at the hip ($p=0.008$; $\eta^2_p=0.355$) and ankle ($p\leq0.001$; $\eta^2_p=0.737$) and in the frontal plane at the knee ($p=0.020$; $\eta^2_p=0.292$) and ankle ($p=0.012$; $\eta^2_p=0.279$).

For the knee-shifted squat significantly lower hip flexion moments (overall: $p=0.004$; specific: $p<0.016$: 30°-70° eccentric phase, 50°,30° concentric phase) and significantly higher hip abduction moments (overall: $p=0.030$; specific: $p<0.016$: 30°-70° eccentric phase) were identified compared to the declined squat. At the ankle significantly higher plantar-flexor moments were identified compared to the standard squat (overall: $p \leq 0.001$; specific: $p<0.016$: 50°-90° eccentric phase; 70°-10° concentric phase) as well as to the declined squat (overall: $p\leq0.001$; specific: $p<0.016$: 30°-70° eccentric phase; 70°-10° concentric phase). Furthermore, significantly lower abduction angles were identified at the concentric phase compared to the standard squat (overall: $p=0.037$; specific: $p<0.016$: 70°-10° concentric phase).

For the declined squat significantly higher hip flexion moments (overall: $p=0.004$; specific: $p<0.016$: 30°-70° eccentric phase, 50°,30° concentric phase) as well as significantly lower abduction moments were identified compared to the knee-shifted squat (overall: $p=0.030$; specific: $p<0.016$: 30°-70° eccentric phase). At the ankle the declined squat showed significantly lower plantar-flexor moments compared to the knee-shifted squat (overall: $p \leq 0.001$; specific: $p<0.016$: 30°-70° eccentric phase; 70°-10° concentric phase). No significant differences in lower limb joint moments were identified between the declined squat and the standard squat.

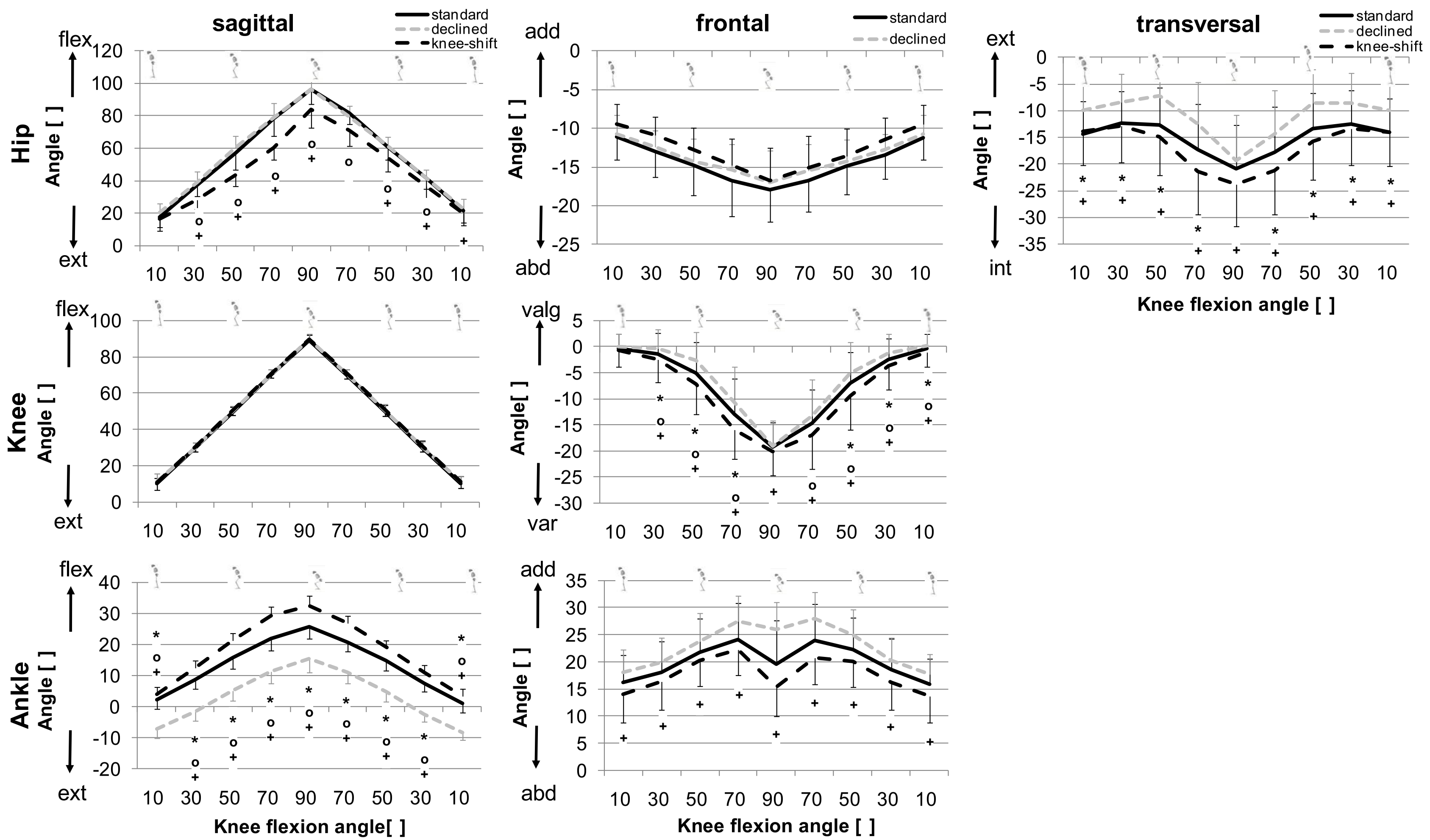


Figure 3.2: Mean angles at the hip, knee and ankle joint over the entire squatting movement. Significant difference between variations are marked as: * sig. difference between standard – declined squat; o: sig. difference between standard – knee-shifted squat; + sig. difference between declined – knee shifted squat

Table 3.1: Mean and standard deviation of the hip, knee and ankle angles over the entire squatting movement.

Angles		sagittal						frontal						transversal					
[°]	kneeangle	standard	knee-shift	declined	p(st.-shift)	p(st.-decl)	p(decl-shift)	standard	knee-shift	declined	p(st.-shift)	p(st.-decl)	p(decl-shift)	standard	knee-shift	declined	p(st.-shift)	p(st.-decl)	p(decl-shift)
Hip	overall p				0.001	1.000	0.000				0.227	0.737	0.775				0.091	0.000	0.000
sagittal:	10°	17.8 (6.2)	16.4 (7.1)	20.2 (5.6)	0.293	---	**0.010**	-11.1 (3.0)	-9.4 (2.5)	-10.8 (2.5)	---	---	---	-14.5 (6.2)	-13.8 (6.4)	-10.0 (5.5)	---	**0.000**	**0.005**
flexion (+)	30°	37.4 (7.2)	28.5 (6.2)	39.1 (6.3)	**0.000**	---	**0.000**	-13.0 (3.3)	-10.9 (2.4)	-12.4 (2.1)	---	---	---	-12.3 (6.1)	-12.8 (6.8)	-8.4 (5.2)	---	**0.000**	**0.001**
frontal:	50°	56.7 (9.9)	43.5 (7.0)	60.0 (7.6)	**0.000**	---	**0.000**	-14.8 (3.9)	-12.8 (2.9)	-14.3 (3.0)	---	---	---	-12.7 (7.0)	-15.0 (7.1)	-7.3 (5.9)	---	**0.013**	**0.000**
add (+)	70°	78.2 (10.5)	60.4 (7.4)	78.8 (9.1)	**0.000**	---	**0.000**	-16.8 (4.5)	-14.7 (3.4)	-15.4 (3.3)	---	---	---	-17.3 (8.6)	-21.4 (8.0)	-12.6 (7.9)	---	**0.000**	**0.000**
transversal:	90°	96.5 (9.1)	83.8 (11.0)	96.4 (7.2)	**0.001**	---	**0.003**	-17.9 (4.2)	-16.8 (4.3)	-17.0 (4.1)	---	---	---	-20.9 (8.2)	-23.8 (7.8)	-19.3 (8.5)	---	0.251	**0.007**
extern (+)	70°	81.6 (6.8)	71.0 (9.6)	79.5 (6.5)	**0.001**	---	0.044	-16.7 (4.1)	-15.1 (4.1)	-15.5 (3.5)	---	---	---	-17.7 (8.5)	-21.3 (8.2)	-14.4 (8.2)	---	**0.004**	**0.000**
	50°	60.8 (8)	54.2 (8.7)	61.4 (5.9)	**0.006**	---	**0.002**	-14.9 (3.6)	-13.6 (3.6)	-14.3 (2.8)	---	---	---	-13.5 (6.8)	-15.8 (7.1)	-8.5 (6.3)	---	**0.000**	**0.000**
	30°	42.0 (6.9)	36.8 (7.8)	41.7 (5.3)	**0.001**	---	**0.001**	-13.4 (3.1)	-11.5 (2.9)	-12.7 (1.9)	---	---	---	-12.5 (6.2)	-13.4 (6.8)	-8.6 (5.7)	---	**0.000**	**0.001**
	10°	21.6 (7.5)	19.8 (7.2)	22.7 (5.9)	0.167	---	**0.005**	-11.2 (2.8)	-9.5 (2.5)	-10.8 (2.5)	---	---	---	-14.0 (6.3)	-14.1 (6.4)	-9.9 (5.7)	---	**0.000**	**0.003**
	overall p				0.377	0.836	1.000				0.000	0.005	0.000						
	10°	10.0 (3.3)	10.8 (4.2)	11.3 (4.6)	---	---	---	-0.3 (2.7)	-0.7 (3.1)	-0.1 (2.6)	0.209	0.371	0.086						
Knee	30°	29.6 (3.2)	30.1 (2.4)	29.8 (3.1)	---	---	---	-1.5 (4.0)	-2.5 (4.3)	-0.4 (3.8)	**0.001**	**0.002**	**0.000**						
sagittal:	50°	49.4 (3)	50.2 (2.2)	50.1 (2.9)	---	---	---	-5.0 (5.9)	-7.2 (5.8)	-2.7 (5.4)	**0.000**	**0.001**	**0.000**						
flexion (+)	70°	69.9 (3.0)	70.6 (1.8)	70.1 (2.7)	---	---	---	-13.0 (6.9)	-16.0 (5.5)	-10.8 (6.9)	**0.001**	**0.001**	**0.000**						
frontal:	90°	89.1 (3.1)	89.9 (1.9)	89.7 (2.9)	---	---	---	-19.3 (4.7)	-20.2 (4.5)	-19.1 (4.9)	0.032	0.726	**0.002**						
valgus (+)	70°	69.9 (3.3)	70.8 (2.7)	70.3 (2.8)	---	---	---	-14.8 (6.5)	-17.0 (6.4)	-13.4 (7.1)	**0.000**	0.017	**0.000**						
	50°	49.7 (3.5)	50.6 (3.2)	50.6 (3.3)	---	---	---	-7.1 (6.0)	-9.5 (6.4)	-5.0 (6.0)	**0.002**	**0.003**	**0.000**						
	30°	30.0 (4.0)	30.7 (3.1)	30.1 (3.4)	---	---	---	-2.4 (4.1)	-3.7 (4.5)	-1.2 (3.7)	**0.001**	**0.001**	**0.000**						
	10°	10.4 (3.6)	11.1 (3.5)	11.0 (3.3)	---	---	---	-0.3 (2.7)	-1.0 (2.8)	0.2 (2.6)	**0.015**	0.184	**0.004**						
	overall p				0.000	0.000	0.000				0.129	0.072	0.000						
	10°	2.1 (3.0)	3.9 (2.4)	-7.3 (3.0)	**0.011**	**0.000**	**0.000**	16.2 (5.0)	14.0 (5.2)	18.1 (4.1)	---	---	**0.000**						
Ankle	30°	8.6 (2.8)	12.5 (2.3)	-1.6 (3.0)	**0.000**	**0.000**	**0.000**	18.0 (5.7)	16.4 (5.3)	19.9 (4.5)	---	---	**0.002**						
sagittal:	50°	15.7 (3.7)	21.1 (2.5)	5.1 (3.3)	**0.000**	**0.000**	**0.000**	21.7 (6.2)	20.3 (4.8)	23.8 (5.1)	---	---	**0.001**						
flexion (+)	70°	21.8 (3.9)	29.3 (2.7)	11.3 (3.9)	**0.000**	**0.000**	**0.000**	24.1 (6.7)	22.2 (4.6)	27.5 (4.7)	---	---	**0.000**						
frontal:	90°	25.6 (3.7)	32.4 (3.1)	15.4 (4.4)	**0.000**	**0.000**	**0.000**	19.5 (8.1)	15.3 (5.4)	25.9 (5.1)	---	---	**0.000**						
add (+)	70°	20.7 (2.9)	26.7 (2.6)	11.1 (3.8)	**0.000**	**0.000**	**0.000**	24.0 (6.6)	20.7 (4.8)	27.9 (5.0)	---	---	**0.000**						
	50°	14.8 (3.2)	19.2 (1.9)	4.9 (3.2)	**0.000**	**0.000**	**0.000**	22.2 (6.0)	20.0 (4.7)	24.9 (4.8)	---	---	**0.000**						
	30°	7.5 (2.8)	11.1 (2.2)	-2.4 (2.6)	**0.000**	**0.000**	**0.000**	18.5 (5.7)	16.1 (5.0)	20.2 (4.3)	---	---	**0.000**						
	10°	1.1 (3.1)	3.4 (2.3)	-8.4 (2.5)	**0.004**	**0.000**	**0.000**	15.9 (4.7)	13.6 (4.9)	17.7 (3.7)	---	---	**0.000**						

* p-level at: overall: 0.05; between degrees: 0.016 (Bonferroni-adjustment)

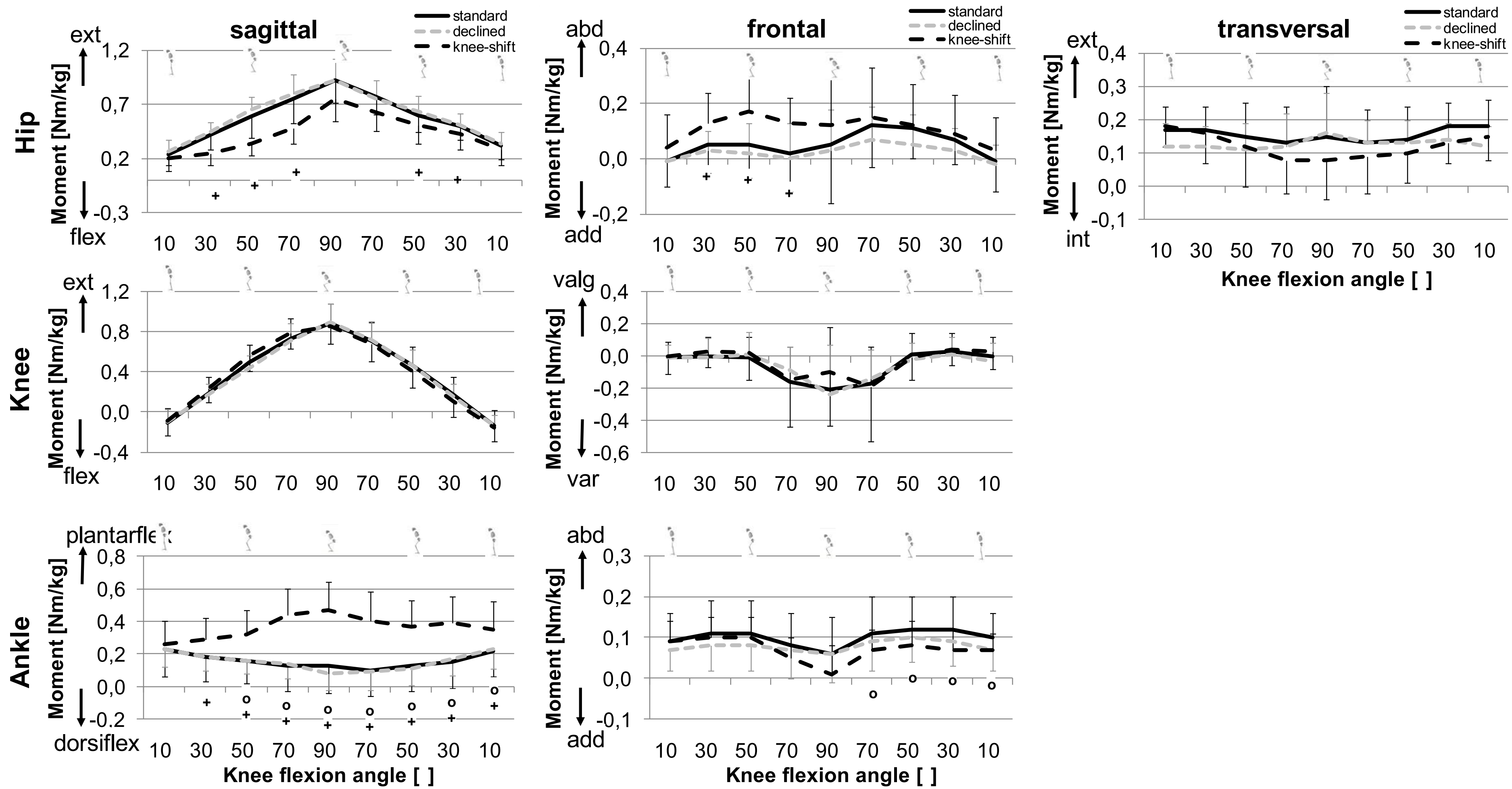


Figure 3.3: Mean internal moments at the hip, knee and ankle joint over the entire squatting movement. Significant difference between variations are marked as: * sig. difference between standard – declined squat; o: sig. difference between standard – knee-shifted squat; + sig. difference between declined – knee shifted squat

Table 3.2: Mean and standard deviation of the hip, knee and ankle internal moments over the entire squatting movement.

Moments		sagittal						frontal						transversal					
[Nm/kg]	kneeangle	standard	knee-shift	declined	p(stand-shift)	p(stand-decl)	p(decl-shift)	standard	knee-shift	declined	p(stand-shift)	p(stand-decl)	p(decl-shift)	standard	knee-shift	declined	p(stand-shift)	p(stand-decl)	p(decl-shift)
Hip	overall p				0.096	1.000	0.004				0.252	0.188	0.030				0.504	0.410	1.000
sagittal:	10°	0.24 (0.1)	0.20 (0.11)	0.27 (0.10)	---	---	0.027	-0.01 (0.09)	0.04 (0.12)	-0.01 (0.05)	---	---	0.047	0.17 (0.07)	0.18 (0.06)	0.12 (0.07)	---	---	---
ext (+)	30°	0.42 (0.14)	0.25 (0.11)	0.45 (0.09)	---	---	**0.000**	0.05 (0.1)	0.13 (0.11)	0.03 (0.07)	---	---	**0.002**	0.17 (0.07)	0.16 (0.09)	0.12 (0.06)	---	---	---
frontal:	50°	0.59 (0.20)	0.34 (0.11)	0.65 (0.12)	---	---	**0.000**	0.05 (0.12)	0.17 (0.12)	0.02 (0.11)	---	---	**0.005**	0.15 (0.10)	0.12 (0.12)	0.11 (0.08)	---	---	---
adb (+)	70°	0.75 (0.22)	0.48 (0.14)	0.80 (0.18)	---	---	**0.000**	0.02 (0.14)	0.13 (0.09)	0.00 (0.13)	---	---	**0.004**	0.13 (0.11)	0.08 (0.10)	0.12 (0.10)	---	---	---
transversal:	90°	0.92 (0.20)	0.75 (0.20)	0.92 (0.20)	---	---	0.032	0.05 (0.21)	0.12 (0.18)	0.03 (0.15)	---	---	0.040	0.15 (0.15)	0.08 (0.12)	0.16 (0.12)	---	---	---
extern (+)	70°	0.77 (0.16)	0.63 (0.17)	0.75 (0.17)	---	---	0.039	0.12 (0.15)	0.15 (0.18)	0.07 (0.12)	---	---	0.048	0.13 (0.10)	0.09 (0.11)	0.13 (0.07)	---	---	---
	50°	0.60 (0.15)	0.51 (0.17)	0.64 (0.14)	---	---	**0.016**	0.11 (0.11)	0.12 (0.15)	0.05 (0.11)	---	---	0.071	0.14 (0.10)	0.10 (0.09)	0.13 (0.07)	---	---	---
	30°	0.50 (0.13)	0.43 (0.15)	0.51 (0.11)	---	---	**0.013**	0.07 (0.09)	0.09 (0.14)	0.03 (0.08)	---	---	0.044	0.18 (0.07)	0.13 (0.06)	0.14 (0.05)	---	---	---
	10°	0.32 (0.13)	0.28 (0.14)	0.34 (0.11)	---	---	0.038	-0.01 (0.11)	0.03 (0.12)	-0.02 (0.07)	---	---	0.058	0.18 (0.08)	0.15 (0.07)	0.12 (0.06)	---	---	---
	overall p				1.000	1.000	1.000				1.000	1.000	1.000						
	10°	-0.11 (0.15)	-0.09 (0.14)	-0.10 (0.13)	---	---	---	-0.01 (0.10)	0.00 (0.11)	-0.01 (0.08)	---	---	---						
Knee	30°	0.18 (0.17)	0.23 (0.13)	0.17 (0.11)	---	---	---	0.00 (0.12)	0.03 (0.10)	-0.01 (0.12)	---	---	---						
sagittal:	50°	0.49 (0.18)	0.56 (0.15)	0.44 (0.12)	---	---	---	-0.01 (0.13)	0.02 (0.17)	0.01 (0.14)	---	---	---						
ext (+)	70°	0.73 (0.20)	0.79 (0.16)	0.71 (0.17)	---	---	---	-0.16 (0.22)	-0.15 (0.29)	-0.09 (0.15)	---	---	---						
frontal:	90°	0.88 (0.20)	0.85 (0.17)	0.89 (0.19)	---	---	---	-0.21 (0.39)	-0.10 (0.33)	-0.24 (0.31)	---	---	---						
adb (+)	70°	0.71 (0.18)	0.68 (0.18)	0.72 (0.18)	---	---	---	-0.17 (0.23)	-0.19 (0.34)	-0.14 (0.18)	---	---	---						
	50°	0.47 (0.18)	0.41 (0.17)	0.47 (0.15)	---	---	---	0.01 (0.13)	-0.01 (0.14)	-0.02 (0.10)	---	---	---						
	30°	0.17 (0.18)	0.11 (0.16)	0.15 (0.13)	---	---	---	0.03 (0.11)	0.04 (0.10)	0.01 (0.11)	---	---	---						
	10°	-0.14 (0.16)	-0.16 (0.13)	-0.14 (0.11)	---	---	---	0.00 (0.12)	0.03 (0.11)	-0.03 (0.11)	---	---	---						
	overall p				0.000	1.000	0.000				0.037	0.113	0.930						
	10°	0.23 (0.17)	0.26 (0.14)	0.23 (0.11)	0.371	---	0.391	0.09 (0.07)	0.09 (0.05)	0.07 (0.05)	0.948	---	---						
Ankle	30°	0.18 (0.15)	0.29 (0.13)	0.18 (0.08)	0.019	---	**0.006**	0.11 (0.08)	0.10 (0.05)	0.08 (0.06)	0.253	---	---						
sagittal:	50°	0.16 (0.14)	0.32 (0.15)	0.16 (0.08)	**0.002**	---	**0.001**	0.11 (0.08)	0.10 (0.05)	0.08 (0.06)	0.321	---	---						
ext (+)	70°	0.13 (0.16)	0.44 (0.16)	0.14 (0.09)	**0.000**	---	**0.000**	0.08 (0.08)	0.05 (0.05)	0.07 (0.07)	0.113	---	---						
frontal:	90°	0.13 (0.17)	0.47 (0.17)	0.08 (0.10)	**0.000**	---	**0.000**	0.06 (0.09)	0.01 (0.07)	0.06 (0.07)	0.034	---	---						
adb (+)	70°	0.10 (0.16)	0.40 (0.18)	0.09 (0.11)	**0.000**	---	**0.000**	0.11 (0.09)	0.07 (0.05)	0.09 (0.07)	**0.015**	---	---						
	50°	0.13 (0.16)	0.37 (0.16)	0.11 (0.10)	**0.000**	---	**0.000**	0.12 (0.08)	0.08 (0.06)	0.10 (0.06)	**0.001**	---	---						
	30°	0.15 (0.16)	0.39 (0.16)	0.17 (0.10)	**0.000**	---	**0.000**	0.12 (0.08)	0.07 (0.05)	0.09 (0.06)	**0.000**	---	---						
	10°	0.22 (0.16)	0.35 (0.17)	0.23 (0.12)	**0.003**	---	**0.000**	0.10 (0.06)	0.07 (0.04)	0.07 (0.05)	**0.005**	---	---						

* p-level at: overall: 0.05; between knee-angles: 0.016 (Bonferroni-adjustment)

3.4 Discussion

The kinematic changes of squatting on a declined surface indicate similar mechanisms on the joint loading as for the knee being pushed anteriorly past the virtual vertical line of the toes. Therefore, the aim of this study was to analyze joint moments of three squatting variations representing a standard squat, a squat with the knee being shifted anteriorly over the virtual vertical line of the toe ('knee-shifted squat') and a squat with elevated heels.

At the knee-shifted squat the knee moved 3.6 (±2.5) cm anteriorly the vertical line of the toes. At the standard squat the knee stayed 4.7 (±3.8) cm posterior to the toes of the subjects, at the declined squat 3.3 (±3.9) cm. This change also caused significantly smaller ankle dorsi-flexion angles, significantly smaller hip flexion angles and significantly higher valgus angles at the knee compared to the standard squat and declined squat. In addition, the hip was significantly more internally rotated than at the declined squat. These kinematic alterations resulted in an anterior shift of 5.3 (±2.9) cm of the center of pressure and respectively the force vector at the foot as base of support. Therefore, lever arms to the knee and hip joint centers are comparable to those of the standard and declined squat and lead to similar joint moments at these two joints. A different situation occurs at the ankle joint, where a by 360% significantly higher plantar-flexor moment was identified, due to a longer lever arm between the ground reaction force vector and the joint center (Figure 3.4). In addition, a decrease of approximately 22% in ankle abduction moment for the concentric phase was observed. Due to the low magnitudes the changes in frontal plane might be negligible though.

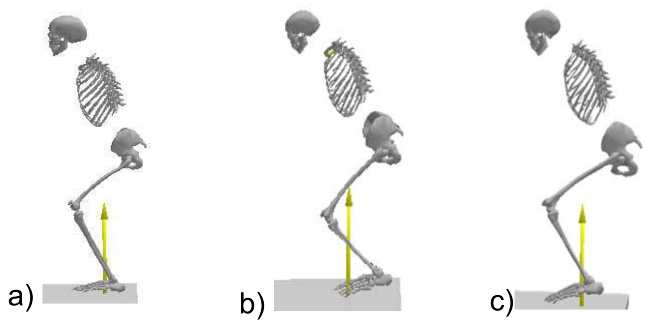


Figure 3.4: a) standard squat, b) knee-shifted squat and c) declined squat at 90° knee flexion angle.

These findings are in contrast to previous studies, which report a significant increase in sagittal knee joint moment by 30% (Fry et al., 2003) and by 38%, respectively, (Hirata & Duarte, 2007) and a significant decrease in hip joint moment by 7% (Fry et al., 2003) for a knee-shifted squat. Parallel squat depths at the studies of Fry et al. (2003) and Hirata & Duarte (2007) were not presented as a function of knee angle and both studies reported higher knee flexion angles for the knee-shifted squat. The respective joint moments were then analyzed at the instant of maximum knee flexion (Fry et al., 2003) and maximum patellar compressive force (Hirata & Duarte, 2007). Since the joint moments for the squat are directly related with knee flexion angle (Escamilla, 2001), it is not clear, whether the reported higher knee and hip joint moments by Fry et al. (2003) and Hirata & Duarte (2007) are caused by the knee-shifted technique or by the more flexed knee. In this study the squatting depth was controlled by knee angle for all three technique variations, and therefore the reported moments are due to technique. However, this might not reflect the normal training situation. It is likely that athletes, executing a knee-shifted squat, have a higher range of motion and therefore a higher knee and hip joint moment. The increase in sagittal ankle moment is considerable though and has not been addressed in previous studies. It clearly indicates higher muscle work that has to be done by the plantar flexor muscles. Hence, this technique can be considered as an adequate variation if the aim of the weight lifting training is to target this muscle group additionally. In order to limit the increase of hip and knee moments, squatting depth should be restricted to a 90° knee angle though.

Squatting on a declined surface compared to the standard squat leads to significantly lower internal rotation at the hip, significantly lower valgus angles at the knee, and significantly lower plantar- and dorsi-flexion angles, but significantly higher eversion angles at the ankle. These angular changes only lead to subtle changes in joint moments, hence, no differences to the standard squat can be identified. The more erect trunk posture reported by Fortenbaugh et al. (2010) for squatting with weightlifting shoes was not realized by this study's subject group. A common error for novice lifters performing a barbell squat is an excessive forward lean of the trunk (Chandler & Stone, 1991; Dunn et al., 1984). It is likely that this happened to our lifters, being novice lifters, in an attempt to restrict anterior movement of the knees for the standard squat as well as for the declined squat, while Fortenbaugh et al. (2010) analyzed resistance training experienced lifters.

For squatting on a declined surface compared to the knee-shifted squat the same mechanism as described for the differences between standard and knee-shifted squat were identified for the ankle plantar-flexor moments. Also, significantly higher hip extension moments during the eccentric phase (30°-70°) and the concentric phase (50°, 30°) of the squat and significantly lower abduction moments during the eccentric phase of the squat (30°- 70°) were identified. At all other parameters no significant differences in joint moments were observed. The higher hip extension moment might be due to the fact that the subjects leaned more forward in order to prevent falling back at the declined squat, as already previously addressed. The forward lean caused an anterior movement of the center of pressure and thereby a longer lever arm to the hip joint center. This might not be relevant when compared to the standard squat, but in comparison to the knee-shifted squat significantly higher hip joint moments in sagittal and frontal plane can be identified for the declined squat.

In the rehabilitation of patella tendinopathy the clinically beneficial effect of a higher load on the knee - and consequently the patellar tendon - when squatting on a declined surface (Purdam et al., 2004; Young et al., 2005; Kongsgaard et al., 2006; Frohm et al., 2007; Richards et al., 2008) cannot be observed in this study. Mostly a declined surface of 25° was used in former studies (Purdam et al., 2004; Young et al., 2005; Kongsgaard et al., 2006; Frohm et al., 2007), while an inclination angle of ~10° was used in this study. Limited research is established on the magnitude of the decline angle (Richards et al., 2008). Based on our findings an inclination of 10° seems to be too low to lead to an alteration in knee joint moments.

One methodological limitation of the inverse dynamics approach is that co-contraction of agonistic muscles are not taken into consideration. This fact may limit the ability to infer specific structural joint loading (such as e.g. changes in ligament forces, compressive forces and shear forces) from external joint moments and consequently means that the effects of different squatting techniques on specific joint structures can only be speculated about. Calculating the net joint moments is the first step in understanding the forces in the respective joints though. By producing an external knee flexor moment a quadriceps extensor moment is also needed for the athlete to keep the equilibrium. Hence, a quadriceps force will be needed to generate this moment, which further has effects on single sub-components of the knee such as tibiofemoral and patellofemoral compression forces or the forces on the anterior and posterior cruciate ligament (Escamilla, 2001). Further, it has been reported that a greater forward tilted trunk position decreases potential ACL strain, in part due to greater hamstrings activity and less quadriceps activity (Ohkoshi et al., 1991; Escamilla et al., 2001). Hence, squatting with greater forward trunk tilt as it was identified for the knee-shifted squat may be appropriate for those whose goal is to minimize ACL stress. This needs to be proven in further studies though.

It also should be noted that this study examined novice lifters squatting with a relatively low barbell mass of 20 kg. There is the assumption that greater absolute loads may result in altered anterior torso tilt, especially for the restricted squat, and may alter the results.

3.5 Conclusion

Squatting and technique variations are considered from different aspects. One aspect is a reduction of joint moments in order to produce less mechanical loading on specific structures to prevent overloading and injury. Another aspect is to increase joint loading to set positive impulses to the musculoskeletal system in order to strengthen or to help regenerate as desired from specific injuries, e.g. patella tendinopathy. The knee-shifted squat shows increased ankle plantar-flexor moments resulting in an increase of work done by the plantar-flexor muscles. Hence, the knee-shifted squat might be a good squat variation to target the plantar-flexor muscles without increasing the joint moments at the knee and hip. Squatting on a 3 cm shim leads to subtle changes, which are only relevant when compared with the knee-shifted squat, and hence no benefit for this variation was detected.

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4 Study III: Stair walking adaptations in obese children: Spatio - temporal, kinematic and kinetic differences to normal-weight children

4.1 Introduction

The prevalence of adult and childhood obesity continues to increase in most countries of the world (Flodmark et al., 2004). In Germany, the prevalence of pediatric obesity increased by 50% in the last decade. One in six children is affected by overweight or obesity (Kurth & Schaffrath, 2007), and four out of five obese teenagers remain obese in adulthood (Flodmark et al., 2004). Obesity may cause orthopedic problems not only during childhood but may also have long-term implications for musculoskeletal health during adolescence and into adulthood.

Mechanical factors such as joint geometry, bone mineral content and body weight have been identified as important factors for the development and progression of foot deformities (Garcia-Rodriguez et al., 1999; Shultz et al., 2009a), varus/valgus angular deformities of the knee (Dietz et al., 1982; Henderson, 1992; Shultz et al., 2009a), slipped capital femoral epiphysis (Shultz et al., 2009a) and have long term implications for developing osteoarthritis (e.g. Oliveria et al., 1999; Sharma et al., 2000; Miyazaki et al., 2002; Powell et al., 2005;). However, only few studies investigated the interrelation of these factors during dynamic activities in children.

The few studies on kinematic and kinetic indices of obese subjects' movement reported differences in movement strategy between normal and obese subjects for rising from a chair (Sibella et al., 2003; Riddiford-Harland et al., 2006) and for level-gait (Hills & Parker, 1991; Spyropoulos et al., 1991; McGraw et al., 2000; DeVita & Hortobágyi, 2003; Gushue et al., 2005; Nantel et al., 2006; Browning & Kram, 2007; Morrison et al., 2008; Shultz et al., 2009b). Additional trunk movement in the preparation phase of rising from a seated position indicated that obese subjects experience a greater difficulty in performing this movement (Riddiford-Harland et al., 2006). To accomplish the sit-to-stand task, obese subjects moved with a reduced trunk flexion and repositioned their feet backwards from the initial position resulting in smaller hip moments and greater knee joint moments (Sibella et al., 2003)

Compensatory movement changes for gait parameters in obese children and adults include slower walking velocities, longer double support phases, wider stance widths (Spyropoulos et al., 1991), higher degree of asymmetry (Hills & Parker, 1991) and all in all a more "tentative ambulation" (Hills & Parker, 1991; McGraw et al., 2000). In addition, obese adults have shorter step lengths, smaller knee range of motion and a more erect walking pattern (Spyropoulos et al., 1991; McGraw et al., 2000; DeVita & Hortobágyi, 2003). However, only few studies investigated the influence of obesity on ambulatory kinetics (DeVita & Hortobágyi, 2003; Nantel et al., 2006; Gushue et al., 2005; Browning & Kram, 2007; Shultz et al., 2009b). While higher absolute joint moments for obese children were identified at the hip, knee and ankle in the sagittal (DeVita & Hortobágyi, 2003; Gushue et al., 2005; Browning & Kram, 2007; Shultz et al., 2009b), frontal (Gushue et al., 2005; Shultz et al., 2009b) and transverse (Shultz et al., 2009b) planes, there are inconsistent findings when the moments are scaled to bodyweight. Modification patterns for obese children have been reported for the hip to reduce work done by the hip flexors (Nantel et al., 2006), the knee

(DeVita & Hortobágyi, 2003) with lower knee extensor moments, and the ankle with either higher peak dorsiflexor moments (Shultz et al., 2009), higher plantarflexor moments (DeVita & Hortobágyi, 2003) or lower plantarflexor moments (Browning & Kram, 2007; Gushue et al., 2005). Although these results indicate that obese children reorganize their walking pattern, the current literature does not support clear conclusions regarding systematic effects of obesity on gait pattern and joint moments. Similarly, while some information is available on functional joint loading during level walking and rising from a chair in obese persons, to date no information is available on joint moments during other daily activities such as, for instance, stair walking. Stair walking is of special interest because greater ground reaction forces and knee moments are required for stair walking tasks compared to those required for level walking (McFadyen & Winter, 1988; Nadeau et al., 2003). In addition, body mass index has been associated with greater difficulty of descending and ascending stairs (Stickles et al., 2001). Therefore, the purpose of this study was to test the hypothesis that during stair walking lower extremity joint moments normalized to body weight in obese children are greater than those in normal-weight children.

4.2 Methods

4.2.1 Subjects

Eighteen obese children (10.5 ± 1.5 yrs; 148 ± 10 cm; 56.6 ± 8.39 kg) and 17 normal-weight children (mean ± standard deviation; age: 10.4 ± 1.3 yrs; height: 143 ± 9 cm; mass: 36.7 ± 7.5 kg) were recruited for this study. Obesity was defined as having a body mass index (BMI) at or above the 95th percentile of BMI for age, and normal-weight was defined as having a BMI between the 15th percentile and the 85th percentile of BMI for age (Ogden et al., 2002). Subjects were excluded if they had experienced any lower extremity injury during the past six months. The study was approved by the institutional review board. One parent of each subject signed an informed consent form prior to participation.

4.2.2 Equipment (Staircase set-up, systems)

The experimental staircase consisted of six steps (Figure 4.1). The step dimensions were 17 cm (riser) and 28 cm (tread) with a stair slope of 31°. No handrail was used. Kinematic and kinetic recordings were collected simultaneously by a ten camera, three-dimensional motion analysis system (VICON, MX camera system, Oxford Metrics Ltd, UK)[a] and two force platforms (AMTI, model BP600900, Advanced Mechanical Technology, Watertown, MA)[b] positioned as the 3rd and 4th stair step. Kinematic data were sampled at 200 Hz, and ground reaction forces were collected with 1000 Hz.

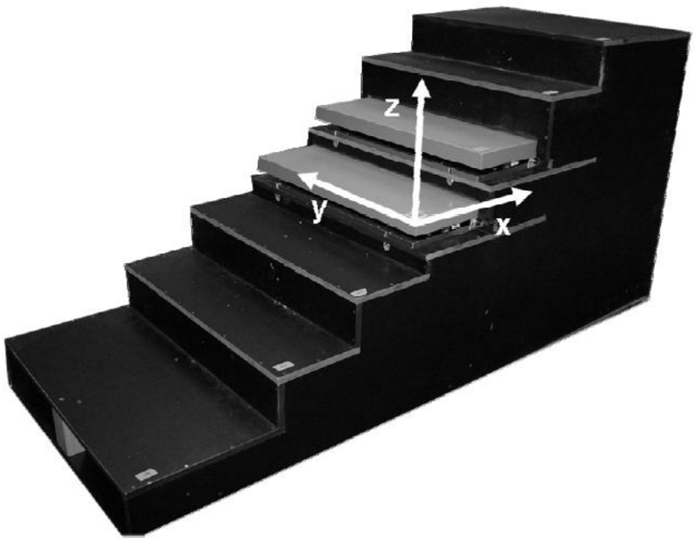

Figure 4.1: Staircase set up. Independent force plates were imbedded in steps 3 and 4 of the six-step staircase

4.2.3 Subject preparation and procedure

All subjects walked barefoot and wore swimsuits to allow an unobstructed attachment of reflective markers to the skin. Reflective markers were placed according to the Vicon Plug In Gait (PIG) full-body marker set[a], and anthropometric measurements were taken. All subjects were asked to ascend and descend the stairs, placing only one foot on each step with a cadence of 110 steps per minute. This cadence was identified as comfortable stair walking speed in pretests and was given by a metronome. No instructions for arm position were given. For each subject, testing consisted of one static standing trial and as many ascending and descending trials as needed until three valid trials for each condition could be recorded. A trial was considered valid when the given cadence was achieved and no visible alterations in the stride characteristics were detected. Sufficiently long rest periods were given between trials to avoid fatigue.

4.2.4 Data analysis

Three-dimensional coordinates of the reflective markers were collected during the locomotion task. All trajectories were filtered using a generalized cross-validated spline technique as reporterd by Woltring (Woltring, 1986). Relative angles were calculated using the analysis package Vicon Nexus[a] PIG. According to the Vicon PIG[a] definitions, the local x-, y- and z-axes corresponded to flexion-extension, abduction-adduction and rotation at the hip and knee, respectively, and dorsiflexion-plantarflexion, eversion-inversion and internal-external rotation at the ankle, respectively.

Prior to parameter calculation, ground reaction forces (kinetic data) were filtered with a 4th-order 7 Hz Butterworth filter to eliminate the slight oscillation of the staircase. The relatively low cut-off frequency also eliminated impact forces, and hence only active forces were analyzed. An inverse dynamics approach of the PIG model[a] was used to calculate net moments at the ankle, knee and hip joints, respectively. Because no anthropometric data set for obese children is available, the standard values for adults (Dempster et al., 1959) implemented in the Vicon PIG model[a] were used for calculation. Even though these values might differ from those of children, it has been shown that during stance phase the effects of these differences are negligible (Ganley & Powers, 2004). Ground reaction forces and net joint moments were normalized to subject body mass and were expressed as internal moments.

The gait cycle was defined as initial foot contact on the step with the embedded platform (3rd or 4th step). The gait cycle ended with the subsequent foot contact of the same foot. All gait events were expressed in percentage of the gait cycle (100%). Ensemble averages of the three trials were calculated for angular displacements and moments at each percent of the gait cycle. Key variables included into the statistical analyses were: time of double and single support; percentage of foot-off during gait cycle; step width; maximum and minimum value for thorax, pelvis, hip, knee and ankle angles in the sagittal and frontal planes, respectively; maximum ground reaction forces in the anterior-posterior, medio-lateral and vertical directions, respectively; and maximum hip, knee and ankle moments in the sagittal and frontal planes, respectively.

4.2.5 Statistical analysis

Gait symmetry was tested using t-tests for paired samples. Variables showing no significant differences were averaged across the sides. Variables showing significant differences between right and left leg are presented in Table 4.1. For these variables further analysis consisted only of the right side, because the right foot was always placed on the 3rd step and therefore was always the 3rd consecutive foot-plant for both stair ascending and descending. Significant differences between obese and normal-weight subjects were detected using a MANOVA with parameters grouped for spatiotemporal parameters and for each joint for kinematic and kinetic parameters. The mean deviation in inter-marker distance was calculated and compared between the two groups using independent t-tests. The significance level was set a priori to 5%.

Table 4.1: Asymmetrical parameters identified by dependent t-test ($p<0.05$) between right and left sides for all subjects.

Spatio-temporal parameters		*Angles*		*GRF & Moments*	
ascent	*descent*	*ascent*	*descent*	*ascent*	*descent*
Foot off	—	Maximum pelvis anterior tilt	Maximum knee varus	Maximum anterior-posterior GRF	Maximum anterior-posterior GRF
—	—	Maximum ankle plantarflexion	Maximum ankle pronation	---	Maximum ankle adduction moment

4.3 Results

4.3.1 Inter-marker distance

While greater skin movement at the hip was observed for the obese group (distance left spina iliaca anterior superior to spina iliaca posterior superior for ascending and distance left to right spina iliaca anterior superior for descending), the inter-marker displacements were still within an acceptable range (Table 4.2). High inter-marker displacements were observed at the hip for both groups (21.5–27.6 mm for ascending and 12.8–13.5 mm for descending, respectively), and the obese group did not show significantly higher artifacts.

Table 4.2: Mean deviations of inter-marker distances during stair ascent and descent. LASI: left spina iliaca anterior superior, RASI: right spina iliaca anterior superior, LPSI: left spina iliaca posterior superior, LKNE: left lateral epicondyle of the knee, RKNE: right lateral epicondyle of the knee, LANK: left malleolus lateralis, RANK: right malleolus lateralis

	ascent			descent		
deviation of distance [mm]	obese	normal-weight	p	obese	normal-weight	p
LASI-RASI	3,5 (2,3)	1,9 (0,7)	.020*	2,0 (0,8)	1,5 (0,5)	.055
LASI-LPSI	2,6 (0,6)	1,4 (0,5)	.000*	1,9 (0,4)	1,2 (0,4)	.000*
LASI-LNKE	24,1 (6,2)	27,6 (4,9)	.102	13,7 (3,8)	13,5 (4,7)	.902
RASI-RNKE	21,5 (4,7)	25,4 (3,7)	.018*	12,8 (2,5)	13,7 (3,1)	.391
LKNE-LANK	4,9 (1,7)	5,7 (2,5)	.343	5,1 (1,3)	5,8 (2,4)	.333
RKNE-RANK	4,2 (1,5)	5,7 (3,0)	.112	4,4 (1,0)	5,7 (2,9)	.129

4.3.2 Spatio-temporal gait cycle parameters

Spatio-temporal gait cycle parameters for stair ascent and descent for both groups are presented in Table 4.3. The MANOVA revealed an overall difference in spatio-temporal gait parameters between obese and normal-weight subject (p=0.043). Obese children spent less time in single support during stair ascent (p=0.010), and more time in double support (p=0.014) with a delayed foot-off (p=0.008) during stair descent compared to normal-weight children. Interestingly, obese children did not have broader step widths compared to normal-weight children.

4.3.3 Joint angles

Table 4.3 summarizes peak angles observed at the thorax, pelvis, hip, knee and ankle joint, respectively, and foot progression angle during stair ascent and descent for both groups. The MANOVA revealed overall significances between groups in pelvis and knee angles (p=0.035 and p=0.003, respectively) during stair ascent and in hip angles during stair descent (p=0.038). While ascending stairs, obese children walked with a slightly more anteriorly tilted pelvis (+3.9°; p=0.041) and with the knee in a more pronounced valgus position (+6.2°; p=0.005) than normal-weight children.

4.3.4 Ground reaction forces and moments

Normalized ground reaction forces were similar for obese and normal-weight children during stair ascent and descent. Mean peak ground reaction forces in anterior-posterior, medio-lateral and vertical direction and peak moments at the hip, knee and ankle in the sagittal and frontal planes are given in Table 4.3.

Mean (sd) sagittal and frontal plane moments at the hip, knee and ankle throughout the gait cycle of stair ascent and descent are illustrated in Figure 4.2 (ascent) and Figure 4.3 (descent). For stair ascent, overall differences were found at the hip (p=0.006) and the knee (p=0.030). Obese subjects had a 23% higher hip abduction moment (p=0.001) and a 22% higher knee extension moment (p= 0.008). All other moments during stair ascent were similar for both groups (Table 4.3).

For stair descent, the MANOVA revealed overall differences in hip moments (p=0.027) and in knee moments (p=0.030). Obese children shifted from hip extension moment into hip flexion moment at 10% of stance, while in normal-weight children this shift occurred at approximately 50% of stance. Hence, obese children had smaller hip extension moments (-53%; p=0.031), greater hip flexion moments (+26%; p=0.016) and greater knee extension moments (15%; p=0.008) compared to normal-weight children. No statistical difference in any other joint moment was observed (Table 4.3).

Table 4.3: Mean (sd) stride characteristics, maximum (sd) and minimum (sd) parameters for angles, ground reaction forces and joint moments in sagittal and frontal plane for obese and normal-weight subjects while ascending and descending stairs. Angle and moment definitions according to Vicon Plug In Gait[a].

	ascent			descent		
	obese	normal-weight	p	obese	normal-weight	p
Spatio-temporal parameters						
Double support [s]	0.26 (0.03)	0.25 (0.02)	.611	0.27 (0.03)	0.24 (0.04)	.014*
Single support [s]	0.39 (0.02)	0.42 (0.03)	.010*	0.39 (0.03)	0.41 (0.03)	.127
Foot off [%]	61.7 (1.43)	61.3 (1.32)	.184	63.3 (1.88)	61.2 (2.29)	.008*
Step width [m]	0.13 (0.05)	0.10 (0.05)	.112	0.15 (0.05)	0.13 (0.05)	.215
Angles [°]						
Thorax ant. tilt max	13.7 (4.0)	14.9 (6.0)	.499	2.8 (4.3)	2.7 (4.0)	.354
Thorax post. tilt max	9.7 (4.0)	10.7 (5.0)	.508	-1.76 (4.3)	-2.8 (3.8)	.913
Thorax right tilt	4.0 (1.6)	3.9 (2.3)	.984	2.2 (1.2)	1.8 (0.7)	.797
Thorax left tilt	-4.2 (1.6)	-3.7 (2.2)	528	-1.7 (0.9)	-1.5 (0.6)	.919
Pelvis ant. tilt max	24.5 (6.2)	21.0 (3.1)	.052	13.6 (5.3)	10.7 (3.5)	.075
Pelvis ant. tilt min	20.1 (6.3)	16.2 (3.2)	.041*	8.6 (5.9)	5.0 (4.4)	.050
Pelvis obl. up max	8.9 (2.6)	8.5 (1.4)	.693	4.9 (2.1)	4.9 (1.8)	.934
Pelvis obl. down max	-8.8 (2.4)	-8.1 (1.6)	.353	-5.4 (2.2)	-5.6 (2.0)	.812
Hip flexion max	76.0 (7.7)	75.2 (3.7)	.912	48.7 (7.6)	47.8 (7.8)	.591
Hip flexion min	17.4 (8.6)	13.1 (4.4)	.065	21.5 (9.0)	16.5 (6.9)	.078
Hip adduction max	8.0 (3.3)	8.3 (2.5)	.479	5.6 (3.8)	7.1 (3.7)	.278
Hip abduction max	-9.2 (3.6)	-9.0 (2.1)	.947	-9.1 (2.4)	-7.9 (2.1)	.140
Knee flexion max	102.1 (8.0)	104.7 (6.3)	.106	90.2 (8.6)	94.4 (9.0)	.206
Knee flexion min	15.8 (4.4)	15.0 (4.4)	.463	16.6 (3.3)	15.1 (5.5)	.140
Knee varus max	11.8 (4.6)	8.7 (3.7)	.063	7.1 (5.3)	5.8 (4.2)	.775
Knee valgus max	-12.9 (5.4)	-6.7 (3.3)	.005*	-9.4 (4.2)	-7.8 (4.0)	.222
Ankle dorsiflex. max	25.8 (5.4)	24.9 (3.0)	.873	38.4 (7.9)	40.7 (6.3)	.608
Ankle plantarflex. max	-25.0 (11.0)	-21.5 (6.0)	.083	-27.6 (6.3)	-26.8 (6.5)	.938
Ankle abduction max	4.0 (3.9)	4.7 (3.3)	.479	3.5 (3.9)	4.8 (3.1)	.330
Ankle adduction max	-3.9 (2.6)	-4.6 (3.5)	.950	-4.4 (3.1)	-3.0 (3.5)	.227
Foot Prog max	-0.7 (5.9)	2.9 (5.0)	.072	-2.1 (8.9)	0.7 (9.1)	.310
GRF [N/kg]						
GRF ant.-post. max	0.06 (0.03)	0.06 (0.02)	.760	0.12 (0.06)	0.13 (0.01)	.904
GRF med.-lat. max	0.05 (0.02)	0.04 (0.02)	.272	0.04 (0.00)	0.04 (0.00)	.363
GRF vertical max	1.14 (0.08)	1.13 (0.05)	.731	1.35 (0.03)	1.35 (0.04)	.872
Moments [Nm/kg]						
Hip extension max	0.57 (0.22)	0.67 (0.21)	.211	0.18 (0.22)	0.38 (0.25)	.031*
Hip flexion max	-0.51 (0.18)	-0.42 (0.17)	.126	-0.49 (0.19)	-0.39 (0.14)	.016*
Hip abduction max	0.70 (0.07)	0.57 (0.11)	.001*	0.81 (0.11)	0.75 (0.12)	.153
Knee extension max	1.13 (0.23)	0.93 (0.18)	.008*	1.1 (0.31)	0.96 (0.25)	.008*
Knee valgus max	0.48 (0.22)	0.49 (0.21)	.834	0.44 (0.16)	0.48 (0.10)	.320
Ankle plantarflexion max	1.15 (0.25)	1.17 (0.13)	.737	1.14 (0.17)	1.27 (0.21)	.059
Ankle adduction max	-0.11 (0.10)	-0.11 (0.09)	.938	0.13 (0.09)	0.12 (0.09)	.865
Ankle abduction max	---	---		0.14 (0.11)	0.10 (0.06)	.266

* indicates significant differences between obese and normal-weight subjects ($p<0.05$).
— indicates no data, - indicates no statistical significance

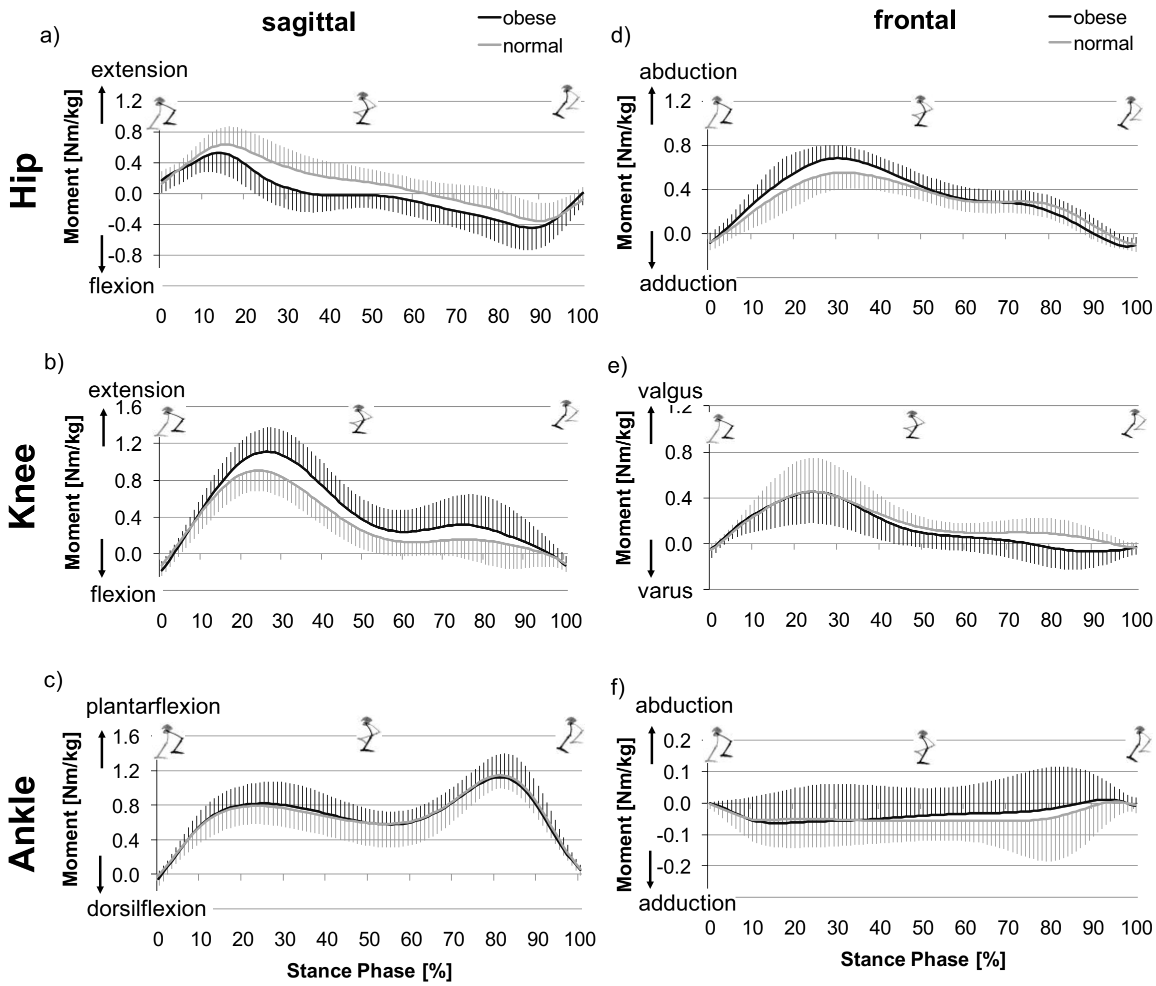

Figure 4.2: Mean hip, knee and ankle joint moments at stair ascent of obese (black) and normal-weight (grey) subjects. a) hip extension-flexion moment, b) knee extension-flexion moment, c) ankle dorsiflexion-plantarflexion moment, d) hip adduction-abduction moment, e) knee varus-valgus moment, f) ankle adduction-abduction moment.

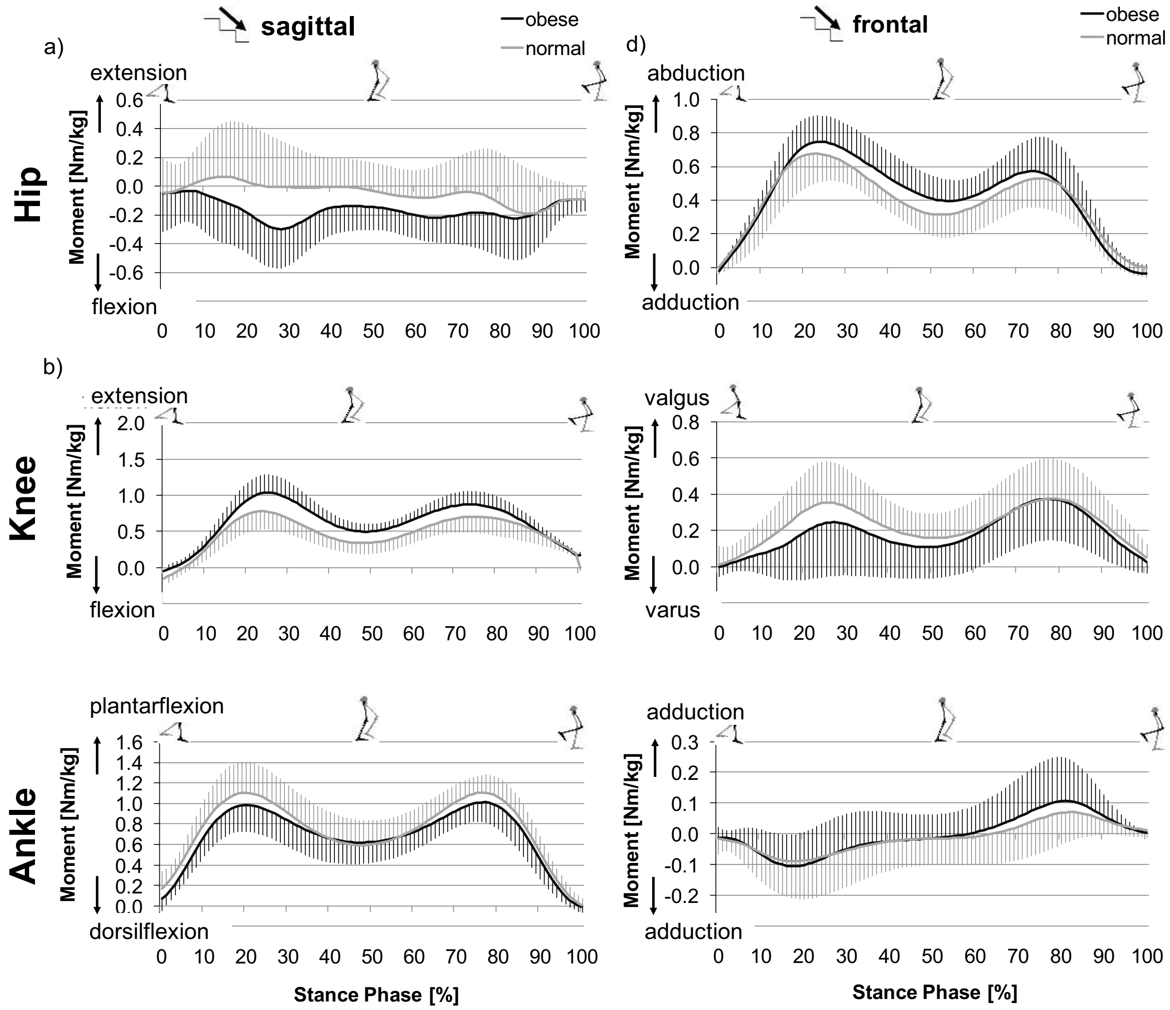

Figure 4.3: Mean hip, knee and ankle joint moments at stair descent of obese (black) and normal-weight (grey) subjects. a) hip extension-flexion moment, b) knee extension-flexion moment, c) ankle dorsiflexion-plantarflexion moment, d) hip adduction-abduction moment, e) knee varus-valgus moment, f) ankle adduction-abduction moment.

4.4 Discussion

The goal of this study was to provide initial data on differences in stair-walking biomechanics between obese and normal-weight children. This comprehensive analysis of spatio-temporal parameters, joint angles, ground reaction forces and joint moments revealed distinct differences in stair-walking mechanics between the two groups.

Obese children spent less time in single support going upstairs, and more time in double support going downstairs, both indicators that obese children spent more time in double stance. These alterations assist the maintenance of dynamic balance, because, unlike during single support, the base of support is bound in between the two feet. Similar results have been reported for level walking and generally interpreted as a representation of an underlying instability in obese persons (Spyropoulos et al., 1991; McGraw et al., 2000; DeVita & Hortobágyi, 2003; Nantel et al., 2006; Morrison et al., 2008). In contrast to results reported for level walking in obese adults (Spyropoulos et al., 1991) in this study obese children had similar step width as normal-weight children. Hence, foot width in this population seems to be unaffected by greater thigh girth associated with excess body weight.

Interestingly, differences in only a few parameters describing joint kinematics and ground reaction forces were found between obese children and normal-weight children. These results indicate that overall kinematics and kinetics are primarily given by the stair walking task. The absence of differences in normalized ground reaction forces between obese and normal-weight children suggests that differences in joint moments are mainly caused by subtle differences in joint angles, such as for instance anterior pelvic tilt, and longer phases of double support. However, absolute ground reaction forces were higher in obese children compared to normal weight children. While the musculoskeletal structures may be able to adapt to such greater absolute loads over time, a rapid weight gain (increase in body mass within a short period of time) may generate absolute loads that exceed tissue strength and lead to injury or tissue damage (Rana et al., 2009). Hence, a child's weight development is critical and should be closely monitored throughout childhood and adolescence.

Time series for the normal-weight group in this study are similar in shape and magnitude to those reported by Costigan et al. (2002), Kowalk et al. (1996) (knee joint moments) and Nadeau et al. (2003) (hip and knee flexion moments). While for level walking and standing up higher moments in one joint appear to reduce the moment of the knee (DeVita & Hortobágyi, 2003; level walking) or the hip (Sibella et al., 2003; standing up from a chair), such compensation adaptations were not observed for stair walking in this study.

Obese children had higher moments at the hip and at the knee compared to normal-weight children. More specifically, this study revealed different mechanisms between walking up and down stairs. For ascending stairs, higher hip abduction moments (Nm/kg) and higher knee extension moments (Nm/kg) were observed. Because obese children climbed the stairs with the same step width as normal-weight children, the increased hip abduction moment may have been caused by greater thigh girth in obese children. The forward-upward progression at the beginning of stance is mainly generated by the knee extension moment, which is 20% greater in obese children presumably representing an increase in muscle work. During stair descending, obese children walked with greater peak hip flexor moments (+26%) and peak knee extensor moments (+15%) possibly leading to an overloading of the joints and earlier fatigue in the stair walking task.

Overweight children have lower relative bone area and bone mass than normal-weight children (Goulding et al., 2000). Expressed in absolute values, greater loads have to be supported and transferred by relatively smaller bone area with lower bone mass. Therefore, greater moments scaled to bodyweight are a strong indication of greater loading of the musculoskeletal structures in obese children compared to normal-weight children. While within a certain window, greater loads stimulate bone to get stronger (Carter & Wong, 2003) excessive loads may cause micro damage of musculoskeletal tissues and lead to long-term degenerative joint disease such as osteoarthritis (Andriacchi et al., 2004). The differences between our results for stair walking and those reported in the literature for other daily activities emphasize the need for assessing joint kinematics and kinetics during a range of daily activities. In addition, different daily activities may be used in the context of weight reduction programs to not only improve general function but should be carefully selected to avoid overloading of musculoskeletal structures during these programs.

Messier et al. (1996) proposed that the greater prevalence of lower limb injuries in the obese is the result of altered frontal plane mechanics of the foot and the lower limb during gait. This study showed differences in the frontal plane only for the hip abduction moment and the maximum knee valgus position during stair ascent. Hence, frontal plane mechanics may not be the only factor contributing to a greater prevalence of lower limb injuries, but rather sagittal plane moments should also be taken into considerations when discussing prevalence of lower limb injuries.

Because of excess body tissue in obese children, errors in kinematic tracking resulting from skin motion artifacts may represent a limitation of this study. To date, no report on actual inter-marker distances between obese and normal-weight children is available. However, skin movement artifacts as assessed by relative marker displacement were similar for both groups. Hence, it can be assumed that the differences in biomechanical parameters between obese and normal-weight children is not caused by the accuracy of the measurement technique but they rather appear to be true differences in stair walking mechanics between the two groups.

To date, anthropometric reference data for obese children are not available. While center of mass location may not change with adiposity, segment mass and moment of inertia certainly do. Because during stance the major contributor to joint moments is the ground reaction force, the effect of anthropometric data can be considered minor (Ganley & Powers, 2004), and thus the results reported in this study are relevant.

One limitation of the inverse dynamics approach is that co-contraction of antagonistic muscles are not taken into consideration. This fact may limit the ability to infer joint loading from net muscle moments. Further, the current study did not report on impact force during stair walking because of the limitations of our technical set-up. It is possible that obese children experience greater impact forces during stair ascent and descent, and future studies should attempt to develop technical set-ups that eliminate stair oscillations.

4.5 Conclusion

To our knowledge, this is the first study to report kinetic and kinematic characteristics of stair walking in obese children. Reported movement characteristics of obese children for level gait and rising from a chair are also evident for stair walking. For descending and ascending stairs, higher hip abduction and knee extension moments during stair ascent and higher hip and knee flexion moments during stair ascent observed in obese children may contribute to a cumulative overloading of the joint until adulthood and to a higher risk of knee and hip osteoarthritis.

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5 Study IV: Effect of two different functional braces on knee joint laxity and functional achievements in patients with anterior cruciate ligament ruptures

5.1 Introduction

The main functions of the anterior cruciate ligament (ACL) are the stabilization of the knee joint for preventing hyperextension, anterior displacement of the tibia in relation to the femur and internal rotation of the tibia. Another important function of the ACL is the restrain of both valgus and varus stresses in all degrees of flexion (e.g. Andersson et al., 2009; Trees et al., 2009; Mir et al., 2008; MacDonald et al., 1996). Apart from pure mechanical stabilization the importance of proprioceptive aspects of the ACL have been highlighted (Mir et al., 2008; Beynnon et al., 2002; Fridén et al., 2001; Fremery et al., 2000; Risberg et al., 1999a, 1999b; MacDonald et al., 1996;). Injury of the ACL implies functional loss in both stability and proprioception and is associated with long term indications of meniscal injury, damage to the cartilage and osteoarthritis (Andersson et al., 2009; Quatman & Hewett 2009; Mayr et al., 2004). ACL injury is the most frequent injury at the knee (Trees et al., 2009; Senter & Hame, 2006; Salmon et al., 2005) and is referred to as the largest single problem in orthopaedic sports medicine (Renström et al., 2008). In the USA about 100.000 ACLs are torn each year, 70% of which occur during athletic activity (Senter & Hame, 2006). Most injuries are of non-contact nature (80%) and occur during lateral pivoting, cutting, landing or deceleration maneuvers during sports games (Renström et al., 2008; Senter & Hame 2006). Female athletes are at higher risk than male athletes (Quatman & Hewett, 2009; Renström et al., 2008; Hewett et al., 1999).

The main goal of ACL injury treatment is to regain functional stability of the knee joint motion with optional return to sport activity. Currently, the treatment following ACL surgery is discussed controversially, but in general ACL injury can be treated surgically or conservatively. In surgical treatment the ruptured ACL is reconstructed by a patellar tendon or hamstring tendon autograft, which should restore the stabilization of the knee joint (Andersson et al., 2009; Trees et al., 2009). The aim of conservative treatment is the achievement of stabilization by improving neuromuscular activity (Zätterström et al., 2000).

In assistance of the rehabilitation process functional knee braces can be applied. After ACL-reconstructive surgery the functional knee brace should protect the graft by reducing strain on the reconstructed ACL. Studies focusing on ACL-reconstructed knees report initial short term beneficial effects for bracing (reduced pain, less swelling), but no differences in long-term rehabilitation were observed between braced and non-braced patients (e.g. McDevitt et al., 2004; Brandsson et al., 2001; Risberg et al., 1999b; Kartus et al., 1997). Consequently, the use of a functional knee brace after ACL reconstruction might not be effective (Birmingham et al., 2008; McDevitt et al., 2004; Risberg et al., 1999a, 1999b; Kartus et al., 1997). Rehabilitation after ACL-reconstruction depends on many different factors such as surgical procedure, initial rehabilitation procedure, type of rehabilitation training, brace use or activity level of patients. Further research on optimum combinations has yet to be established.

The reason for using functional knee braces in ACL deficient knees is to improve knee joint stability and performance by reducing anterior translation of the tibia in relation to the femur

(Nemeth et al., 1997). However, research has shown that the ability of functional braces to reduce tibial displacement is only achieved for low loads but not for high-load conditions as they may appear in active patients (Beynnon et al., 2003, 1997, 1992; Ramsey et al., 2001; Wojtys et al., 1992, 1996). Ramsey et al. (2001) detected in vivo no differences in the anterior tibial displacement due to the brace. Since many patients with ACL-deficiency or ACL-rupture report subjective benefits due to functional braces such as a higher sense of stability or increased performance (Birmingham et al., 2008; Swirtun et al., 2005; Smith et al., 2003) other mechanisms than reducing the anterior tibial translation might also be essential. Functional bracing might improve the neuromuscular control of the knee due to increased proprioceptive mechanisms (Nemeth et al., 1997; DeVita et al., 1996; Wojtys et al., 1990; Branch et al. 1989). Additionally, functional knee bracing in ACL deficient persons reduced the knee extensor moment while increasing the ankle and hip extensor moments during stance phase of gait, which is assumed to be conductive to ACL strain reduction (DeVita et al., 1998; 1992). Branch et al. (1989) reported that ACL deficient patients without braces show a significant increase in hamstring muscle activity but decreased activity in the quadriceps muscle. When wearing a brace, the patient with the ACL deficient knee showed further reduction of quadriceps muscle activity along with a decrease of hamstrings muscle activity. The authors suggested that this change either indicates that braces stabilize the knee and hence lessen the need for muscle control, or braces inhibit muscle performance. In general, no consistent findings of the effect of functional bracing exist, and the discussion if functional braces provide a benefit in the rehabilitation of ACL deficient and ACL-reconstructed knee still remains controversial.

The functional braces studied in the literature are mainly rigid shell functional braces with a hinge joint and straps to hold the brace in place. These braces tend to be bulky (Nelson, 1990), migrate or slip during activity (Greene et al., 2000; Nelson, 1990), lead to an extension deficit more likely (Mayr et al., 2010) and patients often report discomfort while using (Singer & Lamontagne, 2008; Risberg et al., 1999b; Nelson, 1990). An increase in thigh atrophy has been observed for patients using rigid functional braces (Risberg et al., 1999b; Branch et al., 1989). DeVita et al. (1996) observed a reduction in mechanical work and power at the knee in healthy patients as response to functional bracing. This leads the authors to the suggestion that functional bracing may be one factor evoking the gait modifications in ACL deficient patients. This is also supported by a later work of Singer & Lamontagne (2008), who reported that the rigid shell brace appears to exert pressure at its contact points, especially at the gastrocnemius, which might have been the cause for a decrease in plantarflexor moment and a change in the gait pattern.

Based on the current discussion on how functional braces can support the rehabilitation of ACL deficient knees a neuromuscular explanation appears to be favorable over the need to provide mechanical stability (Singer & Lamontagne, 2008). Beneficial effects on proprioception have also been demonstrated by simple ortheses such as sleeves in healthy subjects (Jerosch & Prymka, 1996) and in subjects with different types of knee disorder (Beynnon et al., 2002). Birmingham et al. (2008) reported that patients provided with a neoprene sleeve have similar outcome in functional tests than patients provided with a rigid functional brace. Mayr et al. (2010) reported a superior effect of a water-filled sleeve brace regarding effusion, swelling, extension deficit and patient compliance over a rigid shell brace. Singer & Lamontagne (2008) examined the differences in the effect of the shell and sleeve type functional knee brace on 3D lower limb joint mechanics in comparison to non-braced walking in healthy subjects and found that subjects with the sleeve brace showed a gait pattern more alike to that of the non-braced healthy group. Additionally, they reported that a sleeve brace might distribute the applied force more evenly over the shank and the underlying muscle can expand more freely due to the elastic material of the brace. This might have positive effects on comfort and may lead to a decreased atrophy of the thigh muscles (Reer et al., 2001).

While the reported effects of a sleeve brace seem beneficial, little evidence is available of the effect of the different bracing types on subjects with ACL deficient knees. Therefore, the aim of this study was to investigate the effect of different brace types compared to a non-braced condition on joint laxity and functional achievement for proprioception, stability, strength and daily activity in ACL deficient patients. It is hypothesized that a difference in joint laxity and functional achievement between the sleeve brace and the rigid brace as well as between the braced and the non-braced conditions can be observed.

5.2 Methods

5.2.1 Subjects

Twenty-eight subjects (16 female, 12 male; age: 40±13 years) with ruptured ACL knees participated in this study. Twelve subjects ruptured their ACL within the last three to six months. The ACL rupture of the remaining 16 subjects occurred between 7 to 360 months (median: 36 months) prior to the study. All subjects were non-copers and experienced "giving way" episodes repeatedly. Inclusion criteria were defined as: (a) age between 18-60 years, (b) unilateral tear of the ACL without reconstruction, (c) time of rupture at least three months ago, (d) side-to-side difference in knee laxity ≥ 3 mm evaluated via the KT-1000^{TM} anthrometer (MEDmetric, San Diego, California[a]), (e) functional instability measured via hop test (index of symmetry > 85%), (f) >1 "giving way" since ACL rupture, (g) no signs of gonarthrosis ≥ 2°, (h) no injuries of the posterior cruciate ligament, (i) contralateral side must be free of injuries due to optional impairment of the functional outcome. The study was approved by the ethics board and informed consent was signed for all subjects participating in the study.

5.2.2 Braces and preparation

Subjects were provided with a sleeve brace (SofTec Genu, Bauerfeind Germany Inc, Zeulenroda) (Figure 5.1a) and a rigid shell brace (4Titude Donjoy, ORMED GmbH, Freiburg) (Figure 5.1b). Both braces were individually fitted by an orthopedic technician and subjects were familiarized with the correct positioning of the braces. For reducing learning effects all subjects came to the laboratory for two habituation sessions prior to the actual measurements to become familiar with the tests and the entire procedure. All subjects completed two testing sessions. The tests of each session were completed in one condition (sleeve brace, shell brace, non-braced) before the subjects changed to the next condition. Five minutes were given to the subjects to get accustomed to the new condition prior to the measurements. By using a balanced randomization scheme the order of sleeve braced, shell braced and non-braced condition was randomized.

a) b)

Figure 5.1: a) Sleeve brace (SofTec Genu, Bauerfeind Germany Inc., Zeulenroda); b) rigid shell brace (4Titude Donjoy, ORMED GmbH, Freiburg).

5.2.3 Testing protocol

Static anterior laxity was measured using the KT-1000™ arthrometer[a] with an applied force of 98 N. Data was collected of the injured side in non-braced and both braced conditions as well as of the uninjured side in non-braced condition.

Joint position sense was measured using the angle reproduction test. Subjects were seated with the feet hanging free and visual sight was blinded by an eye mask. The injured leg was moved to a random knee angle between 0 and 90° by the tester and held in position for three seconds by the subject him/herself. After bringing the leg in neutral position the subjects had to recapture the felt knee joint angle. The mean absolute angular difference between given and recaptured angle of 10 trials was taken as criterion.

Postural control was analyzed by testing static and dynamic balance. Static balance was identified by the tests (a) single leg stance on a stable surface (AMTI, model BP600900, Advanced Mechanical Technology, Watertown, MA, 1000 Hz[b]) with eyes closed, 10 seconds (Figure 5.2.a), (b) single leg stance on an instable surface (Posturomed, Haider Bioswing, Pullenreuth[c]), 15 seconds (Figure 5.2.b). Dynamic balance was identified by the tests (c) single leg stance on an instable surface with 2.5 cm lateral perturbation (Posturomed[c]), 1 to 5 seconds after release, (d) landing after a 30 cm forward single leg counter movement jump (CMJ) onto a force plate (AMTI[b]), 5 seconds, and (e) landing after a 30 cm forward single leg CMJ with a 90° inward turn about the longitudinal axis on a force plate (AMTI[b]) (Figure 5.2.c), 5 seconds. For further analysis the path length and the standard deviation of the path in anterior-posterior and medio-lateral direction (of either the center of pressure - COP, when using the force platform AMTI[b] or of the platform, when using Posturomed[c]) were used as variables.

The specific recording times have been identified in pretests to target either the ability of stabilization after activity or general postural control. For non-braced and both braced conditions of the injured leg five valid trials were conducted for each test, of which the best and worst trial (in terms of path length) were not considered for further analysis. Mean values of the remaining three trials were calculated and taken for further analysis.

Injured and uninjured lower limb strength was analyzed in isometric and dynamic conditions. Maximum isometric lower limb extension strength was tested on an instrumented leg press equipped with a left-right-separated force plate (self-construction, BioMotion Center, Karlsruhe, 1000 Hz[e]) (Figure 5.2.d). Subjects were positioned with 120° knee angle and each foot was placed at the respective part of the force plate. Subjects were instructed to press with maximum strength for 3 seconds. To analyze dynamic lower limb strength subjects performed a counter movement jump (CMJ) with arms akimbo on a left-right-separated force platform (self-construction[e]) (Figure 5.2.e). Force dependent variables were normalized to body mass. The best out of three trials regarding peak force (isometric lower limb extension strength) and maximum jump height (CMJ) was taken for further analysis. The criterion for isometric leg extension strength was the peak isometric force (F_{peak}isom). Parameters for the CMJ were maximum jump height (h_{max}), peak force (F_{peak}) and peak rate of force development (RFD_{peak}). In each brace condition data was collected for the injured and uninjured side simultaneously.

Level gait and running were also included in the test protocol to analyze daily living tasks. Horizontal (anterior-posterior, medio-lateral) and vertical ground reaction forces were collected (AMTI[b]) of subjects walking and running at a comfortable self-selected pace. Five valid trials were taken for analysis. A trial was considered valid when the entire foot was placed on the force platform and no visible alterations of movement pattern could be identified. Ground reaction forces were normalized to body mass and peak vertical and horizontal ground reaction forces were identified for each trial. Out of the five trials mean peak vertical and horizontal ground reaction forces were calculated and considered for further analysis. The force plates were mounted in the floor in such a way that in gait both left and right ground contact could be recorded within one single trial, so injured and uninjured side could be measured simultaneously. In running only data for the injured side could be recorded in all three conditions per trial.

a) b) c) d) e)

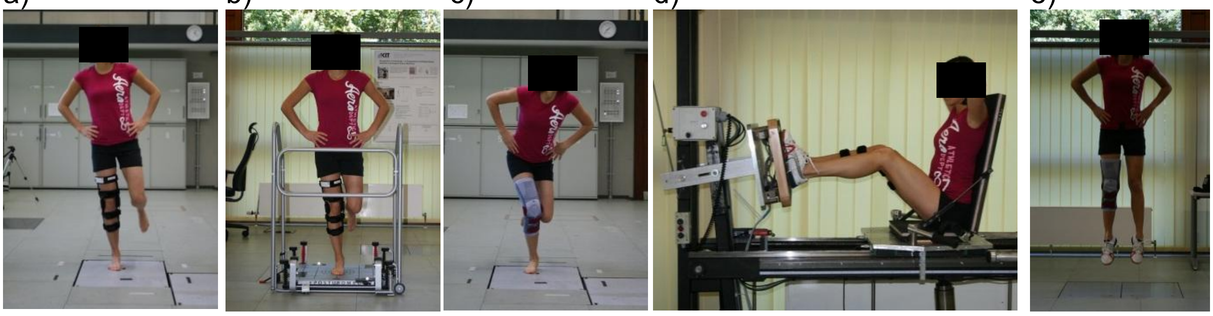

Figure 5.2: a) single leg stance on stable surface with eyes closed, (b) single leg stance on instable surface, c) landing after a 30 cm forward single leg CMJ with a 90° inward turn about the longitudinal axis, d) isometric lower limb extension strength, e) CMJ

5.2.4 Statistical analysis

Overall significance was calculated using a one-way repeated measure ANOVA including a Bonferroni adjustment. The level of significance was set at $p \leq 0.05$ for overall significance. Necessary requirements for normality and sphericity were given. Partial eta² (η^2_p) was calculated for overall significances. Borders for effect size η^2_p were set to be: η^2_p=0.01 for small, η^2_p=0.06 for medium and η^2_p=0.14 for high effect sizes. If a significant effect was found t-tests were used as post-hoc tests to assess pair-wise comparisons. For post-hoc tests a Bonferroni adjustment was performed by reducing the level of significance to $p \leq 0.016$ (three conditions: non-braced, sleeve braced, shell braced). Cohen's d (*d*) with pooled standard deviation for each pair-wise comparison was calculated for effect sizes of post-hoc tests. Cohen's *d* was quantified to be small <0.40, medium between 0.40 - 0.79 and high with effect size $d \geq 0.80$.

5.3 Results

Mean (95%-CI) values of test parameters and respective p-values and effect sizes are presented in Table 5.1. Subjects wearing a sleeve brace showed a significant decrease in joint laxity by 32% compared to the non-braced condition. Results for the dynamic balance showed that sleeve braced subjects significantly reduced medio-lateral standard deviation of the path length when standing on an unstable lateral perturbed surface by 10%. Additionally, a trend to decreased anterior-posterior and medio-lateral standard deviation of the path length when stabilizing the body after a single leg CMJ with 90° medial rotation could be observed for the sleeve braced condition. The strength tests revealed that subjects wearing a sleeve brace significantly increased their rate of force development by 18% at the injured leg.

When subjects were provided with a rigid shell brace no significant differences were observed in all tests compared to the non-braced conditions. Trends were identified for the knee joint laxity and for the postural control test. The medio-lateral standard deviation of the path length was decreased when standing on a perturbed surface and for anterior-posterior and medio-lateral standard deviation of the path length when stabilizing the body after a single leg CMJ with 90° medial rotation.

Significant differences between the two brace types were identified for knee joint laxity and peak rate of force development at the CMJ. The sleeve brace showed a reduction of laxity by 21% and an increase for rate of force development by 19% compared to the rigid shell braced condition.

All other tests revealed comparable values between non-braced, sleeve-braced and shell braced conditions.

Table 5.1: Mean (95%-CI) values with overall p-values (η^2_p) and post-hoc tests (Cohens d).

test parameter	non-braced (95%-CI)	sleeve (95%-CI)	shell (95%-CI)	p overall (e^2)	p nobraced-sleeve (d)	p nonbraced-shell (d)	p sleeve-shell (d)
KT 1000							
98 N [mm]	8.3 (1.3)	5.6 (0.7)	7.1 (1.1)	<0.001 (0.46)	<0.001 (0.94)	0.039 (0.37)	0.001 (0.36)
angle reproduction test							
mean abs difference[°]	2.5 (0.5)	2.3 (0.4)	2.4 (0.3)	0.762 (0.01)	---	---	---
stable, closed eyes							
path length_10s [m]	1.01 (0.14)	1.01 (0.15)	0.98 (0.13)	0.822 (0.01)	---	---	---
ant.-post._10s [mm]	11.84 (1.66)	11.10 (1.20)	0.01 (31.15)	0.907 (0.00)	---	---	---
med.-lat._10s [mm]	11.03 (1.10)	11.21 (1.01)	0.01 (21.10)	0.440 (0.03)	---	---	---
instable							
path length_15s [m]	0.45 (0.08)	0.46 (0.13)	0.47 (0.14)	0.901 (0.00)	---	---	---
ant.-post._15s [mm]	0.86 (0.13)	0.87 (0.21)	0.89 (0.16)	0.992 (0.00)	---	---	---
med.-lat._15s [mm]	1.54 (0.27)	1.54 (0.36)	1.56 (0.30)	0.932 (0.00)	---	---	---
instable, perturbation							
path length_1-5s [m]	0.25 (0.03)	0.22 (0.03)	0.22 (0.03)	0.053 (0.12)	---	---	---
ant.-post._1-5s [mm]	1.63 (0.21)	1.29 (0.16)	1.39 (0.13)	0.071 (0.11)	---	---	---
med.-lat._1-5s [mm]	4.11 (0.57)	3.70 (0.64)	3.71 (0.54)	0.011 (0.19)	0.016 (0.83)	0.180 (0.50)	0.656 (0.16)
landing single leg forward CMJ							
path length_5s [m]	0.62 (0.07)	0.60 (0.06)	0.64 (0.04)	0.411 (0.04)	---	---	---
ant.-post_5s [mm]	16.97 (2.09)	16.46 (1.81)	18.36 (1.67)	0.373 (0.04)	---	---	---
med.-lat_5s [mm]	8.91 (20.91)	8.78 (20.86)	9.54 (10.74)	0.075 (0.12)	---	---	---
landing single leg forward 90° CMJ							
path length_5s [m]	0.75 (0.06)	0.74 (0.07)	0.70 (0.05)	0.059 (0.11)			
ant.-post_5s [mm]	19.55 (2.03)	18.53 (1.05)	17.75 (1.71)	0.022 (0.16)	0.128 (0.49)	0.061 (0.68)	0.820 (0.01)
med.-lat._5s [mm]	11.67 (1.09)	11.07 (1.04)	10.32 (0.90)	0.045 (0.12)	0.121 (0.52)	0.149 (0.53)	1.000 (0.18)
isometric peak force, leg press							
F_peak_total [N/KG]	4.41 (0.50)	4.46 (0.52)	4.53 (0.56)	0.389 (0.04)	---	---	---
F_peak_uninjured side [N/KG]	2.27 (0.28)	2.31 (0.29)	2.32 (0.30)	0.435 (0.03)	---	---	---
F_peak_injured side [N/KG]	2.15 (0.24)	2.19 (0.25)	2.22 (0.26)	0.420 (0.03)	---	---	---
CMJ							
h [m]	0.24 (0.04)	0.26 (0.03)	0.25 (0.04)	0.312 (0.05)	---	---	---
F_peak_total [N/kg]	2.13 (0.07)	2.15 (0.12)	2.08 (0.13)	0.080 (0.11)	---	---	---
F_peak_uninjured side [N/kg]	1.12 (0.04)	1.14 (0.06)	1.11 (0.05)	0.151 (0.07)	---	---	---
F_peak_injured side [N/kg]	1.04 (0.04)	1.06 (0.05)	1.04 (0.05)	0.107 (0.09)	---	---	---
RFD_peak_total [N/s/kg]	10.77 (1.37)	12.10 (1.87)	10.95 (1.82)	0.073 (0.10)	---	---	---
RFD_peak_uninjured side [N/s/kg]	6.12 (0.98)	6.73 (1.36)	6.29 (1.27)	0.181 (0.07)	---	---	---
RFD_peak_injured side [N/s/kg]	5.98 (0.70)	7.02 (1.11)	5.88 (0.90)	0.003 (0.22)	0.015 (0.42)	1.000 (0.04)	0.016 (0.21)
gait							
F_ant-post_noninjured side [N/kg]	0.23 (0.02)	0.24 (0.02)	0.23 (0.02)	0.778 (0.01)	---	---	---
F_ant-post_injured side [N/kg]	0.23 (0.02)	0.23 (0.02)	0.23 (0.02)	0.387 (0.05)	---	---	---
F_med-lat_noninjured side [N/kg]	0.07 (0.01)	0.08 (0.02)	0.08 (0.01)	0.562 (0.03)	---	---	---
F_med-lat_injured side [N/kg]	0.07 (0.02)	0.07 (0.02)	0.07 (0.02)	0.917 (0.00)	---	---	---
F_vert_noninjured side [N/kg]	1.14 (0.06)	1.15 (0.06)	1.15 (0.06)	0.658 (0.02)	---	---	---
F_vert_injured side [N/kg]	1.13 (0.02)	1.15 (0.02)	1.14 (0.02)	0.437 (0.04)	---	---	---
running							
F_ant-post_injured side [N/kg]	0.25 (0.02)	0.24 (0.02)	0.25 (0.02)	0.909 (0.00)	---	---	---
F_med-lat_injured side [N/kg]	0.07 (0.02)	0.07 (0.02)	0.07 (0.03)	0.780 (0.01)	---	---	---
F_vert_injured side [N/kg]	2.15 (0.15)	2.17 (0.13)	2.16 (0.14)	0.402 (0.04)	---	---	---

5.4 Discussion

Research has not yet clearly evidenced a beneficial effect of functional braces in the rehabilitation of ACL injuries. While rigid shell braces seem to lead to an increase of muscle atrophy and a change of movement pattern without improving the laxity under weight-bearing conditions little is known about the effect of sleeve braces on subjects with ACL deficient knees. Therefore, the aim of this study was to investigate the effect of different brace types compared to a non-braced condition on joint laxity and functional tests for proprioception, stability, strength and daily activities in ACL deficient patients.

One reason for using functional braces is to reduce the anterior translation of the tibia with respect to the femur. Passive knee joint laxity was significantly reduced by the sleeve brace and the rigid shell brace showed a reduction trend. However, the passive situation does not reflect the situation at which a reduction of anterior tibial translation needs to be achieved. Functional braces should provide protection in dynamic weight-bearing situations. Wojtys et al. (1996) report a significant reduce of laxity by 29-39% when the knee was braced and muscles were relaxed, with muscle activation bracing reduced laxity significantly by 70%-85%. Beynnon et al. (2003) complement these findings in reporting that bracing a knee with a chronic ACL tear was effective in reducing abnormal anterior-posterior laxity during non-weight-bearing and weight-bearing tasks. When subjects performed a transit from non-weight-bearing to weight-bearing functional braces were not effective in reducing anterior displacement of the tibia relative to the femur, which was 3.5 times more pronounced in injured than in healthy knees. It is difficult to assess the joint laxity since the compliant soft tissues surrounding the knee challenge the ability of braces to control motion of these bones (Fleming et al., 2000). Therefore, Ramsey et al. (2001) implanted Steinmann pins into the bone to directly measure displacement of the tibia relative to the femur during jumps for maximal horizontal distance in subjects with ACL deficient knees. They found that rigid functional shell bracing of the ACL deficient knee did not cause changes in displacement of the tibia relative to the femur compared with the unbraced condition.

Joint position sense is impaired in ACL ruptured knees, mainly due to the loss of the afferent information of mechanoreceptors in the ACL (Beynnon et al., 2002; Fridén et al. 2001, Lephart et al., 1992; Skinner & Barrack, 1991). The effect of bracing on joint position sense in ACL deficient knees is not well understood yet. Studies present contrast findings due to differences in subject group, study design and measurement techniques. While some studies show positive effects on proprioception from wearing a functional brace or a bandage (Birmingham et al., 2008; Lephart et al., 2000; Jerosh & Prymka, 1996; McNair et al., 1996; Barrett, 1991) other studies show no effect at all (Beynnon et al., 2002, 1999; Risberg et al. 1999a, 1999b; MacDonald et al., 1996; Fridén et al., 2001). The latter is supported by this study. We found no evidence of an effect of bracing on joint position sense in our subjects. This might be due to the fact that joint position sense can be at least partially regained in the healing process by compensatory mechanisms. ACL deficient subjects may learn to rely more sensitively on sensory information of the neuromuscular system (muscle spindle afferents) and of remaining joint structures to aid dynamic joint stability (Cooper et al., 2005b; Beynnon et al., 2002). Additionally, it has been shown in patients with ACL reconstruction that deficits in the joint position sense are only reduced after three to six months (Fremery et al., 2000) and eleven months, respectively (Mir et al., 2008). Therefore an improvement in joint position sense might not be necessary. The ACL of the subjects in this study was torn between 3 and 360 months prior to the study, hence the subjects might have already learned to use alternative mechanisms to identify knee joint position. Another aspect might be that the passive positioning test used in this study does not have the sensitivity necessary to detect changes in proprioception. Other tests such as threshold to detection of passive motion might have resulted in more sensitive data (Cooper et al. 2005b; Beynnon et al., 2002; Fridén et al., 2001), but this cannot be addressed in this study.

Little is known about the effect of functional bracing on static and dynamic balance of ACL deficient subjects. No study has been identified by the authors to address the influence of bracing on static balance and only little is known about the influence of bracing in dynamic balance. While bracing in the rehabilitation of ACL reconstruction does not seem to provide superior results for the hop test for maximum distance (McDevitt et al., 2004; Risberg et al., 1999a, 1999b; Kartus et al., 1997), biomechanical effects of braces in patients with ACL deficient knees on dynamic stability is limited. Smith et al. (2003) showed that in some subjects brace use indicated a more favorable muscle firing pattern than without brace use in hop tests for maximum distance. Ramsey et al. (2001) did not observe an effect of a rigid functional shell brace in jumps for maximum distance in the anterior translation of the knee as reported earlier. Detailed information of the ability to stabilize can be provided in analyzing the body sway. Patients with both chronic and symptomatic unilateral ACL deficiency were shown to have a bilateral defect in postural control when body sway was measured during the single limb stance on a force plate (Zätterström et al., 1994). No significant difference between sleeve braced, shell braced and non-braced condition were identified in the presented study for static and most of dynamic stability tasks. This might be due to the fact that patients of this study adapted to the loss of afferent information of the torn ACL and managed to provide stability based on neuromuscular feedback. The only significant difference was identified for the dynamic stability test on an instable platform responding to a lateral perturbation. For the sleeve braced condition the medio-lateral standard deviation of the platform's sway was significantly reduced by 10%. It seems that for the ability to react under weight-bearing situation on a perturbation impulse the sleeve brace leads to enhanced stabilization. It also seems that the mechanism to adapt to the sensory loss of the ACL only takes effect in less complex situations. It has been suggested that bracing and bandaging stimulates cutaneous receptors around the joint (Fridèn et al., 2001; McNair et al., 1996), which might lead to proprioceptive benefits. In addition to the flexible textile of sleeve braces that allow undisturbed muscle function this might be the cause why only the sleeve brace led to significantly better body stabilization than the shell brace.

The main muscles to stabilize the knee joint are the quadriceps, the hamstrings and the gastrocnemius muscle. While the quadriceps provides stability of the joint and also can increase anterior translation of the tibia, the hamstring muscles are the antagonist muscles of the anterior drawer. Both, quadriceps strength (Keays et al., 2003) and hamstring strength (Wilk et al., 1994) have been reported to correlate highly with functional stability. The most significant effect of an ACL tear is thigh muscle atrophy (Lautamies et al., 2008; Keays et al., 2003, 2000; Fridén et al., 2001) leading additionally to a decrease in joint stability. Again the effect of bracing on lower limb strength is discussed controversially. Some shell braces improved reflex contraction time of the muscle, which was most pronounced in the quadriceps (Beynnon et al., 2002; Wojtys et al., 1996). Nemeth et al. (1997) reported that ACL deficient skiers wearing a functional knee brace demonstrated increased lateral hamstrings electromyographic activity during periods of increased knee flexion. It was also suggested that functional knee bracing may dynamically stabilize the ACL deficient patient by stimulating increased hamstrings activity (Nemeth et al., 1997; O'Connor, 1993; Solomonow et al., 1987). Negative effects of rigid shell braces were identified by an increased loss of muscle volume and muscle strength after ACL reconstruction wearing a brace compared to the unbraced group (Risberg et al., 1999b) as well as by a delayed onset of voluntary contraction for hamstrings muscles indicated by some rigid shell braces (Beynnon et al., 2002; Wojtys et al., 1996). McDevitt et al. (2004) did not find a difference in isokinetic strength testing between braced and non-braced groups after ACL reconstruction.

Some evidence is provided that sleeve braces do not show detrimental effects on muscle strength after ACL reconstruction. Swirtun et al. (2005) did not find differences between a braced and an unbraced group in isokinetic maximum quadriceps and hamstrings peak torque. Also a study by Reer et al. (2001) showed a 25% lower reduction in circumference of the thigh muscles in patients with ACL deficient knees using a functional sleeve brace of the same type as used in the presented study. Additionally, Singer & Lamontagne (2008) suggested that the force applied by this type of brace can be more evenly distributed over the shank and the gastrocnemius due to the elastic material of the brace. The main finding of the present study is a significant increase of peak rate of force development in the injured leg when provided with the sleeve brace compared to the non-braced condition. This indicates that muscle force can be produced significantly faster with a sleeve brace than in the unbraced condition, which might be an important factor in stabilizing the knee joint and preventing further injuries. Consequently, this ability might also explain the significant decrease in medio-lateral standard deviation of the path length in response to a perturbation. No significant differences were identified for isometric and dynamic peak force between all three conditions.

The gait of subjects with ACL deficient or ACL reconstructed knees is altered compared to normal gait. DeVita et al. (1996) suggested that the gait alterations are induced by brace use. Fridén et al. (2001) proposed, however, that the flexed knee is a protective mechanism designed to avoid a dangerous zone near full knee extension at which proprioception deficits have been detected. Another mechanism might be the possibility to reduce shock with a more flexed knee (Potthast, 2005). Additional research on ACL reconstructed knees on the influence of bracing revealed that a functional brace caused an increase in extensor angular impulse at the hip and ankle, an increase in the work produced at the hip, and a decrease in the work produced at the knee. It has been shown that these effects are about three times larger in subjects with ACL reconstructed knees than in healthy subjects and may be beneficial to patients with recent ACL reconstructions (DeVita et al., 1998, 1996). The reduced extensor moment at the knee in the braced condition indicated that the load on the recently reconstructed ligament was reduced and that the brace protected the ligament during the stance phase of walking (DeVita et al., 1998). Lu et al. (2006) identified a significant increase of the peak knee abductor moments in ACL deficient knees with bracing, but also a reduction in bilateral kinetic asymmetry. Singer & Lamontagne (2008) studied the influence of sleeve and shell braces on gait of healthy subjects. They identified a more flexed knee in both bracing conditions. With the shell brace the peak knee adduction angle was significantly greater, and the peak knee internal rotation angle was significantly smaller than during non-braced walking. Both alterations are likely due to the positioning of the shell brace, which might have positioned the shank in an increased adduction and decreased internal rotation throughout the gait cycle. The kinematic pattern with the sleeve brace resembled more closely that of the non-braced condition, which should be a main goal. The differences, however were very small and the clinical and functional relevance is not clear. The present study only analyzed the ground reaction forces during gait and running in ACL deficient subjects. No effect of bracing regarding peak vertical and horizontal ground reaction forces could be detected. Ground reaction forces, however, only provide little information on gait patterns. While kinetic parameters might be the same between all three conditions, kinematic patterns can change, leading to differences in joint moments and ACL strain, respectively. Given the increased impact braces generally had on ACL deficient gait compared to healthy subjects (DeVita et al., 1998), future research should address different brace types on joint moment patterns of ACL deficient patients to complement findings of Singer & Lamontagne (2008).

The very inhomogeneous group of subjects in terms of age, activity level and time of rupture might be a limitation of this study. It is representative though reflecting the group of patients clinicians treat in daily life. It has been suggested that bracing might have different effects in early and late rehabilitation (Cooper et al., 2005a; Liu-Ambrose et al., 2003), which probably might have also influenced the results of the present study.

Even though the effect of bracing was tested under dynamic situations in the present study the loads applied were well below those that cause injury or those that might appear in a sportive task. Athletes with an injured, disrupted, or reconstructed ACL depend on a functional brace for protection during activities that produce substantial compression and shear loads across the knee, and not only moderate loads as created in this study (Cooper et al., 2005b; Beynnon et al., 2003). It could be shown that ACL deficient subjects were able to stabilize the body in simple tasks even without bracing support. The positive effects of sleeve bracing with respect to balance and stabilization only emerged when the system is more challenged as it might happen in more exposed situations and sportive tasks. This aspect should be considered in future studies.

Since subjects used the different braces solely for testing, only short time effects of different brace types can be reported. It is of specific interest to study the long term effects when different braces are used in normal rehabilitation routine in ACL deficient and ACL reconstructed patients. An additional limitation of the study is that only one type each of rigid shell and sleeve brace was examined. Therefore the results cannot be generalized since different braces might have revealed different findings (Beynnon et al. 2003).

5.5 Conclusion

The current study investigated the effect of sleeve and rigid shell brace design on knee joint laxity and functional achievements in ACL deficient subjects. The results showed that the sleeve brace led to a significant decrease in knee joint laxity, significant decrease of medio-lateral standard deviation of platform sway after perturbation and an increase in peak rate of force development. The shell brace showed no significant changes compared to the non-braced condition. It is suggested that the sleeve brace enhances the proprioceptive benefit in strength, balance and stabilization tasks. This might be caused by the flexible area of support and the incorporated mechanisms to address proprioceptive aspects. The effects, however, were only observed in complex situations, which might indicate that subjects adjusted partly to the loss of sensory information, usually provided by an unaffected ACL. Sleeve braces might not be needed in simple daily life tasks, but could provide – compared to a shell brace – effective and beneficial support in more dynamic settings like exposed situations or when patients return to sportive activities after an ACL injury.

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6 Summary and prospective on future research

6.1 Summary

The forces acting on the human muskulo-skeletal system and the corresponding responses that are evoked by these forces such as locomotion, deformation of soft and hard tissue, growth and development of biological tissue, but also acute and chronic injuries are a major field of interest in sports biomechanics (Nigg, 2000). In connection with the progress in measurement instrumentation and in computer technology new possibilities are feasible for the advancement of the study of human locomotion and for the extension of kinetic analysis from the purely sportive application to the analysis of clinical problems (Andriacchi & Alexander, 2000; Gollhofer & Müller, 2009).

Not directly observable inter-segmental forces and moments are usually approximated by measuring the external ground reaction force as well as the three-dimensional positioning of the segments with tracking markers placed on the skin. Both serve as input variables for models to describe the kinematics and kinetics of the musculo-skeletal system (Andriacchi & Alexander, 2000). Several modeling approaches to approximate these parameters currently exist. The accuracy of the approximations highly depends on measurement instruments, assumptions taken in the model and the model itself (Cappozzo et al., 2005).

This thesis wants to give an insight into the broad spectrum of possible applications of sports biomechanics. This is realized by a methodological part in *Chapter 2,* in which the issue of different model approches used for analysing kinematics and kinetics is adressed and by an applied-science oriented part in *Chapter 3* to *5*, in which the focus is set on prevention and rehabilitation. These chapters represent three independent studies, each of them invesitgates a seperate currently discussed research question and uses different methods. The three subject groups “healthy subjects”, “subjects at risk for overuse injury” and “injured subjects” are adressed in these studies to embrace the broad spectrum of application possibilities. Every chapter provides new insights of human movement and increases the knowledge sports biomechanists can provide in order to develop adequate training programs, to understand mechanisms of injury and to enhance rehabilitaiton programs. *Chapter 3* analyzes lower limb loading of healthy subjects performing variations of squats. *Chapter 4* adresses the lower limb loading of obese children as “subjects at risk for overuse injury” who are climbing stairs. The effect of brace type on functional tests and knee joint laxitiy of patients with anterior cruciate ligament ruptured knees as “injured subjects” is reported in *Chapter 5.* In the following section each chapter is summarized independently.

Study I: The effect of calculating kinematics and kinetics of the squat movement with three different models

A variety of different models exist to analyze human movement. Models differ among other aspects in marker-position, measured variables, degrees of freedom assigned to the joints, as well as in anatomical and technical references, joint rotation conventions and terminology. Hence data assessed by different models is difficult to compare. The purpose of this study was to identify the effect of three currently used models on joint kinematic and kinetics in a sportive relevant setting. Fifteen male subjects performed a squat with an additional mass of 20 kg. Kinematic and kinetic recordings were collected using an infrared camera system (VICON, 200 Hz) and two force platforms (AMTI, 1000 Hz). The parameters of interest were calculated using (a) the Vicon Plug-in Gait (PiG) model, (b) a recursive multi-body algorithm (MKDtools) and (c) the Visual3D (V3D) model. Results showed significant kinematic differences in sagittal, frontal and transversal plane for the hip, knee and ankle joint. Kinetic results revealed significant differences in sagittal and transversal plane for the hip and the knee joint moments. Results clearly showed that different models lead to distinct differences in kinematic and kinetic outcome. Additionally, it was not possible to either identify systematic differences between the three different models, or draw conclusions that two models consistently lead to the same results. This study enforces the need for great caution when comparing data of studies using different analyzing methods. Especially for comparing and interpreting cross sectional studies, specifications of the used model have to be provided in order to enable a good judgment of the data.

Study II: Lower limb loading of three different variations of squat

Squats are commonly incorporated in general fitness training and in rehabilitation programs and are performed in many different variations. To what extent these lead to different lower extremity joint loading has yet to be established. The aim of this study was to compare the lower extremity joint kinematics and moments of three squat variations. 16 subjects performed with an additional mass of 20 kg a standard squat (“standard squat”, a squat with the knee shifted over the vertical line of the toe (“knee-shifted squat”) and a squat on a declined surface (“declined squat”). Kinematic and kinetic recordings were collected using an infrared camera system (10 cameras, VICON, 200 Hz) and two force platforms (AMTI, 1000 Hz). A recursive multi-body algorithm (MKDtools) was used to calculate lower limb kinematics and inverse dynamics as a function of the knee angle. Sagittal ankle moments significantly increased for the knee-shifted squat in comparison to the standard squat by 360% (overall $p \leq 0.001$) and to the declined squat by 587% (overall $p \leq 0.001$). No significant differences were detected between standard and declined squat. The knee-shifted squat can be recommended if an increase in plantar flexor muscles is desired without increasing knee and hip joint moments. Squatting on an inclination of approx. 10° does not lead to alterations of the joint moments compared to a standard squat and, hence, does not lead to beneficial effects with respect to joint loading.

Study III: Stair walking adaptations in obese children: Spatio-temporal, kinematic and kinetic differences to normal-weight children

Mechanical factors and body weight are important factors in the development and progression of varus/valgus angular deformities of the knee and have long term implications on the person’s health including increased risk of osteoarthritis. However, limited information is available on the interrelation of these factors during dynamic activities in children. The purpose of this study was to test the hypothesis that during stair walking lower extremity joint moments normalized to body mass are greater in obese children than those in normal-weight children. Eighteen obese children (10.5±1.5 yrs, 148±10 cm, 56.6±8.4 kg) and 17 normal-weight children (10.4±1.3 yrs, 143±9 cm, 36.7±7.5 kg) were recruited. A Vicon camera system (200 Hz) and two AMTI force plates (1000 Hz) were used to record and analyze the kinematics and kinetics of ascending and descending stairs. Significant differences in spatio-temporal, kinematic and kinetic parameters during ascending and descending stairs between obese and normal-weight children were detected. For stair ascent, significantly higher hip abduction moments (+23%; p=0.001) and significantly higher knee extension moments (+20%; p=0.008) were observed for the obese children. For stair descent, significantly lower hip extension moments (-52%; p=0.031), significantly higher hip flexion moments (+25%; p=0.016) and significantly higher knee extension moments (+15%, p=0.008) were observed for obese subjects. Until today, it is unclear if the body adapts to greater joint moments in obese children and how. Nevertheless, these differences in joint moments may contribute to a cumulative overloading of the joint through adolescence into adulthood, and potentially result in a higher risk of developing knee and hip osteoarthritis.

Study IV: Effect of brace type on knee joint laxitiy and functional test of patients with anterior cruciate ligament ruptured knees

Research has not yet clearly shown a beneficial effect of functional braces in the rehabilitation of ACL injuries. While rigid shell braces seem to lead to an increase of muscle atrophy and to a change of movement pattern without improving the laxity under weight-bearing conditions, sleeve braces could lead to benefits in rehabilitation due to an increased proprioceptive stimulation. However, little is known about the effect of sleeve braces on subjects with ACL deficient knees. Therefore, the aim of this study was to investigate the effect of different brace types compared to a non-braced condition concerning joint laxity and functional achievements in ACL deficient patients. Twenty-eight subjects with ACL ruptured knees were provided with a sleeve brace (SofTec Genu, Bauerfeind) and a rigid shell brace (4Titude Donjoy, Ormed). Data was collected of the injured side in a non-braced condition and two braced conditions in tests for knee joint laxity, joint position sense, static and dynamic balance, isometric and dynamic lower limb extension strength, as well as in the daily activity tasks gait and running. The results showed a significant decrease in knee joint laxity (32%; p<0.001), a significant decrease of medio-lateral standard deviation of platform sway after perturbation (10%; p=0.016) and a significant increase in peak rate of force development (18%; p=0.015) for the sleeve brace. The shell brace showed no significant changes compared to the non-braced condition. It is suggested that the sleeve brace enhances the proprioceptive benefit in strength, balance and stabilization tasks. The effects, however, were only observed in complex situations. Sleeve braces might not be needed in simple daily life tasks, but – compared to a shell brace – could provide effective and beneficial support in dynamic activities.

6.2 Limitations and prospective on future research

- One methodological limitation of the inverse dynamics approach is that co-contraction of agonistic muscles are not taken into consideration. Studies reporting joint moments calculated by inverse dynamics as well as measured in vivo report an underestimation of the loads by the calculation approach.

- Calculating the net joint moments via inverse dynamics is the first step in understanding the forces acting in the respective joints. However, the ability to conclude on specific structural joint loading (such as changes in ligament forces, compressive forces and shear forces) by just referring to net joint moments is limited. Further detailed calculations to approximate the load of specific structures need to be applied, especially when exercises are used in order to adequately load or unload these structures as it might be intended in rehabilitation processes. The lower limb loading in three variations of the squat served as one example for these types of exercises. However, only net joint moments were reported and the data needs to be extended with specific structural analyzes. Also further research has to be established on the effect of higher lifting loads, of different inclination angles of the standing surface, and of the effects further variations of the squat impose on the musculo-skeletal system.

- Furthermore, the approach to approximate inter-segmental loads highly depends on the measurement instruments, assumptions taken in the model and the model itself. A variety of instrumental errors occur during measuring, data processing and analyzing the respective data. One limitation is the error emerging from concluding the segmental movement by tracking markers on the skin. This approach is based on the assumptions that (a) the markers can be placed directly over the corresponding bony landmark and (b) skin movement is either negligible or is reduced by an optimization approach in the calculation. Both assumptions might not reflect the real situation. Depending on the subject's body composition, it might not always be possible to palpate the correct landmarks for marker placement, resulting in errors for e.g. calculating joint center locations and continuatively joint moments. Additionally, skin movement in relation to the underlying bone is not constant for each segment of interest as well as for each subject. For example the thigh wobbling mass might have an increased impact compared to the shank wobbling mass, furthermore, this might be different in lean compared to obese subjects. In the understanding of obese peoples' locomotion and skeletal loading situation an improvement of methods in order to decrease this error is of practical importance and further research should focus on this aspect.

- The prevalence of adult and childhood obesity continues to increase in most countries of the world, hence the understanding of obese subjects' musculo-skeletal loading and its effect on long-term implications for musculo-skeletal health will provide important information for clinicians and physiologists. Future research of the obese locomotion and possibilities to reduce joint loading by weight loss or by specific training programs is of practical importance.

- The use of functional bracing in the rehabilitation of ACL ruptured and ACL reconstructed knees is discussed controversially. On the one hand the general need for braces is questioned, since studies conducted on tibial translation and functional performance of patients with ACL ruptured or reconstructed knees could not consistently show improved results for brace use. On the other hand subjects report a subjective improvement in stability. Additionally, limited studies are available on the effect different concepts of brace types focusing on either mechanical stabilization or neuromuscular stimulation might impose on the ACL injured knee. This opens a wide field of research questions, a few named here:

  (a) The used test methods might not have addressed the appropriate parameters (e.g ground reaction force for gait and running), or might not have been reliable enough to detect the supportive effect of functional bracing as was shown for some tests for dynamic stability (Duvinage, 2010; unpublished data). These aspects need to be addressed in future research. (b) Given the short-term effect functional sleeve braces already reveal on ACL deficient subjects (when only worn once), it is of great interest to investigate the effect induced by a functional sleeve brace on thigh atrophy, stability, muscle force and daily living tasks when it is used in the long-term of a rehabilitation process. (c) Research on functional braces was conducted using low loading tasks. However, functional braces also should support the athletes during dynamic situations and when returning to their sportive activity. Hence further studies on activities resembling this demand need to be established. The knowledge that will be gained in addressing these questions could help to improve the rehabilitation concept of ACL ruptured and reconstructed knees as well as to help the manufacturers to improve functional ortheses continuously.

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7 Appendix

7.1 Abstract ECSS 2008: Influence of body weight on joint loading in stair climbing

Strutzenberger, G.; Schneider, M. & Schwameder, H. (2008). Influence of body weight on joint loading in stair climbing. In: Cabri. J., Alves, F., Araújo, D., Barreiros, J., Diniz, J., & A. Veloso (2008) *Book of Abstracts of the 13th annual Congress of the European College of Sport Science – 9-12 July 2008 Estoril – Protugal.* p.506

Introduction: In Germany 15% of the children between 3-17 years are overweight and 6.3% are recorded as obese (Kurth & Schaffenrath Rosario, 2007). Excessive weight leads to numerous health problems not only for the metabolic system but also for the locomotor system (Wabitsch et al., 2005). Obesity is a known risk factor for osteoarthritis but the specific mechanisms are still not well understood. Research on joint loading of overweight children has been performed in level walking and standing, but little research has been done in other daily living activities (e.g. Morrison et al. 2007, Nantel et al, 2006). Therefore, the purpose of this study was to examine whether or not obese children exhibit altered patterns of joint loading in the lower extremities during stair climbing, compared with children of normal weight.

Methods: 5 normal weight children (11.4 ± 0.9 yrs, 41.6 ± 7.2 kg) and 4 obese children (age: 12.1 ±0.8 years; weight: 71.6 ± 14.7 kg) participated at this study. A staircase with 6 steps (17 x 28 cm per step) was built. Two force plates (AMTI, 1000 Hz) were embedded in the 3rd and 4th step. The kinematic data was collected by 10 infrared cameras (Vicon, 200 Hz). The children performed 3 valid trials walking upstairs and downstairs at 3 different speed levels: 90, 110,130 steps/min. The most representative out of 3 valid trials was taken for further analysis. Dynamic data was normalized to bodyweight and both kinematic and dynamic data was time-normalized to the stance phase. Inverse dynamics was calculated and peak values of the ankle, knee and hip joint moments were identified. Independent t-tests were used to examine differences between the two groups within each speed-condition.

Results: Obese children show more pronounced hip flexor moments, especially in the downstairs condition. Analogous to level walking (Nantel et al., 2006) the change between extensor and flexor moment occurs earlier in the gait cycle in obese children than in their normal-weight counterparts. Obese children reveal a significantly higher maximal varus-moment in the knee when walking downstairs. The obese children pronounce a tendency to higher plantarflexor moments and higher pronation moments in walking upstairs.

Conclusion: In obese children a clear trend to higher lower extremity joint loading in stair climbing can be observed. In combination with misalignments of the joints this might cause local overloading within the joints. Furthermore, this might be related to a greater occurrence of lower extremity joint problems and osteoarthritis, which has to be proven in long-term studies.

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7.2 Abstract ISBS 2009: Influence of body weight on joint loading in stair climbing

Strutzenberger, G., Richter, A., Lang, D. & Schwameder, H. (2009). Influence of body weight on joint loading in stair climbing. In D. Harrison, R. Anderson & I. Kenny (eds.). *Scientific Proceedings of the 27th International Conference on Biomechanics in Sports, Limerick*, p Limerick: University of Limerick.

KEY WORDS: joint loading, stair climbing, obesity

INTRODUCTION: Exercise is an essential treatment in childhood obesity. Due to the low impact on joint loading exercise recommendations are aerobic exercise such as swimming, cycling and walking (Hassink et al, 2008). Little is known though about the effect of adiposity on the function of the locomotor system (Wearing et al., 2006). Only limited research has been done on obese gait in children (Nantel et al. 2006) and even less is known about other weight bearing tasks such as climbing stairs. Therefore, the aim of this study was to examine the influences of obesity on the load pattern of the lower extremity joints of obese children while ascending and descending stairs.

METHODS: 17 normal weight children (10.4 ± 1.3 yrs, 143 ± 9 cm, 36.7 ± 7.5 kg) and 18 obese children (10.5 ± 1.5 yrs, 148 ± 10 cm, 56.6 ± 8.39 kg) participated in this study. A staircase with 6 steps (17cm x 28 cm per step) was built. Two force plates (AMTI, 1000 Hz) were embedded in the 3rd and 4th step. The kinematic data was collected using 10 infrared cameras (Vicon, 200 Hz). The children performed 3 valid trials walking up- and downstairs with a given speed of 110 steps/min. Dynamic data was normalized to body weight and time-normalized to stance phase. Inverse dynamics were calculated and mean peak values of ankle, knee and hip joint moments were identified. Independent t-tests were used to check for differences between the two groups.

RESULTS: The analysis of this study is still in progress. First results of 9 subjects (5 normal weight, 4 obese) can be reported (Table 1). Due to the low number of subjects no statistical analysis was performed. The transverse plane shows slightly higher peak moments in all joints. Additionally changes of the load pattern in the hip and knee while descending appeared in that plane.

Table 1 Mean peak moments of the hip, knee and ankle in sagittal and transverse plane.

	Hip M_{flex}	Hip M_{add}	Knee M_{flex}	Knee M_{varus}	Ankle $M_{dorsalext}$	Ankle M_{pron}
upstairs: max obese (Nm/BW)	0.81±0.24	0.61±0.15	0.85±0.24	0.50±0.15	1.61±0.32	-0.29±0.12
upstairs: max normal weight (Nm/BW)	0.93±0.21	0.49±0.04	0.97±0.21	0.42±0.04	1.38±0.15	-0.21±0.08
downstairs: max obese (Nm/BW)	0.32±0.24	1.01±0.20	0.96±0.22	0.69±0.18	1.40±0.10	-0.26±0.08
downstairs: max normal weight (Nm/BW)	0.60±0.39	0.98±0.12	0.93±0.30	0.61±0.10	1.67±0.26	-0.21±0.07

DISCUSSION: The differences of joint loading parameters between the two groups are small, but should not be neglected considering the higher body weight of the obese group. Therefore, weight bearing tasks challenge the obese musculoskeletal system, and could overload it when done too excessively. Exercise and sport performed by obese children should hence focus on training in load reduced conditions.

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Wearing, S.C., et al. (2006). The biomechanics of restricted movement in adult obesity. *Obesity Rev.* 7,13-24.

7.3 Abstract ISBS 2010: Joint loading at different variations of squats

Strutzenberger, G., Simonidis, C., Krafft, F., Mayer, D. & Schwameder, H. (2010). Joint loading at different variations in squat. In: *Scientific Proceedings of the 28th International Conference on Biomechanics in Sports, Marquette*

The purpose of this study was to identify the effect of squatting in a common, in a knee-shifted position and in an inclined position (3 cm heel lift) on joint loading. 16 male subjects were tested during squatting with an additional mass of 20 kg. Kinematic and kinetic recordings were performed by two force platforms (AMTI) and a ten infrared camera system (VICON). Inverse dynamics were calculated using a recursive multibody algorithm. Results showed significantly higher ankle dorsiflexion moments as well as higher knee varus moments for the knee-shifted performance. Due to the higher load on the ankle and the knee joint the knee-shifted variation should be avoided in squat training. The inclination of 3 cm does not lead to alterations of the joint moments and therefore does not lead to beneficial effects with respect to joint loading.

KEY WORDS: squats, joint loading, weight training.

INTRODUCTION: Due to the high biomechanical and neuromuscular similarity to sportive movements, such as running and jumping, and to daily living tasks, such as walking, getting up from a chair or step up or down stairs (Flanagan et al., 2003), squats are commonly used as exercise in general fitness training and in rehabilitation programs (Escamilla et al., 1998; Gullet et al., 2008). Also for elderly people the exercise is recommended to maintain their functional ability and, hence, help to provide their physical independence (Flanagan et al., 2003; Salem et al., 2003). Another aspect is the training of young athletes, at which muscle training always has to be considered with respect to functional and adequate joint loading. Squats can be conducted in many different ways. Variations include e.g. different squatting angles (Cotter et al., 2009; Salem & Powers, 2001), lifting additional weight (Cotter et al., 2009) as well as foot posture and stance width (Escamilla et al., 2001). Research focusing on these variations mostly shows little effect of the variations on joint loading. Cotter et al. (2009) prove different joint loading situations while performing squats with additional weight for varying squat angles of the knee. Escamilla et al. (2001) assert that a narrow stance for squat is characterized by lower tibio-femoral compressive forces than for a wide stance. No significant effects of variations in squatting angles (Salem & Powers, 2001) and in foot posture (Escamilla et al., 2001) are observed. Depending on the training goal, different effects might be aimed. In rehabilitation or recreational training the loading on the knee joint should be reduced, while in rehabilitating the patella tendinopathy more loading on the patella tendon seems to enhance the rehabilitation outcome. Good results are therefore reported for squatting on a declined surface, which increases the strain loading in the patella tendon (Kongsgaard et al., 2006; Frohm et al., 2007). Besides the different variations, recommendations are given for the common squat not to let the knee move across the virtual vertical line of the toe to minimize knee joint loading. Escamilla et al. (2009a, 2009b) analyzed cruciate ligament forces (2009a) and the patellofemoral joint force (2009b) at a long wall squat (feet farther from the wall - knee behind vertical line of toe) and short wall squat (feet closer to the wall - knee shifted over vertical line of toes). For the long wall squat higher PCL-forces, but lower patellofemoral joint forces compared to the short wall squat are exhibited, while no research is found though to study the effect of an “incorrect” performance of the common squat in weight training. Given the effect these variations might show, the knee-shift performance also might have an impact on joint loading. In case of any effects, however, they are supposed to be in a similar range as squatting on a declined surface. Therefore, the aim of this study was to analyze joint moments of three squatting variations representing a common squat, a squat with the knee being shifted over the virtual vertical line of the toe (‘knee-shifted’) and a squat with elevated heels by positioning them on a block of 3 cm.

METHOD: 16 healthy male physically active students (25.1 ± 2.2 years, 183.0 ± 5.8 cm, 80.3 ± 7.6 kg) with no lower extremity injuries participated in this study. Kinematic and kinetic recordings were collected simultaneously by a 10 camera, three-dimensional motion analysis system (VICON, MX camera system, Oxford Metrics Ltd, UK; 200 Hz)[a] and two force platforms (AMTI, model BP600900; Advanced Mechanical Technology, Watertown, MA, 1000 Hz) embedded in the floor. Reflective markers were placed according to the Vicon Plug In Gait (PIG) markerset[a] including additional markers on the medial epicondyles of the knee and on the barrel. Subjects were instructed to perform 3 different types of squats standing in a natural position. Each foot was standing on one force platform and an additional mass of 20 kg was lifted. The three variations consisted of a common performance (“common” - knees stay behind a virtual vertical line of the toes), a knee anterior shifted performance (“knee-shift” - knee moves across the vertical line of the toes) and a performance, where the subjects position their heels on a wooden block of 3 cm (“block” - block positioned under each heel). Squats were performed to a knee angle of 90°. Tactile feedback was given by a pole, which was positioned horizontally according to the subject’s body height. 8 repetitions were performed for each condition, with each repetition taking 4 seconds and 5 min resting period between each condition.

Sagittal and frontal plane moments were calculated for the hip, knee and ankle using the recursive multibody algorithm MkdTools (Simonidis & Seemann, 2010) and a model based on Zatsiorsky / Seluyanov Parameters (de Leva, 1996). Movements were filtered with a 4 Hz Butterworth filter. Maximum joint moments were identified for each repetition. Ensemble averages of the eight trials were calculated for each parameter. Peak moments are identified in sagittal plane as flexion moments of the hip and knee and as dorsiflexion moment at the ankle. In the frontal plane peak moments are identified as hip abduction moment, knee varus moment and ankle adduction moment (Figure 1) and are presented in Table 1.

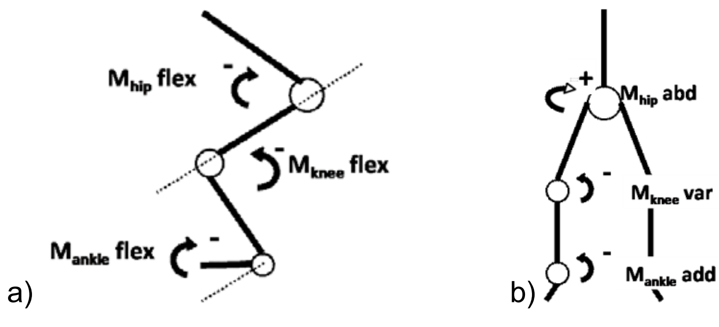

Figure 2: Peak moments of the ankle dorsiflexion moment (a), the hip adduction moment (b), the knee varus moment (c) and the ankle adduction moment (d).

Only the data of the right leg were considered for statistical analysis. The differences between the variations were statistically tested using an ANOVA with repeated measurements ($p < 0.05$).

RESULTS: Regarding the sagittal plane the 'knee-shifted' squat leads to a significant increase of the dorsiflexion moment in the ankle. This is indicated by an increase of 78% compared to the 'common' condition and an increase of 104 % compared to the 'block' condition (Figure 2.a). No other effects on the joint kinetics have been identified in the sagittal plane. In the frontal plane significant differences are observed at the knee for the 'knee-shifted' condition. Compared to the 'common' squat the knee varus moment is 127% increased and compared to the 'block' condition it is increased by 94% (Figure 2.c). The variation of the squat also leads to alterations in the ankle abduction moment, with the least abduction moment for the 'knee-shifted' squat and the highest abduction moment for the 'block' condition (Figure 2.d).

Table 1: Mean maximum moments of hip, knee and ankle joint in sagittal and frontal plane of the right leg; mean (SD)

	common [Nm/BW]	block [Nm/BW]	knee-shift [Nm/BW]
Hip flexion moment	-0.92 (0.25)	-0.90 (0.28)	-0.93 (0.26)
Knee flexion moment	-0.70 (0.23)	-0.76 (0.26)	-0.75 (0.28)
Ankle dorsiflexion moment	-0.32 (0.14)	-0.28 (0.15)	-0.57 (0.2)
Hip abduction moment	0.19 (0.08)	0.17 (0.09)	0.16 (0.09)
Knee varus moment	-0.30 (0.27)	-0.36 (0.27)	-0.69 (0.51)
Ankle adduction moment	-0.03 (0.07)	-0.08 (0.07)	0.00 (0.06)

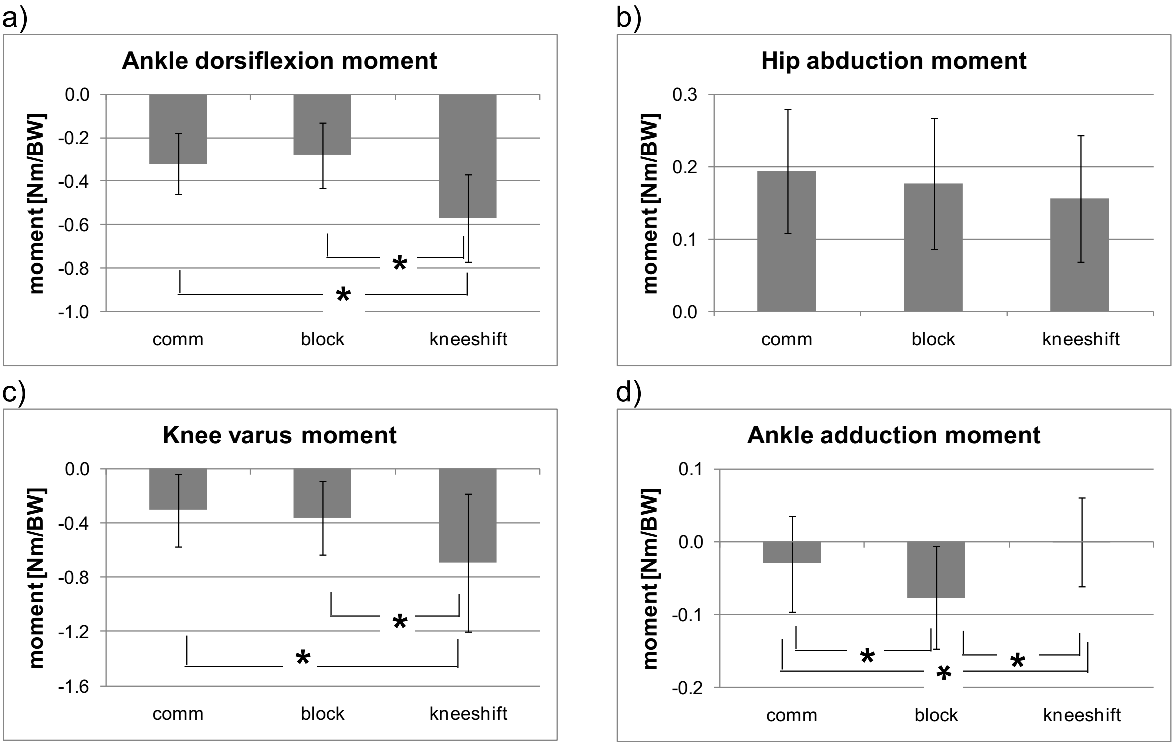

Figure 2: Peak moments of the ankle dorsiflexion moment (a), the hip adduction moment (b), the knee varus moment (c) and the ankle adduction moment (d).

DISCUSSION: The ‘knee-shifted’ condition with a relatively low additional weight does not affect the knee joint loading in the sagittal plane as one might expect. The only effect in the sagittal plane is a higher dorsiflexion moment at the ankle due to the extended anterior shift of the knee. In the frontal plane the maximum knee varus moment increases significantly. Due to the anterior movement the stabilisation of the knee might be reduced. Furthermore, this positioning leads to higher joint loading. The squat on a block does not lead to alterations in the joint kinetics in this study. Konsgaard et al. (2006) used a declined surface of 25° inclination, while the inclination at this study was ~10°. This inclination seems to be too low to lead to an alteration in the joint kinetics. At the ankle joint significant differences regarding the adduction moment are observed in each condition. Considering the relatively low values of these moments the relevance of these alterations might be neglected. Calculating the knee joint moments is the first step in understanding the forces in the knee joint. By producing a knee flexor moment a quadriceps extensor moment is also needed to hold the subject in equilibrium. Hence, a quadriceps force will be needed to generate this moment, which further has effects on single subcomponents of the knee such as tibiofemoral and patellofemoral compression forces or the forces on the anterior and posterior cruciate ligament (Escamilla, 2001). Therefore this paper only can give a first indication of the effects of variation in squatting technique. For further insight more specific knee models need to be applied to the present data.

CONCLUSION: The ‘knee-shifted’ squat does lead to higher ankle dorsiflexion moments and higher knee varus moments and should, consequently, not be recommended especially for the fitness training in juvenile and elderly athletes. No effect was found in squatting with the heel standing on a block of 3 cm height, despite a higher, but still very low adduction moments at the ankle joint. The chosen heel elevation seems to be too low to lead to signifcant alterations of joint loading. Most likely, however, the stress on the Achilles tendon might be reduced. Alterations in the ankle adduction angle at the three variations exist, but the relevance of these alterations has to be investigated more specifically.

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7.4 Abstract ÖSG 2010: Wirkung verschiedener Orthesentypen auf die passive Instabilität, die Stabilisation und die Kraft bei Patienten mit nicht operativ versorgter vorderer Kreuzbandruptur

Strutzenberger, G., Braig, M., Sell S. & Schwameder, H. (2010). Wirkung verschiedener Orthesentypen auf die passive Instabilität, die Stabilisation und die Kraft bei Patienten mit nicht operativ versorgter vorderer Kreuzbandruptur. In: Titze S, & Tilp M (eds*). Abstractband des 13. Kongress der ÖSG 2010*, Bruck an der Mur

Einleitung: Funktionelle Knieorthesen werden u.a. zur Behandlung der Kniegelenkinstabilität oder in der Heilungsphase nach Ersatz des Kreuzbandes eingesetzt. Dabei soll die Orthese die normale Gelenkkinematik nicht einschränken, muss das Gelenk aber vor unerwünschten Bewegungen schützen. Hartrahmenorthesen haben eine hohe mechanische Stabilität (Fleming et al.1999), weisen aber einen geringen Tragekomfort auf und verrutschen leicht am Bein (Berschin et al. 2003). Neben den herkömmlichen Hartrahmen-Orthesen, gibt es von der Fa. Bauerfeind mit der Orthese „SoftTec Genu“ eine Soft-Orthese mit vectororientiertem Bandagengestrick und eingebauter Gelenkschiene mit individuell einstellbaren Gelenkdrehpunkten. Diese Orthese erreicht ähnliche Stabilitätswerte wie Hartrahmen-Orthesen (Luder et al., 1998; Fleming et al., 1999), erlaubt eine propriozeptive Stimulierung (Jerosch & Prymka, 1995) und wirkt der Muskelatrophie entgegen (Reer et al., 2001). Allerdings kann durch die textile Konstruktion der Eindruck mangelnder Stabilisation bei funktionalen Bewegungen entstehen. Bisherige Studien überprüfen die Stabilitätsleistung der „SoftTec Genu“ nur in unbelasteten bzw. statisch belasteten Tests, es existieren allerdings keine Studien, die die Wirkung der „SoftTec Genu“ bei funktionalen Bewegungen prüfen. Ziel dieser Studie ist daher ein Vergleich der „SoftTec Genu“ mit einer herkömmlichen Hartrahmen-Orthese hinsichtlich der kurzfristigen Effekte u.a. auf die passive Instabilität, auf die dynamische Stabilisationsfähigkeit und auf die Kraft der Beinstreckerschlinge.

Methode: An der Studie nahmen 28 Probanden (Geschlecht: ♀: 12, ♂: 16; Alter: 41 ± 13 Jahre) mit nicht operativ versorgter ACL-Ruptur und instabilem Kniegelenk teil. Jeder Proband absolvierte vor Beginn der Studie zwei Einheiten zur Testgewöhnung und wurde mit einer Soft-Orthese („SoftTec Genu“, Fa. Bauerfeind, SoftO) und einer Hartrahmen-Orthese („4Titude“, Fa. Donjoy, HartO) versorgt. Die Probanden wurden an zwei Terminen zu ca. 90 min getestet. Als Parameter für die passive Instabilität wurde der Tibiavorschub durch den KT 1000 Test gemessen. Als Parameter für die Stabilisationsfähigkeit dienten die Pfadlänge sowie die Standardabweichung der Pfadlänge anterior-posterior und medio-lateral der drei Tests: a) Posturomed mit 2.5 cm Auslenkung (Pfadlänge: Wackelweg der Standfläche), b) Landung nach einem einbeinigen Sprung ca. 30 cm vorwärts und c) Landung nach einem einbeinigem Sprung mit ¼ Drehung (bei b) und c) Pfadlänge: Wackelweg des Center of Pressures (COP)). Die Pfadlänge der Standfläche des Posturomeds wurde mittels darauf fixierter reflektierender Marker und einem 3D Infrarot-Kamerasystem (VICON, 200 Hz) 15 s aufgenommen. Die Landungen erfolgten auf einer AMTI Kraftmessplatte (1000 Hz) bei einer Aufnahmezeit von 5 s. Es wurden jeweils fünf gültige Versuche durchgeführt. Für die Auswertung wurde der beste und schlechteste Versuch (Pfadlänge) gestrichen und der Mittelwert aus den verbleibenden drei Versuchen berechnet. Die Kraft wurde anhand der beidbeinigen Tests d) isometrische Maximalkraft (isom. F_{max}) am Kraftmessschlitten mit geteilter Kraftmessplatte (Eigenbau, BMC, 1000 Hz), und e) Counter Movement Jump (CMJ) auf geteilter Kraftmessplatte (Eigenbau, BMC, 1000 Hz) durchgeführt. Zur Analyse wurde aus drei Versuchen der beste gewählt (isom. F_{max}: max. Gesamtkraft; CMJ: max. Sprunghöhe). Die Unterschiede zwischen den Versorgungssituationen wurden mittels einer ANOVA mit Messwiederholung und Bonferronikorrektur bei einem Signifikanzniveau von p=0.05 statistisch überprüft.

Ergebnisse: Die passive Instabilität (KT 1000) wurde durch das Tragen von der Soft-Orthese um 29% (± 25%) und durch das Tragen der Hartrahmen-Orthese um 8% (± 37%) signifikant verringert. Beim CMJ zeigt sich eine signifikante Steigerung der Explosivkraft des verletzten Beines beim Tragen der Soft-Orthese um 18% (± 35%) im Vergleich zur unversorgten Situation und um 24% (± 37%) im Vergleich zur mit Hartrahmen-Orthese versorgten Situation. Kein Unterschied konnte zwischen unversorgter und mit Hartrahmen-Orthese versorgter Situation festgestellt werden (Tabelle 1). In den übrigen Parametern zeigten sich keine signifikanten Unterschiede zwischen unversorgter Situation und mit Orthese versorgter Situation (SoftO, HartO) sowie zwischen den beiden Orthesentypen.

Tabelle 1
Passive Instabilität (KT 1000) und Explosivkraft des verletzten Beins (CMJ)

Parameter	Ohne	SoftO	HartO	p: Ohne-SoftO	p: Ohne-HartO	p: SoftO- HartO
KT 1000 20 p [mm]	8.3 (3.5)	5.6 (2.0)	7.1 (3.0)	0.000	0.026	0.004
CMJ $F_{explosiv}$ verletztes Bein [Nm/BW/s]	6.11 (1.92)	7.3 (2.94)	6.09 (2.4)	0.015	1.000	0.018

Diskussion und Schlussfolgerungen: Bei den getesteten funktionellen Aufgaben haben die beiden Orthesen im Allgemeinen keinen signifikanten Einfluss auf die Leistung. Ausnahmen bestehen in der passiven Stabilität und in der Explosivkraft beim CMJ. Dies indiziert, dass obwohl in der passiven Situation beide Orthesen eine Stabilisierungsfunktion haben, diese bei leichten funktionellen Aufgaben nicht in Anspruch genommen wird. Bei komplexer Bewegung (CMJ, Explosivkraft) konnte nur durch die Orthese SofTec eine Veränderung erreicht werden. Möglicherweise ist die propriozeptive Stimulierung der Muskulatur durch die SoftOrthese (Jerosch & Prymka, 1995) dafür verantwortlich, was auch die verbesserte Stabilität beim KT 1000 Test erklären könnte.

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7.5 Abstract VSOU 2011: Wirkung verschiedener Orthesentypen auf die passive Instabilität, die Stabilisation und die Kraft bei Patienten mit nicht operativ versorgter vorderer Kreuzbandruptur

Strutzenberger G, Braig M, Sell S, Schwameder H. Wirkung verschiedener Orthesentypen auf die passive Instabilität, die Stabilisation und die Kraft bei Patienten mit nicht operativ versorgter vorderer Kreuzbandruptur. Abstractband 59. Jahrestagung der VSOU, 2010. (in press)

Fragestellung

Die Wirkung verschiedener Orthesenkonstruktionen ist bei funktionalen Bewegungen noch ungenügend geklärt. Ziel der Studie ist daher ein Vergleich einer Soft-Orthese mit einer Hartrahmen-Orthese hinsichtlich kurzfristiger Effekte auf die passive Instabilität, auf die dynamische Stabilisationsfähigkeit und auf die Kraft der Beinmuskulatur.

Methode

28 Probanden (♀: 12, ♂: 16; Alter: 41 ± 13 J.) mit nicht operativ versorgter ACL-Ruptur und instabilem Kniegelenk nahmen an der Studie teil. Sie wurden mit einer Soft-Orthese („SoftTec Genu“, Fa. Bauerfeind, SoftO) und einer Hartrahmen-Orthese („4Titude“, Fa. Donjoy, HartO) versorgt. Nach einer Habituationsphase absolvierten sie u.a. die Tests a) KT 1000, b) einbeinige Stabilisation auf b1) stabiler Standfläche - geschlossene Augen (AMTI, 1000 Hz), b2) stabiler Standfläche nach Sprung mit ¼ Drehung (AMTI, 1000 Hz), b3) labiler Standfläche mit Irritation (Posturomed und Vicon, 200 Hz) und c) Counter Movement Jump (CMJ, Kraftmessplatte Eigenbau, 1000 Hz). Die statistische Analyse wurde mittels einer ANOVA mit Messwiederholung und Bonferronikorrektur bei einem Signifikanzniveau von p=0.05 durchgeführt.

Ergebnisse

Der Tibiavorschub (KT 1000, Zug 20p) wurde durch das Tragen beider Orthesen signifikant verringert. Die HartO verringerte den Tibiavorschub um 14% (von 8.3 (±3.5) mm auf 7.1 (±3.0) mm), die SoftO um 32% (von 8.3 (±3.5) mm auf 5.6 (±2.0) mm) und erreicht damit zusätzlich signifikant geringere Werte als die HartO. Keine signifikanten Unterschiede zwischen unversorgter und versorgter Situation, sowie zwischen den beiden Orthesentypen (HartO, SoftO) konnten sowohl für die Pfadlänge, als auch die anterio-posteriore und medio-laterale Schwankung des Kraftangriffpunktes für die Tests „einbeinige Stabilisation auf stabiler Standfläche mit geschlossenen Augen“ und „einbeinige Stabilisation auf stabiler Standfläche nach Sprung mit ¼ Drehung“ gezeigt werden.

Ein anderes Bild entsteht, wenn die Probanden eine labile Standfläche nach einer Auslenkung von ca. 2.5 cm stabilisieren müssen. Zunächst besteht in der ersten Sekunde nach dem Störungsreiz kein Unterschied zwischen den verschiedenen Versorgungssituationen. Hier erfolgt die Auslenkung und das System hat noch nicht die Möglichkeit zu reagieren. Erst in den darauffolgenden vier Sekunden wird durch das Tragen der Orthesen die Pfadlänge (12% HartO, 14% SoftO), die Schwankung in anterior-posteriore Richtung (15% HartO, 21% SoftO) und die Schwankung in medio-lateraler Richtung (9.7% HartO, 10.0% SoftO) verringert. Statistisch abgesicherte Unterschiede zeigen sich allerdings nur durch das Tragen der SofTec Genu Orthese in medio-lateraler Richtung.

Beim CMJ treten keine Veränderungen in der Sprunghöhe und in der Maximalkraft der Beine auf. Jedoch zeigt sich eine signifikante Steigerung der Explosivkraft des verletzten Beines. Hier verhalten sich die beiden Orthesen unterschiedlich im Vergleich zur nichtversorgten Situation. Während durch das Tragen der SofTec Genu Orthese die Explosivkraft des verletzten Beines um 18% signifikant gesteigert werden kann (von 5.98 (±1.90) N/s/BW auf 7.02 (±3.00) N/s/BW), treten keine Veränderungen durch das Tragen der HartO ein (-2%, 5.88 (±2.42) N/s/BW). Übertragen auf eine Alltagssituation könnte das bedeuten, dass die Patienten bei ungewollten Bewegungen schneller Muskelkraft aufbringen können und somit das Kniegelenk schneller stützen können als ohne Versorgung oder auch schneller als mit der HartO-Versorgung.

Schlussfolgerung

In Übereinstimmung mit Ergebnissen von Fleming et al. (1999) und Luder et al. (1998) zeigt diese Studie, dass eine rein mechanische Funktion der Verringerung des Tibiavorschubs durch beide Orthesen erreicht wird. In dynamischer Situation bestehen bei den Tests mit geringeren Komplexitätsgrad keine signifikanten Unterschiede zwischen den verschiedenen Versorgungsituationen (ohne Versorgung, HartO, SofTO). Bei diesen Aufgaben wird eine unterstützende Wirkung durch die beiden Orthesen nicht in Anspruch genommen, beziehungsweise ist diese so gering, dass sie bei der Probandenzahl nicht nachgewiesen werden kann. Bei den beiden komplexen Bewegungsabläufen „Stabilisation nach Auslenkung“ und bei der Explosivkraft können Vorteile der SofTec Genu gegenüber der unversorgten Situation und der Versorgung mit der Orthese Donjoy festgestellt werden. Möglicherweise ist die propriozeptive Stimulierung der Muskulatur durch die SoftOrthese (Jerosch & Prymka, 1995) dafür verantwortlich, was wiederum auch die verbesserte Stabilität beim KT 1000 Test erklären könnte.

Literatur:

Fleming, B., Beynnon, B., Chruchill, D., Webster, J. & Renström, P. (1999). A biomechanical investigation of the effect of functional bracing on the ACL-deficient knee with and without weight bearing. Lecture No. 82 at ORS/AAOS Congress. Anaheim, California, USA.

Jerosch, J. & Prymka, M. (1995). Propriozeptive Fähigkeiten des gesunden Kniegelenks: Beeinflussung durch eine elastische Bandage. *Sportverletzung Sportschaden*. 9 (3), 72-76.

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Zusammenfassung

Die Sportbiomechanik befasst sich unter anderem mit den auf den menschlichen Bewegungsapparat wirkenden Kräften und den damit verbundenen Reaktionen wie Fortbewegung, Deformation der weichen und der knöchernen Strukturen, Wachstum und Entwicklung von biologischem Gewebe sowie akute und chronische Verletzungen (Nigg, 2000). In Verbindung mit dem Fortschritt in der Entwicklung von Messinstrumentarien und in der Computertechnologie sind neue Möglichkeiten entstanden, die menschliche Bewegung zu erfassen sowie die Erweiterung der kinetischen Analyse von einer rein sportlichen Anwendung bis hin zu der Analyse von klinischen Problemstellungen (Andriacchi & Alexander, 2000; Gollhofer & Müller, 2009) auszuweiten.

Die nicht direkt sichtbaren intersegmentären Kräfte und Momente werden üblicherweise durch mathematische Annäherungen bestimmt. Dazu dienen als Eingangsvariablen für die Berechnung kinematischer und kinetischer Parameter die externen Bodenreaktionskräfte sowie die dreidimensionale Position der Segmente, die durch das Erfassen von auf der Haut angebrachten Markern gemessen wird (Andriacchi & Alexander, 2000). Gegenwärtig existieren mehrere Modellierungsmethoden, die versuchen sich diesen Parametern zu nähern. Die Genauigkeit dieser Annäherungen hängt allerdings bedeutend von den Messinstrumentarien, den getroffenen Annahmen und der Berechnungsmethode selbst ab (Cappozzo et al., 2005).

Diese Arbeit möchte einen Einblick in das breite Spektrum der Anwendungsmöglichkeiten der Biomechanik geben. Realisiert wird das zum einen anhand eines methodisch orientierten Teils in Kapitel 2, indem die Problematik von der Vergleichbarkeit verschiedener Modellierungsmethoden behandelt wird, und zum anderen anhand eines anwendungsorientierten Teils in Kapitel 3 bis 5, in dem der Schwerpunkt auf Prävention und Rehabilitation gelegt ist. Dieser Teil behandelt drei voneinander unabhängige Studien. Jede Studie untersucht eine für sich freistehende Forschungsfrage und es finden jeweils unterschiedliche Methoden Anwendung. Um das breite Spektrum der biomechanischen Anwendungsmöglichkeiten zu umspannen, befassen sich die Studien mit den drei Probandengruppen „gesunde Probanden“, „Probanden mit Risiko für Überlastungsverletzungen“ und „verletzte Probanden“. Jedes Kapitel gibt neue Einblicke in die menschliche Bewegung und erweitert das Wissen, zu dem Sportbiomechaniker beitragen können, um adäquate Trainingsprogramme zu erstellen, Verletzungsmechanismen zu verstehen und Rehabilitationskonzepte zu verbessern. Kapitel 3 analysiert bei „gesunden Probanden“ die Belastung der unteren Extremität bei der Ausführung von Variationen der Kniebeuge. Kapitel 4 befasst sich mit der Belastung der unteren Extremität von „Probanden mit Risiko für Überlastungsverletzungen“, indem es das Treppensteigen von adipösen und normalgewichtigen Kindern analysiert. Der Effekt von verschiedenen Orthesentypen auf funktionelle Tests und auf die Kniegelenkinstabilität von Probanden mit gerissenem vorderem Kreuzband als „verletzte Probanden“ wird in Kapitel 5 dargestellt. Im Folgenden wird eine eigenständige Zusammenfassung für jedes Kapitel gegeben.

Studie I: Der Effekt drei verschiedener Berechnungsmodelle auf die Kinematik und Dynamik der Kniebeuge

Es existiert eine Vielzahl an Modellen zur Analyse menschlicher Bewegung. Die Modelle unterscheiden sich unter anderem in Aspekten der Markerposition, der gemessenen Variablen, der für die Gelenke freigegebenen Freiheitsgrade sowie in den anatomischen und technischen Bezugnahmen, der Gelenkrotationsgrundsätze und ihrer Begriffsdefinition. Ein Vergleich der verschiedenen Modelle fällt daher schwer. Das Ziel dieser Studie war es, in einem sportrelevanten Bezug den Effekt von drei gegenwärtig verwendeten Modellen auf die Kinematik und Dynamik zu bestimmen. Fünfzehn männliche Probanden führten eine Kniebeuge mit einem Zusatzgewicht von 20 kg aus. Kinematische und dynamische Daten wurden anhand eines Infrarot-Kamerasystems (VICON, 200 Hz) und anhand von zwei Kraftmessplatten (AMTI, 1000 Hz) aufgenommen. Die vordefinierten Parameter wurden mit (a) dem Vicon Plug-in Gait (PiG) Modell, (b) einem rekursiven Mehrkörper-Algorithmus (MKDtools) und (c) dem Visual3D (V3D) Modell berechnet. Die Ergebnisse zeigten signifikante kinematische Unterschiede in sagittaler, frontaler und transversaler Ebene für das Hüft-, Knie- und Sprunggelenk. Die Ergebnisse der dynamischen Parameter zeigten signifikante Unterschiede in sagittaler und transversaler Ebene für Hüft- und Kniegelenkmomente. Die Resultate zeigten deutlich, dass verschiedene Modelle zu Unterschieden in kinematischen und dynamischen Ergebnissen führen. Außerdem war es weder möglich eine Systematik in den Unterschieden zwischen den Modellen zu identifizieren, noch eine Aussage treffen zu können, dass zwei Modelle beständig zu den gleichen Resultaten führen würden. Diese Studie verleiht der Notwendigkeit zur Vorsicht Nachdruck, wenn Daten von Studien, die verschiedene Analysemethoden verwenden mit einander verglichen werden. Besonders für den Vergleich und die Interpretation von Querschnittsstudien, müssen die Spezifikationen von den verwendeten Modellen angegeben werden, um ein gutes Verständnis der Daten zu ermöglichen.

Studie II: Die Belastung der unteren Extremität bei drei unterschiedlichen Kniebeuge-Variationen

Kniebeugen sind häufig Bestandteil von einem allgemeinen Fitnesstraining sowie von Rehabilitationsprogrammen und werden in vielen Variationen durchgeführt. In welchem Ausmaß diese Variationen zu unterschiedlichen Belastungen der unteren Extremität führen ist allerdings noch unbekannt. Das Ziel dieser Studie war es die Kinematik und Dynamik der unteren Extremität von drei Kniebeugevariationen zu vergleichen. 16 Probanden führten mit einem Zusatzgewicht von 20 kg eine Standardkniebeuge („Standard“), eine Kniebeuge, in der das Knie über die vertikale Linie der Zehe geführt wurde („Knie vorgeführt“) und eine Kniebeuge auf einer geneigten Standfläche durch („geneigt“). Kinematische und dynamische Daten wurden anhand eines Infrarot-Kamerasystems (VICON, 200 Hz) und anhand zweier Kraftmessplatten (AMTI, 1000 Hz) aufgenommen. Ein rekursiver Mehrkörper Algorithmus (MKDtools) wurde für die Berechnung der Kinematik und der inversen Dynamik verwendet. Bei der „Knie vorgeführten“ Kniebeuge erhöhte sich im Vergleich zur der „Standard“ Kniebeuge das sagittale Sprunggelenkmoment signifikant um 360% (gesamt $p \leq 0.001$) und um 587% (gesamt $p \leq 0.001$) im Vergleich zu der „geneigten“ Kniebeuge. Es wurden keine signifikanten Unterschiede zwischen den Variationen „Standard“ und „geneigt“ festgestellt. Die „Knie vorgeführte“ Kniebeuge kann dann empfohlen werden, wenn eine Steigerung der

Plantarflexoren-Muskulatur - ohne dabei das Kniegelenk- und das Hüftgelenkmoment zu erhöhen - erwünscht ist. Die Kniebeuge auf einer 10° geneigten Ebene führte zu keiner Veränderung der Gelenkmomente im Vergleich zur „Standard“ Kniebeuge und führt darum, in Bezug auf die Gelenkbelastung, auch nicht zu günstigen Effekten

Studie III: Bewegungsanpassungen von adipösen Kindern beim Treppensteigen: Zeitlich-räumliche, kinematische und dynamische Unterschiede zu normal-gewichtigen Kindern

Mechanische Faktoren und das Körpergewicht sind bedeutende Faktoren in der Entwicklung von varus/valgus Knieachsenveränderungen und führen bei einem Missverhältnis zu gesundheitlichen Langzeitauswirkungen, darunter ein erhöhtes Risiko für Osteoarthrose. Bei Kindern ist allerdings noch sehr wenig über die Wechselbeziehung dieser Faktoren während dynamischer Aktivitäten bekannt. Das Ziel dieser Studie war die Überprüfung der Hypothese, dass die gewichtsnormalisierte Gelenkbelastung der unteren Extremität während des Treppensteigens von adipösen Kindern größer ist als die von normalgewichtigen Kindern. 18 adipöse Kinder und 17 normalgewichtige Kinder nahmen an der Studie teil. Ein Vicon Kamerasystem (200 Hz) und zwei AMTI Kraftmessplatten (1000 Hz) wurden verwendet, um die kinematischen und dynamischen Daten beim treppauf und treppab Gehen aufzunehmen und zu analysieren. Signifikante Unterschiede zwischen adipösen und normalgewichtigen Kindern wurden in den räumlich-zeitlichen, kinematischen und dynamischen Parametern während des treppauf und treppab Gehens deutlich. Beim treppauf Gehen wurden signifikant höhere Hüftabduktionsmomente (+23%; p=0.001) und signifikant höhere Knieextensionsmomente (+20%; p=0.008) für adipöse Kinder beobachtet. Beim treppab Gehen zeigten sich für die adipösen Kinder signifikant niedrigere Hüftextensionsmomente (-52%; p=0.031), signifikant höhere Hüftflexionsmomente (+25%; p=0.016) und signifikant höhere Knieextensionsmomente (+15%; p=0.008). Bis heute ist es noch unklar, ob und wie sich der Bewegungsapparat adipöser Kinder an die größere Gelenkbelastung anpasst. Trotzdem können diese Unterschiede in den Gelenkmomenten zu einer über das Jugendalter bis hin zum Erwachsenenalter stetig steigenden Gelenküberlastung führen, und möglicherweise in einem erhöhten Risiko für die Entwicklung von Knie- und Hüftgelenkarthrose resultieren.

Studie IV: Die Wirkung von Orthesentypen auf die Kniegelenkinstabilität und auf funktionelle Tests bei Patienten mit vorderer Kreuzbandruptur

In der Rehabilitation von Verletzungen des vorderen Kreuzbands (VKB) ist es der medizinischen Forschung noch nicht gelungen einheitlich günstige Effekte auf die funktionelle Kniegelenkstabilität durch den Gebrauch von Funktionsorthesen nachzuweisen. Die Verwendung von Hartrahmenorthesen kann scheinbar zu erhöhter Muskelatrophie und zu einer Veränderung das Bewegungsmuster führen, ohne dabei die Stabilität unter gewichtstragenden Situationen zu verbessern. Im Gegensatz dazu könnten Softorthesen durch eine verbesserte propriozeptive Stimulierung möglicherweise einen günstigeren Nutzen für die Rehabilitation bieten. Allerdings ist über die Effekte einer Softorthese, wenn sie von Probanden mit Verletzungen des VKB getragen wird, noch wenig bekannt. Daher war das Ziel dieser Studie den Effekt von verschiedenen Orthesentypen im Vergleich zu einer nicht mit Orthesen versorgten Situation im Hinblick auf die Kniegelenkstabilität und auf die funktionellen Fähigkeiten bei Patienten mit VKB-Ruptur zu untersuchen. Es wurden 28 Probanden, die eine nichtversorgte VKB-Ruptur hatten, eine Softorthese (SofTec Genu, Bauerfeind) und eine Hartrahmenorthese (4Titude Donjoy, Ormed) angepasst. Daten wurden von der verletzten Seite in nichtversorgter Situation und in den beiden mit Orthesen versorgten Situationen für Tests zur Kniegelenkstabilität, zum Gelenkpositionssinn, zum statischen und dynamischen Gleichgewicht, zur isometrischen und dynamischen Kraftfähigkeit der Beinstreckerschlinge sowie zu den täglichen Aufgaben Gehen und Laufen aufgenommen. Die Ergebnisse zeigten für die Softorthese eine signifikante Reduktion der Kniegelenkinstabilität (-32%; $p<0.001$), eine signifikante Minderung in der Standardabweichung der medio-lateralen Standflächenschwankung nach einem Auslenkungsreiz (-10%; $p=0.016$) und eine signifikante Steigerung der maximalen Kraftanstiegsrate (+18%; $p=0.015$). Bei der Hartrahmenorthese wurden keine signifikanten Veränderungen im Vergleich zur nichtversorgten Situation festgestellt. Es ist möglich, dass die Softorthese die Propriozeption positiv beeinflusst und so zu günstigeren Leistungen bei Aufgaben zur Gelenkstabilität, zum Gleichgewicht und zur Kraft führt. Allerdings wurden diese Effekte nur in komplexen Situationen beobachtet. Softorthesen werden möglicherweise nicht bei der Ausführung von täglichen Aufgaben benötigt, aber – im Vergleich zur Hartrahmenorthese – könnten sie eine effektive und günstige Unterstützung bei dynamischen Aktivitäten bieten.

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Eidesstattliche Erklärung

Hiermit erkläre ich, dass ich die vorliegende Dissertation mit dem Thema

"Kinematic and kinetic analyses of human movement with respect to health, injury prevention and rehabilitation aspects"

selbständig verfasst, nur die angegebenen Hilfsmittel benutzt, wörtlich oder inhaltlich übernommene Stellen als solche gekennzeichnet und die Satzung des Karlsruher Institut für Technologie (KIT) [ehemals Satzung der Universität Karlsruhe (TH)] zur Sicherung guter wissenschaftlicher Praxis beachtet habe. Des Weiteren wurde diese Arbeit nicht bereits anderweitig als Prüfungsarbeit verwendet.

Gerda Strutzenberger
Karlsruhe, den 10.01.2011

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