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Development of a Gait Analysis Model and its Clinical Relevance for the Treatment of Patients With Varus Malalignment of the Knee

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1 Introduction

Lower limb malalignment in the frontal plane has been clearly identified as a risk factor for the progression of osteoarthritis (OA) of the knee (Cicuttini, Wluka, Hankin, & Wang, 2004; Sharma et al., 2001; Tetsworth & Paley, 1994). Specifically, excessive varus malalignment is associated with higher than normal medial compartment knee joint loading and the prevalence of OA (McNicholas et al., 2000; Miyazaki et al., 2002; Sharma et al., 1998).

Typically, lower limb alignment is measured statically from radiographs without representing aspects of the joint loading characteristics during dynamic situations. It has been suggested that relying solely on the current clinical practice of assessing alignment statically may be inappropriate (Andriacchi, Lang, Alexander, & Hurwitz, 2000). As a result, the use of quantitative gait analysis as an adjunct to static radiographic measures of alignment has been investigated as a means in the study and treatment of knee OA (Hunt, Birmingham, Jenkyn, Giffin, & Jones, 2008; Mündermann, Dyrby, & Andriacchi, 2008). The external knee adduction moment is an often-used predictor of knee joint loading (Hurwitz, Ryals, Case, Block, & Andriacchi, 2002) and a commonly used outcome measurement reported from gait analysis data in adults with knee OA. Dynamic loading characteristics during gait may significantly influence prognosis of disease progression and the static mechanical axis alignment and joint space width do not reflect biomechanical loading on the diseased medial compartment as strongly as the adduction moment does (Miyazaki et al., 2002).

The role of dynamic factors in the pathogenesis of knee OA has been well documented in the literature. Most studies investigating the gait of adult patients with established knee OA have focused on kinematics and kinetics of the knee (Kaufman, Hughes, Morrey, B. F., Morrey, M., & An, 2001), and the sagittal or frontal plane (Al-Zahrani & Bakheit, 2002; Baliunas et al., 2002; Childs, Sparto, Fitzgerald, Bizzini, & Irrgang, 2004; Gok, Ergin, & Yavuzer, 2002). However, there is a lack of research on gait data in the transverse plane (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007) and sparse attention has been paid to the changes in the mechanical environment of other joints of the affected limb, which presumably occur concomitantly with changes in knee-joint mechanics. Furthermore, the author of this thesis is unaware of previous three-dimensional gait analysis studies focused on the dynamic loading characteristics of the knee and hip joints as well as potential compensatory mechanisms in children and adolescents with pathological varus alignment of the knee but no signs of knee OA.

A precondition to perform gait analysis on patients with varus malalignment is the application of a useful model defining marker positioning and the calculation of skeletal motion. The standard Plug-in-Gait (PiG) model used by a vast majority of clinical gait laboratories is prone to errors arising from inconsistent anatomical landmark identification and knee axis malalignment (Leardini, Chiari, Della Croce, & Cappozzo, 2005; Piazza & Cavanagh, 2000).

Consequently, this thesis comprises two studies:

1. Development and evaluation of a lower body model for clinical gait analysis (Section 6).

   In this section, the relevance of the development of a lower body model is demonstrated. In addition to the detailed characterization of the model, it will be evaluated and compared with the standard PIG model within a study population of 25 subjects.

Parts of this study were accepted for publication in the Journal of Applied Biomechanics on November 30th 2011 (Stief, Böhm, Michel, Schwirtz, & Döderlein, 2011).

2. Application of the lower body model for gait analysis in children and adolescents with varus malalignment of the knee (Section 7).

   In this section, three-dimensional knee and hip joint angles and moments are investigated in 14 patients and compared to 15 healthy control subjects. Moreover, this section shows if potential mechanisms of gait compensation in the present patient group are different with those reported in adult patients with established medial knee OA.

   Parts of this study were published in Gait & Posture (Stief, Böhm, Schwirtz, Dussa, & Döderlein, 2011).

Before the characterization and realization of these two experimental studies in Section 6 and 7, the thesis starts with clinical basics of leg alignment such as the physiological development of the mechanical axis during growth and the illustration of different options for the treatment of pathological varus alignment of the leg (Section 2). After a short visualization of fundamentals of clinical gait analysis in Section 3, the current state of research regarding gait analysis studies in patients with knee OA and varus malalignment is shown in Section 4. This section is used as transition to the concrete aims of this thesis (Section 5). The thesis ends with the identification of the clinical significance of the results (Section 8), a case study (Section 9), an overview of limitations of the present study design (Section 10) and an outlook on future studies in this field of research (Section 11).

2 Clinical Basics of Leg Alignment

This section gives a fundamental clinical knowledge of leg alignment and the resultant knee OA. It contains the physiological development of the mechanical axis during growth, the relationship between pathological leg alignment in the frontal plane and knee OA and possible methods of treatment of pathological varus alignment of the knee and knee OA, respectively.

2.1 Physiological Development of the Mechanical Axis of the Leg During Growth

At birth the lower extremity is characterized by a varus axis in the knee joint at an average of 15°. A varus position of the knee is still noted soon after the child begins to stand (Salenius & Vankka, 1975; Schmidt & Yngve, 1986; Sherman, 1990). For the next two years, the knee continuously moves medially until it reaches a slightly exaggerated valgus position. Then, after reversing its direction, it travels slowly in a varus direction for two more years. From age 5 years on, it comes to rest at a point at which the tibia is vertical and its proximal surface is nearly horizontal (Kling, 1987; Salenius & Vankka, 1975). At an age between 8 and 10 years a normal valgus axis of 5-7° has developed in the lower limb (Westhoff, Jäger, & Krauspe, 2007).

As it is an important question in clinical orthopedic surgery whether to correct extreme varus or valgus knees, Salenius and Vankka (1975) graphically show the physiological development of the tibiofemoral angle during growth based on 1480 measurements of 1279 patients (Figure 1). The tibio-femoral angle was measured on the roentgenogram by drawing a longitudinal axis mid-way between the femoral and tibial diaphyseal cortices. The angle between these two longitudinal lines was measured in degrees.

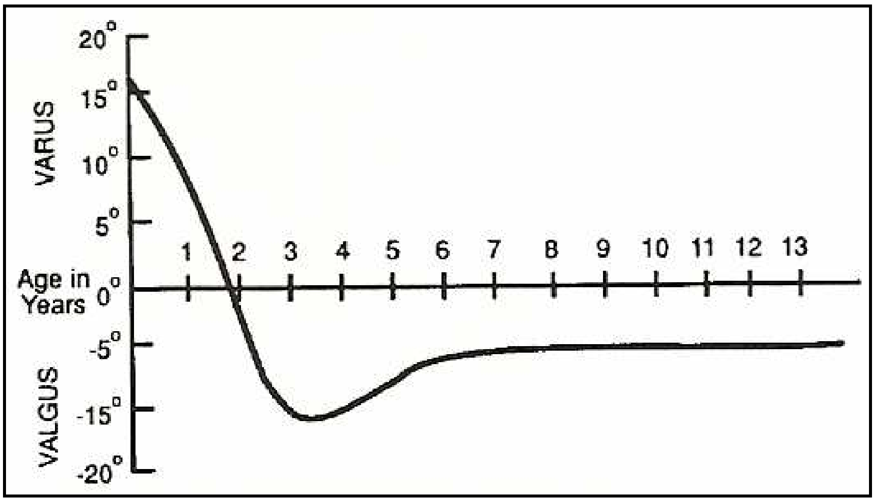


Figure 1. The development of the tibiofemoral angle (anatomic axis) in children during growth. The mean of 1480 measurements of 1279 patients are presented. Adapted from "The development of the tibiofemoral angle in children," by P. Salenius and E. Vankka, 1975, *The Journal of Bone and Joint Surgery (American volume), 57*, p. 260.

The development of the tibiofemoral angle was similar in boys and girls (Salenius & Vankka, 1975) and is radiographically shown in a representative patient in Figure 2.

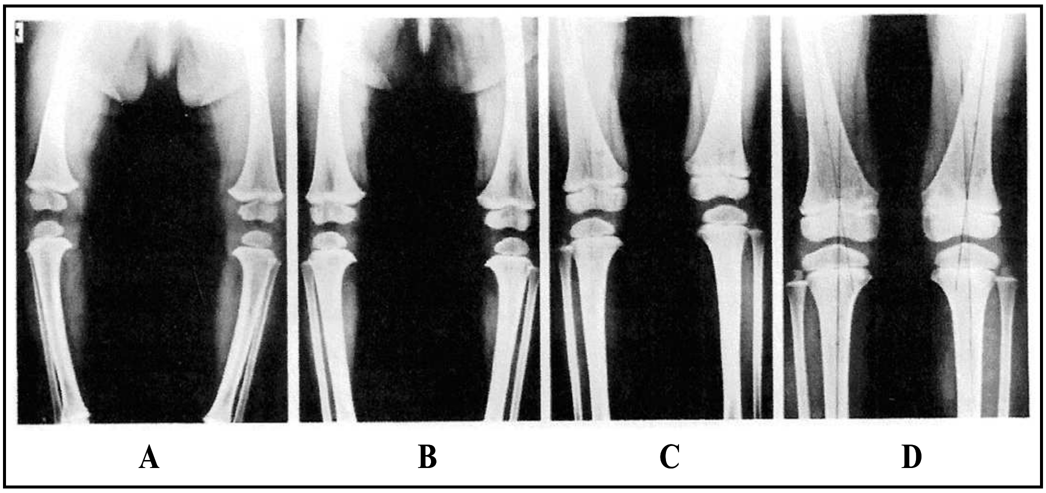


Figure 2. The radiographic development of the tibiofemoral angle in a child. A: The tibiofemoral angle in a child one year and two month old. The child has not yet learned to walk. The angle is in 21 degrees of varus on the right and 28 degrees of varus on the left. B: The tibiofemoral angle of the child six months later. The child has been walking for a few months. The angle is in 13 degrees of varus on both sides. C: The child has been walking for more than a year. The tibiofemoral angle is in 12 degrees of valgus on the right and 13 degrees on the left. The child is three years old. D: The child is now five years old. In the right knee the valgus angle is 11 degrees and in the left, 12 degrees. Adapted from "The development of the tibio-femoral angle in children," by P. Salenius and E. Vankka, 1975, *The Journal of Bone and Joint Surgery (American volume), 57*, p. 261

In accordance with MacMahon, Carmines, and Irani (1995), this motion of the lower femur and the upper tibia resembles the behavior of a pendulum, which comes to a stop in an upright position (Figure 3).

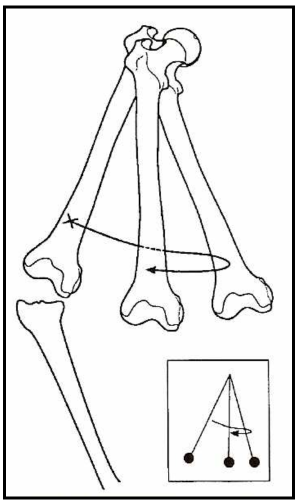


Figure 3. Varus to valgus to neutral motion of the femur. Adapted from "Physiologic Bowing in Children: An Analysis of the Pendulum Mechanism," by E. B. MacMahon, D. V. Carmines, and R. N. Irani, 1995, *Journal of Pediatric Orthopaedics Part B, 4*, p. 101.

The evolution of the axis values in the lower extremity is influenced by changing compressive and propelling forces acting on the growth plates as the child adopts an upright posture. Therefore, gravitational forces and their bending moments must be involved (MacMahon et al., 1995). In particular, physiologic bowing is a yardstick of the behavior of a physis and metaphysis to normal loading and their time-dependent response to these forces. Conditions outside these limits showing in Figure 1 must be considered pathologic (MacMahon et al., 1995).

2.2 Leg Malalignment in the Frontal Plane as Biomechanical Risk Factor for the Onset and Progression of Knee Osteoarthritis

OA is a degenerative joint disease that affects an increasing proportion of the population (Felson et al., 1987; Fife, Klippel, Weyand, & Wortmann, 1997; Peyron et al., 1993) and one of the major causes of pain and physical disability (Felson et al., 1987). The probability of developing knee OA in people aged over 65 years is approximately 30% (Felson et al., 1987) and by age 85 nearly one in two (Murphy et al., 2008).

Although most joints of the lower extremity, including the ankle and hip, may be involved, the knee is the most common site for OA (Oliveria, Felson, Reed, Cirillo, & Walker, 1995). It can occur in either compartment of the tibiofemoral joint, but is most common in the medial compartment (Dearborn, Eakin, & Skinne, 1996). Moreover, loads transferred through the medial compartment during walking are substantially higher than loads transferred through the lateral compartment (Schipplein & Andriacchi, 1991). The major clinical features of OA are pain, restricted joint motion and deformity leading to a decline in physical function (Guccione et al., 1994).

The pathogenesis of OA is poorly understood and the treatment is currently limited to the management of symptoms rather than reducing disease progression and in more severe and late-stage disease, a knee replacement may be required. Therefore, prevention of knee OA should be one of the major aims of health care, and requires clear knowledge of the risk factors of this disease. There is increasing interest in the contribution of biomechanical variables to the pathogenesis and management of OA in addition to biologic factors. The term "biomechanics" refers to the forces acting upon and within biological structures, including the effects produced by these forces (Hay, 1993).

Changes in lower limb alignment can redistribute the medial and/or lateral loads at the knee joint, resulting in altered joint moments and thus articular loads. While a varus knee alignment is argued to increase medial tibio-femoral compartment load, a valgus alignment is believed to increase lateral compartment load (Sharma, Lou, Cahue, & Dunlop, 2000). Anatomically, a varus angulation compresses medial structures such as the medial tibio-femoral compartment, and distracts lateral structures such as the lateral collateral ligament. Therefore, lower limb malalignment in the frontal plane has been clearly identified as a risk factor for the progression of OA of the knee (Cicuttini et al., 2004; Sharma et al., 2001; Tetsworth & Paley, 1994). The relationship between moments in the knee and the internal joint load distribution is shown in Figure 4.

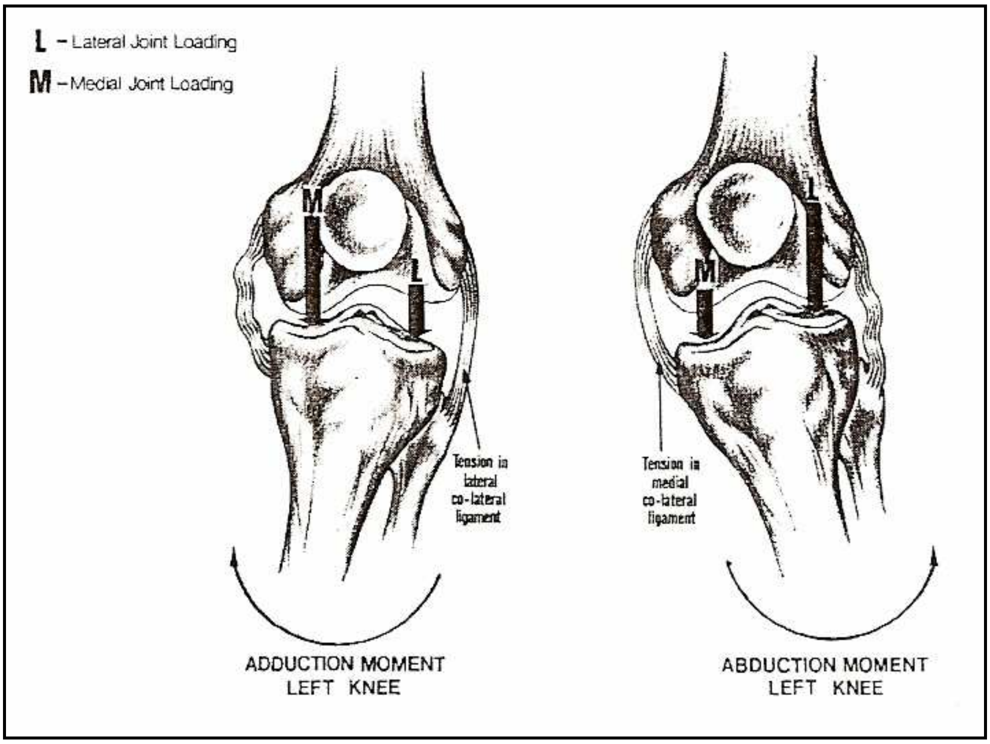


Figure 4. Schematic diagram of the relationship between moments in the knee and the internal joint load distribution. Adapted from "Gait Analysis Study on Patients With Varus Osteoarthrosis of the Knee," by J. C. H. Goh, K. Bose, and B. C. C. Khoo, 1993, Clinical Orthopaedics and Related Research, 294, p. 227.

Specifically, excessive varus malalignment is associated with higher than normal medial compartment knee joint loading and thus, may cause progressive medial femoral-tibial OA (McNicholas et al., 2000; Miyazaki et al., 2002; Moreland, Bassett, & Hanker, 1987; Sharma et al., 1998). In early OA, varus alignment implies a fourfold increased risk of progression during the subsequent 18 months (Cerejo et al., 2002; Sharma et al., 2001).

Although subjects with varus-aligned knees generally exhibit a medially-located center of pressure during static and dynamic tasks, the sole use of a static measurement, such as a weight-bearing radiograph, to predict knee joint load distribution is unreliable. This is primarily due to subjects who often demonstrate medially-located centers of pressure, despite having valgus alignment of their lower limb (Harrington, 1983; Johnson, Leitl, & Waugh, 1980). Only 50%

of knee adduction moment variability is accounted for by the mechanical axis of the lower limb in subjects with medial tibiofemoral OA, emphasising the need for dynamic evaluation of the knee joint loading (Jackson, Wluka, Teichtahl, Morris, & Cicuttini, 2004). Analysis of functional dynamic tasks such as walking represent a non-invasive method for establishing the distribution of tibiofemoral load and helps to better understand the biomechanical factors that contribute to the pathogenesis of knee OA.

2.3 Treatment of Leg Malalignment in the Frontal Plane and Knee Osteoarthritis

Leg malalignment not only causes a cosmetic problem, but also alters the knee biomechanics as shown in Section 2.2. The present section comprised an overview of different common treatment strategies of leg malalignment in the frontal plane and knee OA subdivided in surgical interventions (Section 2.3.1) and non-invasive treatments (Section 2.3.2).

2.3.1 Surgical interventions

During childhood and with remaining growth, asymmetrical suppression of the physeal growth offers an elegant solution for the treatment of leg malalignment. Lateral hemiepiphyseodesis provides a growth tether allowing for continued growth to provide correction of the varus deformity (Blount & Clarke, 1949). Moreover, it is a technique with minor operative trauma and without the need for a special after-treatment and weight-bearing curtailing.

Permanent hemiepiphyseodesis is effective, but relies on precise calculation of remaining growth and perfect surgical timing (Ferrick, Birch, & Albright, 2004; Inan, Chan, & Bowen, 2007). Temporary hemiepiphyseodesis, using staples or plates, has been shown to provide gradual deformity correction, yet may allow resumption of growth if the implants are removed in a timely fashion (Blount & Clarke, 1949; Stevens, 2007). Problems with stables have included implant failure, extrusion, and physeal damage resulting in permanent closure of the physis (Wiemann, Tryon, & Szalay, 2009). The eight-plate is purported to allow guided growth with the prospect of reducing the complications related to physeal stapling, and in one series provided more rapid correction than stapling (Stevens, 2007). One plate per physis is used. The placement in distal femur, proximal tibia, or both is based on the location of primary deformity. Figure 5 shows a typical postoperative radiograph for varus deformity treated with eight-plate hemiepiphysiodesis.

Figure 5. Eight-plate implanted for correction of a varus deformity. The hardware was subsequently removed in a routine fashion after completion of deformity correction.

However, when using hemiepiphyseodesis it is possible that patients do not achieve the desired axis correction because of insufficient remaining growth potential. Moreover, the definition of the ideal point in time for plate removal is still open (Wiemann et al., 2009).

After growth is completed or the remaining growth potential is too low, the gold standard for knee angular deformity is the corrective osteotomy. In contrast to hemiepiphyseodesis, it is a major surgical intervention that requires internal or external fixation and restricted weight-bearing. Osteotomies, especially of the proximal tibia, are high-risk surgeries, with significant incidence of compartment syndrome, neurovascular injury, and overcorrection or undercorrection (Pinkowski & Weiner, 1995).

High tibial osteotomy (HTO) – opening or closing – is a biomechanically focused surgical intervention for leg malalignment and early medial compartment OA of the knee in young and active patients (Aleto, Berend, Ritter, Faris, & Meneghini, 2008). It is called high because it is carried out high on the tibia, close to the knee joint. The high tibial medial open-wedge valgus osteotomy for correction of distal malalignment in the varus knee using a medial plate-fixator represents a popular surgical technique. It avoids detachment of the tibialis anterior muscle, the risk of peroneal nerve damage, leg shortening, and loss of correction when compared with the lateral closing-wedge osteotomy (Lobenhoffer, Agneskirchner, & Zoch, 2004).

The rationale for using a HTO to treat OA of the medial compartment of the knee stems from studies of load distribution between the medial and lateral compartments according to the malalignment of the mechanical axis in the frontal plane as shown in Section 2.2. High eccentric load concentration of the medial compartment can be reduced by lateral shift of the axial load. Figure 6 shows a schematic representation of a HTO.

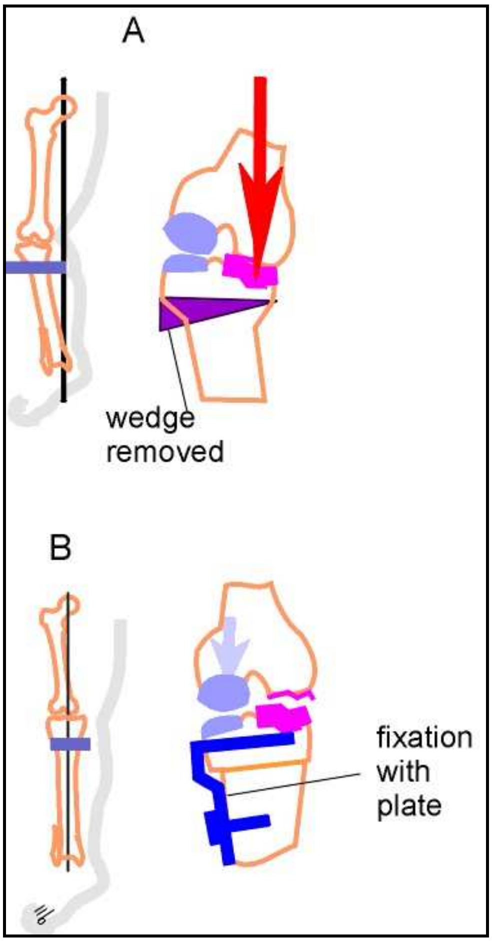


Figure 6. Schematic representation of a high tibial osteotomy. The medial compartment is damaged, the joint cartilage in the lateral compartment is healthy. The mechanical axis passes outside the knee joint. The body weight (red arrow) passes through the already damaged medial compartment. The tibia is divided below the knee joint and a wedge of bone is removed (A). The limb axis is changed to the outside and the divided bone is stabilized in that corrected position by a plate and screw (B). In this new position of the tibia, the body weight now passes through the healthy cartilage on the outside of the knee joint (blue arrow). Adapted from "Knee joint alternative operations," retrieved from http://www.totaljoints.info /Knee_alternative_oper.htm

The indications for surgery have been broadened to treat varus malalignment of the knee in younger and more active patients without signs of knee OA or milder OA allowing for decreased pain with sporting and recreational activities (Mont et al., 2004). When successful, HTO can improve clinical symptoms, prevent progression of OA, and postpone total joint replacement of the knee (Agneskirchner, Hurschler, Wrann, & Lobenhoffer, 2007). With this intervention, a high adduction moment can be also reduced to normal. A reduced adduction moment seems important for a favorable outcome after HTO (Prodromos, Andriacchi, & Galante, 1985). Catani et al. (1998) verified the influence of surgical correction of the tibiofemoral alignment on the adduction moment of the knee (Table 1). These data show that there is a correlation between surgical correction of the varus malalignment and loading of the knee joint.

Table 1
Contingency Table as Related to the Phase of Abduction/Adduction and Surgical Correction

TFA	160°-170°	171°-174°	175°-183°	Total
Abduction moment	100%	60%	16.67%	66.67%
Adduction moment	0%	40%	83.33%	33.33%
Total	100%	100%	100%	100%

Note. TFA = outer tibiofemoral alignment; TFA: 160°-170° = hypercorrection; TFA: 171°-174° = normal correction; TFA: 175°-183° = hypocorrection. Adapted from “The influence of clinical and biomechanical factors on the results of valgus high tibial osteotomy,” by F. C. Catani, M. Marcacci, M. G. Benedetti, A. Leardini, A. Battistini, F. Iacono, and S. Giannini, 1998, *La Chirurgia degli organi di movimento, 83*, p. 259.

The literature confirms that hypercorrection equal to 2-4 degrees favors the maintenance of correction in time (Berman, Bosacco, Kirshner, & Avolio, 1991; Brueckmann & Kettelkamp, 1982; Insall, Joseph, & Msika, 1984; Stuart et al., 1990).

In general, the clinical outcome after HTO is satisfying. For the treatment of medial compartment knee OA in the active patient this approach demonstrated favourable clinical results and allowed patients to return to sports and recreational activities similar to the preoperative level (Salzmann et al., 2009). However, results of HTO tended to deteriorate over time (Hernigou, Medevielle, Debeyre, & Goutallier, 1987). Factors associated with deterioration have been an older age at time of the surgery (Naudie, Bourne, R. B., Rorabeck, & Bourne,

T. J., 1999; Pfahler et al., 2003), obesity (more than 1.32 times the ideal weight; Coventry, Ilstrup, & Wallrichs, 1993), less constitutional preoperative tibial varus (<5°), advanced femoral-tibial OA of the medial compartment with >50% reduction in the joint space (Jenny, J. Y., Tavan, Jenny, G., & Kehr, 1998), and severe limitation of motion before surgery (Naudie et al., 1999). Goutallier, Manicom, Sariali, Bernageau, and Radier (2006) reported that obtaining stable angle correction over time and therefore good functional and radiographic outcomes may require modification of the correction angle according to the degree of femoral anteversion. Femoral torsion, in this context, was positive (anteversion) when the femoral neck axis (the line joining the center of the neck to the center of the head) fell anterior to the bicondylar line (the line through the most prominent part of the two condyles). The data of this study suggest the need for more valgus angulation when femoral anteversion is small (<14°) and less valgus angulation when femoral anteversion is marked (≥14°; Figure 7). This is an important finding recognizing the fact that arthritis increases as the femoral anteversion decreases (Eckhoff, Kramer, Alongi, & VanGerven, 1994) and should be therefore considered for surgical reconstruction of the knee.

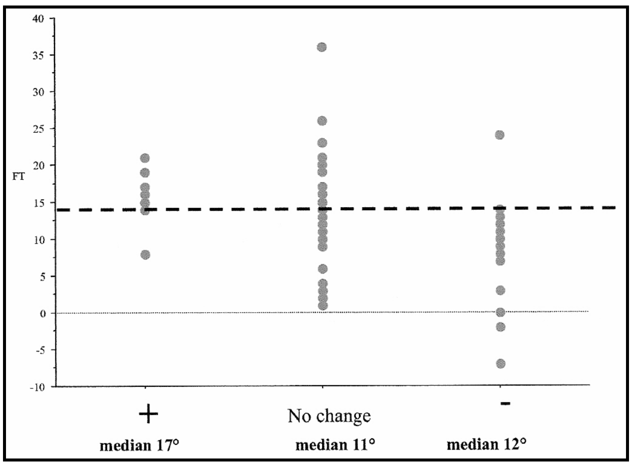


Figure 7. Distribution of patients who gained valgus angulation (+), had no change, or lost valgus angulation (-), according to the amount of femoral torsion (FT). Adapted from "Influence of Lower-Limb Torsion on Long-Term Outcomes of Tibial Valgus Osteotomy for Medial Compartment Knee Osteoarthritis," by D. Goutallier, S. Van Driesche, O. Manicom, E. Sariali, J. Bernageau, and C. Radier, 2006, *The Journal of Bone and Joint Surgery (American volume), 88*, p. 2442.

2.3.2 Non-invasive treatments

The effect of non-invasive treatments of pathological varus alignment and knee OA is generally controversial. The aim of current non-invasive treatments of pathological medial compartment knee loading is to reduce knee adduction moments and to stabilize knees with medial compartment OA or knees with varus alignment. These load-modifying techniques include bracing (Fantini Pagani, Potthast, & Brüggemann, 2010; Lindenfeld, Hewett, & Andriacchi, 1997; Pollo, Otis, Backus, Warren, & Wickiewicz, 2002), footwear modifications (Fisher, Mündermann, & Andriacchi, 2004; Kerrigan et al., 2002; Kerrigan, Karvosky, Lelas, & Riley, 2003) and gait training (Mündermann, Dyrby, Hurwitz, Sharma, & Andriacchi, 2004). Figure 8 shows an example of an orthosis model designed to reduce the knee adduction moment. A three-point bending system is used, which incorporates unilateral tubular frames with straps for fixation of the orthosis on the subject's leg (Fantini et al., 2010).

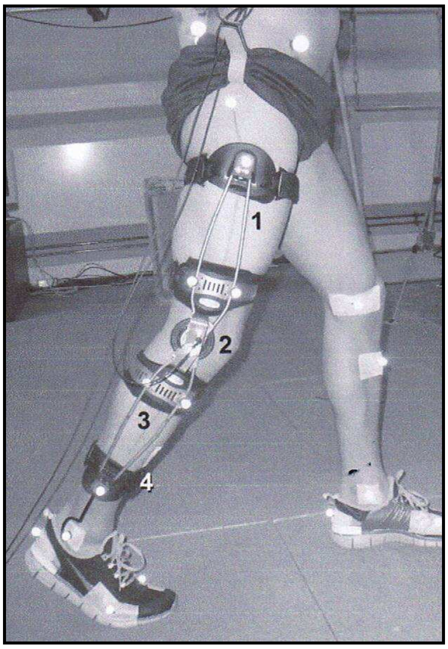


Figure 8. Orthosis model designed to reduce the knee adduction moment. Femoral module (1), orthosis' axis with the strain gauge instrumentation (2), tibial module (3) and straps (4). Adapted from "The effect of valgus bracing on the knee adduction moment during gait and running in male subjects with varus alignment," by C. H. Fantini Pagani, W. Potthast, and G.-P. Brüggemann, 2010, Clinical Biomechanics, 25, p. 72.

A regression equation obtained in the study of Mündermann et al. (2004) predicts a reduction of the maximum knee adduction moment of 10.2% when walking at 0.8 meters/second compared with 1.2 meters/second. In comparison, the maximum knee adduction moment was reduced by 13% using bracing (Pollo et al., 2002) and by 6% or 8.2% using 5° or 10° valgus insoles (Kerrigan et al., 2002), respectively. Laterally wedged orthoses have the effect of increasing rearfoot eversion (Butler, Marchesi, Royer, & Davis, 2007), which is likely a compensation for the greater inclination of the tibia and may help to reduce the knee adduction moment in young patients with varus malalignment (Barrios, Davis, Higginson, & Royer, 2009). Based on a regression equation of a previous investigation (Andrews, Noyes, Hewett, & Andriacchi, 1996) involving healthy subjects, a reduction in maximum knee adduction moment by 10% with a 10° greater foot progression angle (more toe-out foot placement) can be expected.

Additionally, neuromuscular factors play a major role in active knee joint stability and include the strength and coordination of the muscles. (Jackson et al., 2004). One of the major muscle groups involved in knee joint stability is the quadriceps femoris. Shortly after heel-strike there is rapid knee joint flexion and the rate and degree of this flexion is controlled by eccentric activity of the quadriceps muscles. The quadriceps muscles act to prevent excessive or rapid flexion of the knee, and muscular weakness is a feature common to symptomatic individuals with knee OA (Felson et al., 2000; Slemenda et al., 1997). Cross-sectional studies that examined the effect of quadriceps strength in people with established knee OA demonstrated an association between weak quadriceps and both radiographic and symptomatic OA of the knee (Slemenda et al., 1998). Another study conducted on women without knee OA found that individuals with significantly less concentric and eccentric quadriceps and hamstring muscle strength had significantly higher knee joint loads than did women with stronger quadriceps and hamstring muscles (Mikesky, Meyer, & Thompson, 2000). This inverse relationship between joint load and muscular strength implies that strong muscular support of the knee reduces the load being applied to articular cartilage, which may be a risk for the development of joint pathology. Nonetheless, it must be acknowledged that muscle inhibition due to pain may contribute to muscle weakness in people with established cases of knee OA and inadvertently cause disuse atrophy. Whether the knee joint benefits from quadriceps strengthening once OA is apparent is unclear, but it may be an effective treatment for maintaining knee joint stability in patients with varus malalignment of the knee without signs of knee OA (Fisher et al., 1993; Fransen, Crosbie, & Edmonds, 2001; Huang, Lin, Yang, & Lee, 2003).

Although previous discussion has focused on muscle strength, coordination of muscle contraction may also influence the pathogenesis of knee OA. A review by O'Conner and Brandt (1993) concluded that if muscle contraction is not properly coordinated, the joint will exceed its normal extreme of excursion and the loading of its cartilage will be excessive. However, further work is required to gain a better understanding of the muscular contribution toward the pathogenesis and possible treatment of knee OA (Jackson et al., 2004).

3 Fundamentals of Gait Analysis

The basic descriptions of motion will be provided in this section helping to describe and understand normal and pathological function and to facilitate the interpretation of the identified gait deviations in patients with varus malalignment of the knee.

3.1 From Marker Placement to Gait Kinematics

Human movement analysis begins by dividing the body into a series of segments. Each segment is then assumed to behave as a rigid body with fixed center of mass and inertial properties. Defining the position of a point in space requires three coordinates (x,y,z). Any individual point on a segment may be defined in this way and three individual points are required to locate the position and orientation of a segment. The interrelationship of these segments finally results in joint angles often expressed using kinematic graphs. In practice points are identified by retroreflective markers. The placement of the markers on standardized, constant defined locations on the subject's body is therefore the basis for the determination of joint centers, definition of body segments (computerized model) and the calculation of kinematic and kinetic parameters (Figure 9).

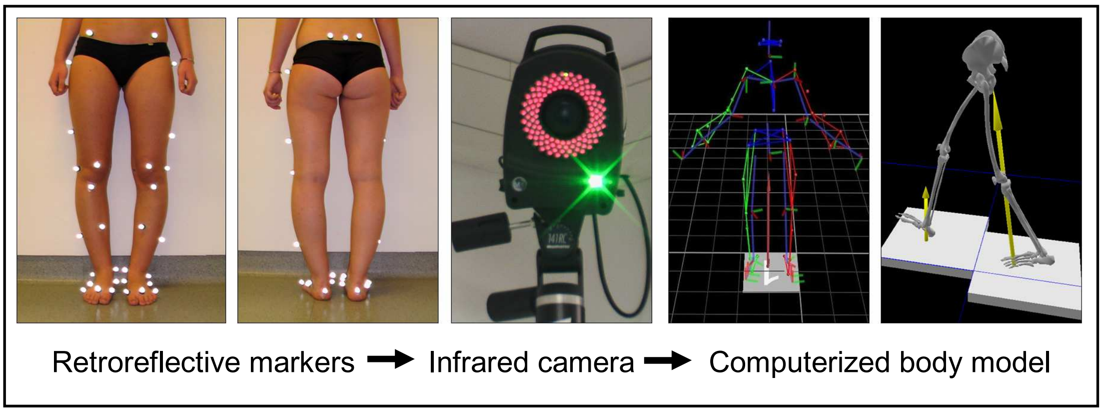


Figure 9. Development of a computerized body model.

3.2 Gait Cycle

Each gait cycle is divided into two periods, the stance and swing phase. *Stance* is the term used to designate the entire period during which the foot is on the ground. Stance begins with initial contact (Figure 10). The word *swing* applies to the time the foot is in the air for limb advancement. Swing begins as the foot is lifted from the floor (toe-off).

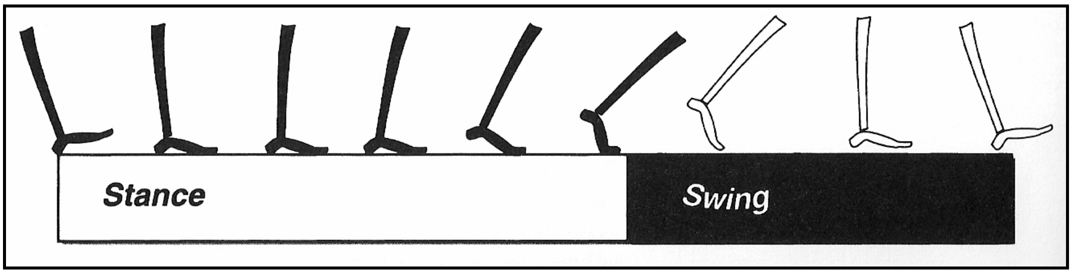


Figure 10. Divisions of the gait cycle. Clear bar represents the duration of stance. Shaded bar is the duration of swing. Limb segments show the onset of stance with initial contact, end of stance by roll-off of the toes, and end of swing by floor contact again. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 4.

The gross normal distribution of the floor contact periods is 60% for stance and 40% for swing (Murray, Drought, & Kory, 1964). The precise duration of these gait cycle intervals varies with the person's walking velocity (Andriacchi, Ogle, & Galante, 1977). At the customary 1.3 m/s rate of walking, the stance and swing periods represent 62% and 38% of the gait cycle, respectively (Perry, 1992). The duration of both gait periods shows an inverse relationship to walking speed. That is, both total stance and swing times are shortened as gait velocity increases. The change in stance and swing times becomes progressively greater as speed slows (Perry, 1992).

The gait cycle also has been identified by the descriptive term stride and step (Murray et al., 1964; Figure 11).

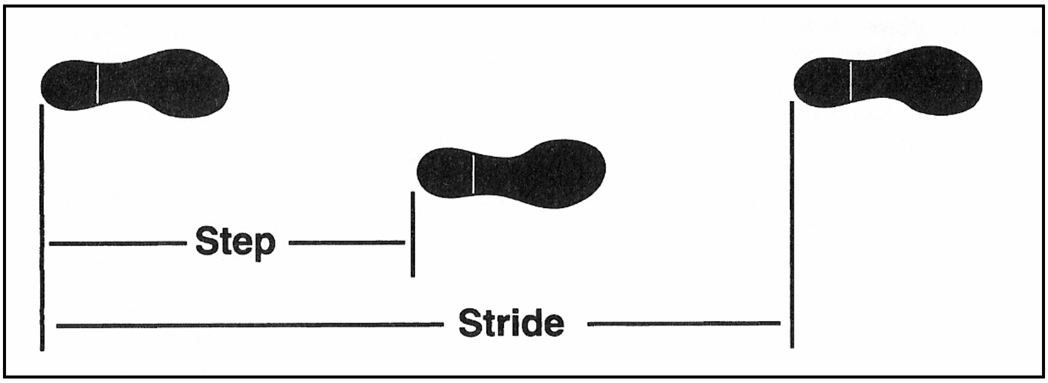


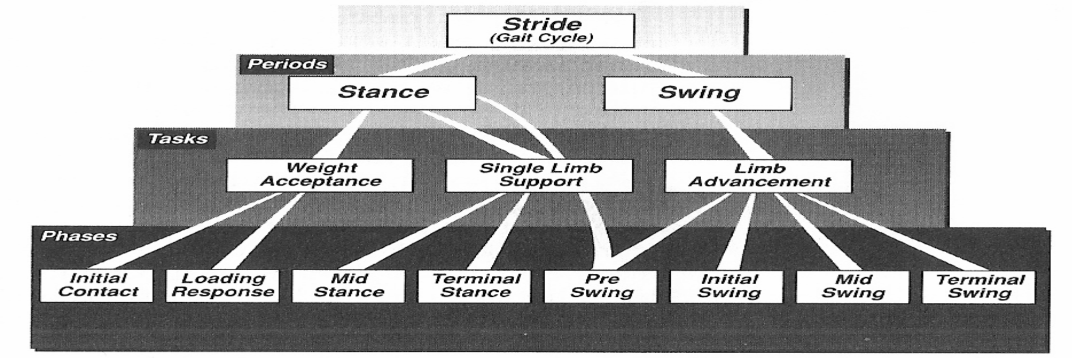
Figure 11. A step versus a stride. Step length is the interval between initial contact of each foot. Stride length continues until there is a second contact by the same foot. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 6.

Stride is the equivalent of a gait cycle. It is based on the actions of one limb. The duration of a stride is the interval between two sequential initial floor contacts by the same limb. *Step* refers to the timing between the two limbs. There are two steps in each stride (or gait cycle). The interval between an initial contact by each foot is a step (i.e., left and then right).

3.3 Phases of Gait

The phases of gait related to a different functional demand. It now is evident that each stride contains eight functional patterns. Technically these are sub phases, as the basic divisions of the gait cycle are stance and swing. The sequential combination of the phases also enables the limb to accomplish three basic tasks. These are weight acceptance, single limb support and limb advancement (Table 2).

Table 2
Divisions of the Gait Cycle



Note. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 10.

Weight acceptance begins the stance period and uses the first two gait phases (initial contact and loading response). Single limb support continues stance with the next two phases of gait (mid stance and terminal stance). Limb advancement begins in the final phase of stance (pre-swing) and then continues through the three phases of swing (initial swing, midswing and terminal swing).

Weight acceptance is the most demanding task in the gait cycle. Three functional patterns are needed: shock absorption, initial limb stability and the preservation of progression. The challenge is the abrupt transfer of body weight onto a limb that has just finished swinging forward and has an unstable alignment (Perry, 1992). Two gait phases are involved, initial contact (Figure 12) and loading response (Figure 13).

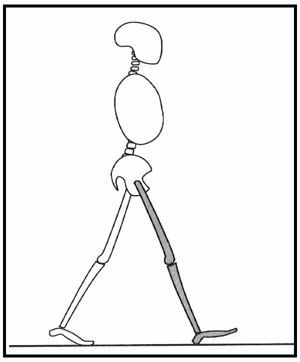


Figure 12. Initial Contact. Interval: 0-2% of the gait cycle. This phase includes the moment when the foot just touches the floor. Shading indicates the reference limb. The other limb (clear) is at the end of terminal stance. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 12.

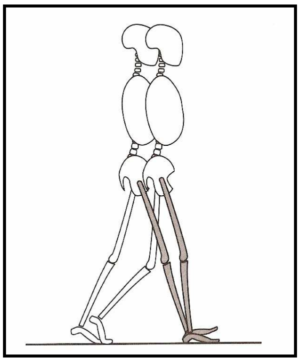


Figure 13. Loading Response. Interval: 0-10% of the gait cycle. The phase begins with initial floor contact and continues until the other foot is lifted for swing. Body weight is transferred onto the forward limb (shaded). The opposite limb (clear) is in its pre-swing phase. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 12.

Lifting the other foot for swing begins the *single limb support* interval for the stance limb. This continues until the opposite foot again contacts the floor. During the resulting interval, one limb has the total responsibility for supporting body weight while progression must be continued (Perry, 1992). Two phases are involved in single limb support: mid stance (Figure 14) and terminal stance (Figure 15).

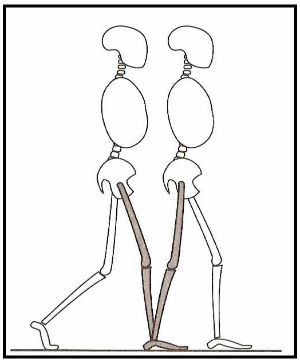


Figure 14. Mid Stance. Interval: 10-30% of the gait cycle. The phase begins as the other foot is lifted and continues until body weight is aligned over the forefoot. In the first half of single limb support, the limb (shaded) advances over the stationary foot. The opposite limb (clear) is advancing in its mid swing phase. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 13.

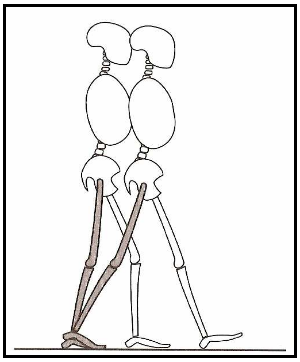


Figure 15. Terminal Stance. Interval: 30-50% of the gait cycle. The phase begins with heel rise and continues until the other foot strikes the ground. During the second half of single limb support, the heel rises and the limb (shaded) advances over the forefoot. The other limb (clear) is in terminal swing. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 13.

Limb advancement begins in the final phase of stance. To meet the high demands of advancing the limb, preparatory posturing begins in stance. The limb swings through three postures as it lifts itself, advances and prepares for the next stance interval (Perry, 1992). Four gait phases are involved: pre-swing (Figure 16), initial swing (Figure 17), mid swing (Figure 18) and terminal swing (Figure 19).

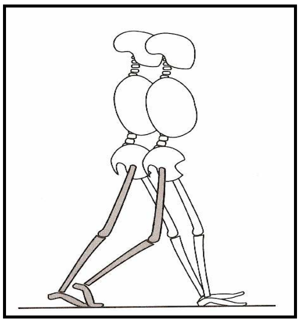


Figure 16. Pre-Swing. Interval: 50-60% of the gait cycle. This final phase of stance begins with initial contact of the opposite limb and ends with ipsilateral toe-off. Floor contact by the other limb (clear) has started terminal double support. The reference limb (shaded) responds with increased ankle plantar flexion, greater knee flexion and loss of hip extension. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 14.

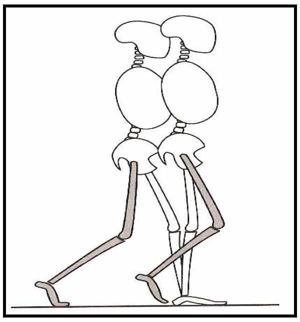


Figure 17. Initial Swing. Interval: 60-73% of the gait cycle. This phase begins with lift of the foot (shaded) from the floor and ends when the swinging foot is opposite the stance foot. The other limb (clear) is in early mid stance. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 14.

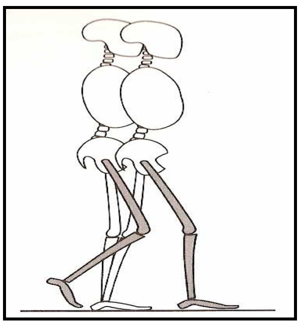


Figure 18. Mid Swing. Interval: 73-87% of the gait cycle. This second phase of the swing period begins as the swinging limb is opposite the stance limb. The phase ends when the swinging limb is forward and the tibia is vertical (i.e., hip and knee flexion postures are equal). Advancement of the limb (shaded) anterior to the body weight line. The other limb (clear) is in late mid stance. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 15.

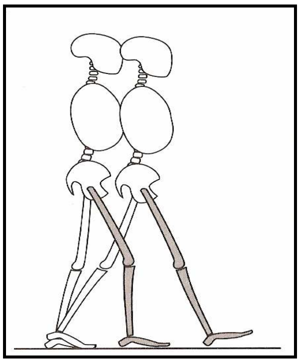


Figure 19. Terminal Swing. Interval: 87-100% of the gait cycle. This final phase of swing begins with a vertical tibia of the reference limb (shaded) and ends when the foot strikes the floor. The other limb (clear) is in terminal stance. Adapted from "*Gait Analysis. Normal and Pathological Function*," by J. Perry, 1992, Thorofare, NJ: SLACK Incorporated, p. 15.

To sum up the above presented phases of gait, Figure 20 shows the division of the gait cycle on the basis of the ankle angle in the sagittal plane.

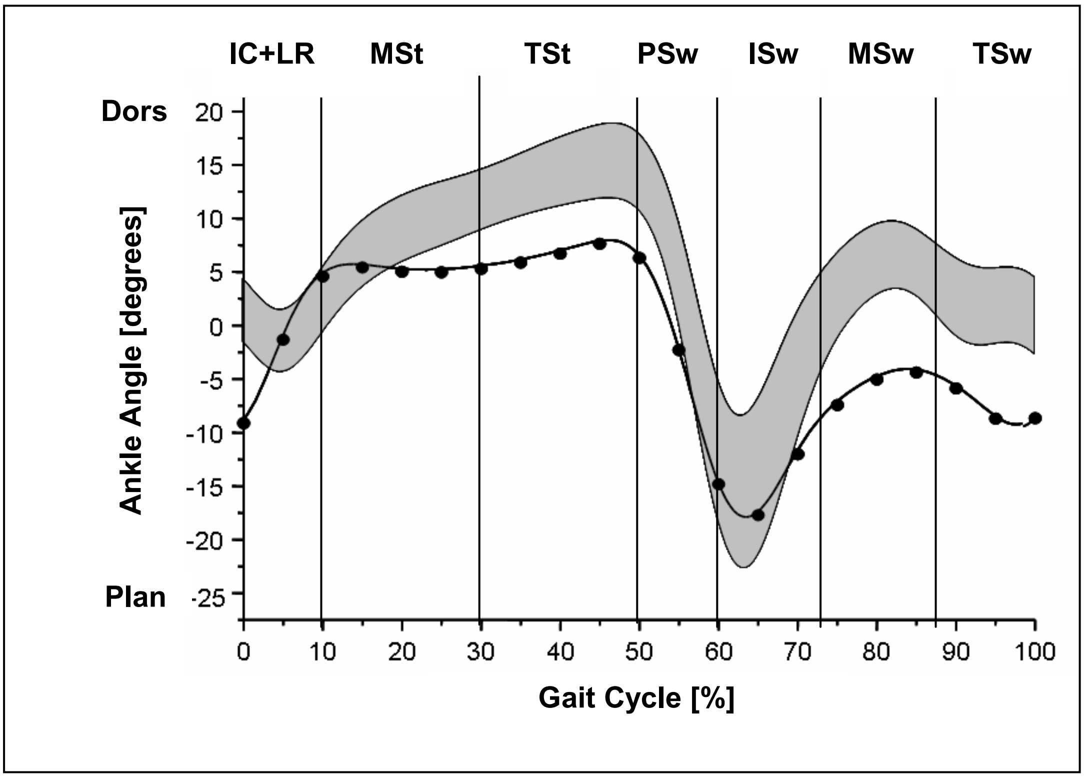


Figure 20. Division of the gait cycle on the basis of the ankle angle in the sagittal plane. The gray graph indicates the mean with standard deviation of a healthy subject group (*n* = 15). The dotted graph indicates a patient with pathological pes equines. IC = Initial Contact; LR = Loading Response; MSt = Mid Stance; TSt = Terminal Stance; PSw = Pre-Swing; ISw = Initial Swing; MSw = Mid Swing; TSw = Terminal Swing; Dors = Dorsiflexion; Plan = Plantarflexion.

4 Previous Gait Analysis Studies on Patients With Knee Osteoarthritis

Due to the lack of three-dimensional gait analysis studies in children and adolescents with pathological varus alignment of the knee but no signs of knee OA, the following section exhibits the current state of research regarding gait analysis studies on adult patients – typically older than 40 years – with knee OA.

4.1 Relevance of the Knee Adduction Moment Regarding Articular Cartilage Degeneration and Disease Progression in the Medial Compartment of the Knee Joint

Recently, quantitative gait analysis has been investigated as a means to quantify limb alignment (Hunt et al., 2008; Mündermann et al., 2008; Vanwanseele, Parker, & Coolican, 2009). The measurement of lower limb alignment during walking can be combined with kinetic data to help with the understanding of the local loading environment at the ankle, knee and hip joint.

The external knee adduction moment is an often-used predictor of knee joint loading (Hurwitz et al., 2002) and a commonly used outcome measurement reported from gait analysis data in adults with knee OA. Furthermore, it is a strong contributing factor to articular cartilage degeneration and disease progression in the medial knee compartment (Andriacchi & Mündermann, 2006; Hurwitz et al., 2002; Miyazaki et al., 2002; Mündermann et al., 2004; Sharma et al., 1998). There is also a significantly correlation between joint space narrowing of the medial knee compartment during a six year period with the knee adduction moment at entry. In addition, logistic regression analysis showed that the risk of progression of knee OA increased 6.46 times with a 1% increase in adduction moment and 1.22 times with a one year increase in age (Miyazaki et al., 2002). As a result, previous studies found increased maximal adduction moment in knees with medial OA (Andrews et al., 1996; Baliunas et al., 2002; Miyazaki et al., 2002; Sharma et al., 1998; Weidenhielm, Svensson, Broström, & Mattsson, 1994). Consequently, the external adduction moment of the knee is a major determinant of medial to lateral load distribution (Schipplein & Andriacchi, 1991) and it is responsible for the biomechanical abnormality of the medial compartment knee OA (Andriacchi, 1994). Figure 21 illustrates the calculation of the knee adduction moment.

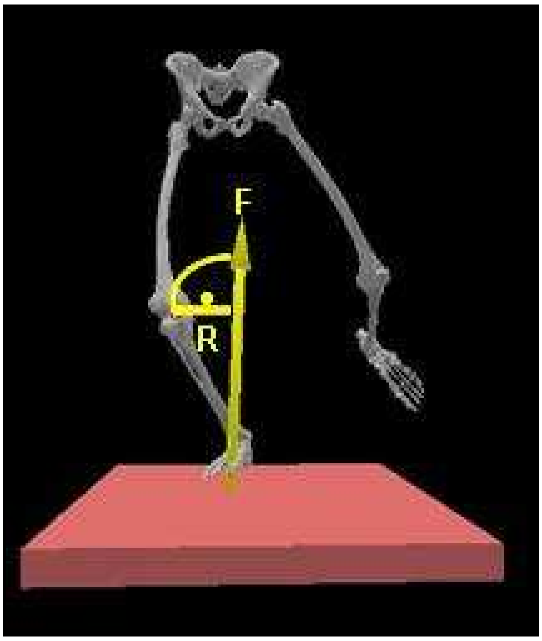


Figure 21. Schematic representation of the external knee adduction moment. The calculation is based on the cross product of the ground reaction force (F) and its vertical distance (R) to the knee joint center. Adapted from "Nordic Walking, Walking, Laufen – Biomechanische Betrachtung. 3-dimensionaler Vergleich der Gelenkbelastung der unteren Extremitäten," by F. Stief, 2008, Saarbrücken: Verlag Dr. Müller, p. 57.

4.2 Relationship Between Static Varus Malalignment and Dynamic Knee Adduction Moment

Previous studies have to a certain extent shown a relation between the degree of knee deformity and the force acting at the knee. A positive correlation between mechanical axis alignment (varus alignment) and maximum external knee adduction moment has been reported (Andrews et al., 1996; Hurwitz et al., 2002; Miyazaki et al., 2002; Noyes, Barber-Westin, & Hewett, 2000; Wada et al., 2001). Hurwitz et al. (2002) found that the radiographic measures of OA severity in the medial compartment were predictive of peak knee adduction moments (r = .46 [.43, .48], p < .001). Weidenhielm et al (1994) found a weak correlation between dynamic peak knee adduction moment and the static, radiographic hip-knee-ankle (HKA) angle (r = .32, p < .05) in patients with medial OA, a clinically stable joint and comparatively mild deformity (Ahlbäck grades 1 to 3). They also observed a moderately high correlation between knee adduction moment and HKA angle in midstance (r = .46, p < .001) and reported that one fifth of the mediolateral knee load in midstance could be explained by the varus knee deformity accompanying OA.

However, there are conflicting views regarding the influence of lower limb alignment on external moments in the knee during gait. Prodromos et al. (1985) and Wang, Kuo, Andriacchi, and Galante (1990) reported no correlation between limb alignment and knee adduction moment. Andrews et al. (1996) postulates that this may be due to their inclusion of patients with anterior cruciate ligament deficiency. Similarly, McNicholas et al. (2000) also found no correlation between the dynamic gait parameter of the knee adduction moment in early stance and the static HKA angle in individuals with tibio-femoral OA and no anteroposterior (AP) laxity who had undergone unilateral total meniscectomy.

These controversial results indicate that the effect of static varus malalignment on dynamic knee joint loading is not completely understood. Individual gait compensatory mechanisms could be a reason for partial differences between static alignment and the dynamic joint loading situation.

4.3 Biomechanical Compensatory Mechanisms in Patients With Knee Osteoarthritis

Previous authors investigating the relationship between static lower limb alignment and dynamic knee joint loading point to the potentially confounding influence of compensatory gait characteristics that may explain differences between static and dynamic measures (Andrews et al., 1996; Hurwitz et al., 2002; Mündermann, Dyrby, & Andriacchi, 2005). Some abnormal mechanics in individuals with knee OA appear related to malalignment, while others appear compensatory in nature. Those related to alignment should be present in healthy, but varus-aligned knees, and mechanics related to OA impairments should be absent in children and adolescents with varus malalignment of the knee.

Specifically, patients with knee OA develop compensatory mechanisms such as an increased *external foot progression angle* (Figure 22; Guo, Axe, & Manal, 2007; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008; Wada et al., 1998; Wang et al., 1990) to reduce the dynamic loading on the medial knee compartment.

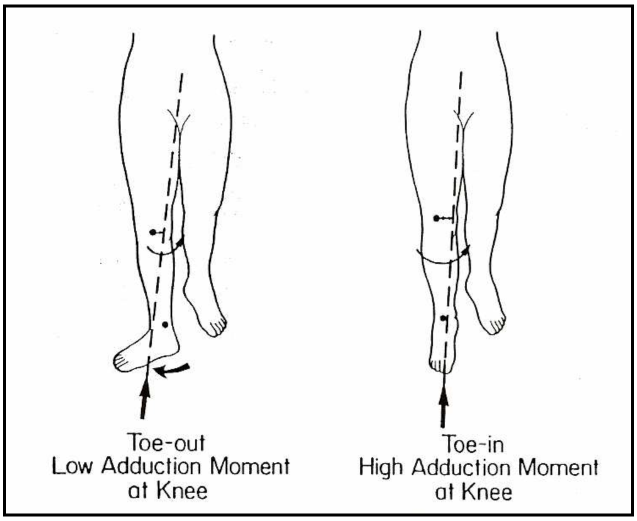


Figure 22. A mechanism used to lower the adduction moment at the knee. Adapted from “The Influence of Walking Mechanics and Time on the Results of Proximal Tibial Osteotomy,” by J. W. Wang, K. N. Kuo, T. P. Andriacchi, and J. O. Galante, 1990, *The Journal of Bone and Joint Surgery (American volume), 72A*(6), p. 908.

These results suggest that walking with a toe-out strategy may benefit persons with knee OA. Moreover, based on a regression equation of a previous investigation (Andrews et al., 1996) involving healthy subjects, a reduction in maximum knee adduction moment by 10% with a 10° greater foot progression angle (more toe-out foot placement) can be expected.

Another compensating strategy to reduce dynamic joint loading is reduced *walking speed* (Al-Zahrani & Bakheit, 2002; Kaufman et al., 2001). Different authors reported that patients with medial knee OA walk at slower speeds (Al-Zahrani & Bakheit, 2002; Gok et al., 2002; Kaufman et al., 2001; Weidow, Tranberg, Saari, & Kärrholm, 2006) and with shorter *stride lengths* (Weidow et al., 2006) compared to healthy controls. In addition, Prodromos et al. (1985) showed that patients in the low adduction-moment group had significantly shorter mean stride length than patients in the high adduction-moment group and it was significantly below normal in the low adduction-moment (Figure 23).

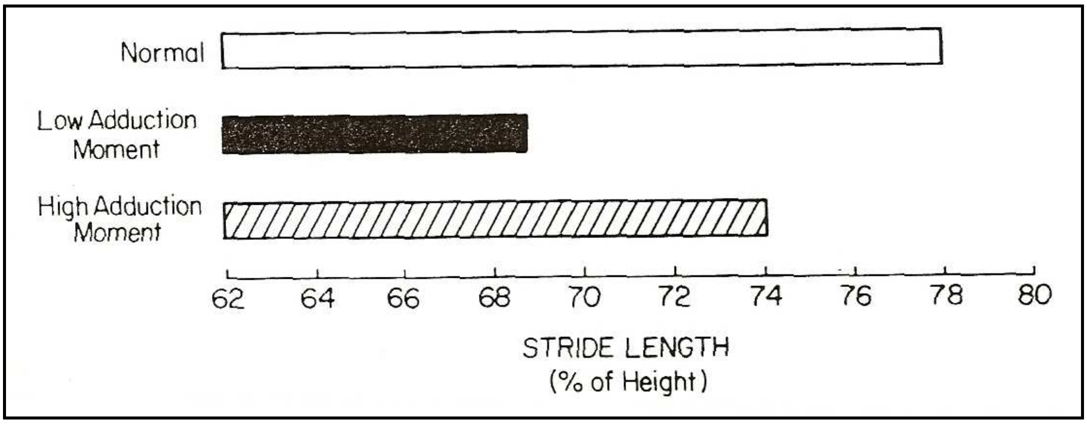


Figure 23. Preoperative relationship between the external knee adduction moment and stride length. Before tibial osteotomy, patients in the low adduction-moment group had a significantly shorter stride length than did patients in the high adduction-moment group. Adapted from “A Relationship between Gait and Clinical Changes following High Tibial Osteotomy,” by C. C. Prodromos, T. P. Andriacchi, and J. O. Galante, 1985, *The Journal of Bone and Joint Surgery (American volume), 67*, p. 1192.

Other studies indicated that patients with medial knee OA have a smaller range of *knee flexion* during the stance phase of walking (Al-Zahrani & Bakheit, 2002; Childs et al., 2004; Kaufman et al., 2001), make initial contact with the ground with the knee in a more extended position, and have higher peak external hip extension moments during terminal stance than do control subjects (Mündermann et al., 2005). Patients with symptomatic medial knee OA often stiffen their knees to reduce the demands on the quadriceps muscles and diminish pain (Al-Zahrani & Bakheit, 2002; Baliunas et al., 2002; Gok et al., 2002; Kaufman et al., 2001; Mündermann et al., 2005). Moreover, patients with medial knee OA in the study of Mündermann et al. (2005) had significantly greater first peak external knee adduction moments and lower first and second peak external hip adduction moments in the frontal plane.

Since these altered kinematic and kinetic data are not captured in a static radiograph, the true relationship between joint loading and lower limb alignment is still somewhat unclear. Therefore, an accurate assessment of the patient's lower limb biomechanics, including alignment during dynamic activities such as walking may be advantageous.

4.4 Pre- and Postoperative Gait Analysis Following High Tibial Valgus Osteotomy

As shown in Section 2.3.1, high tibial valgus osteotomy (HTVO) is an effective treatment for medial compartment knee OA. High eccentric load concentration of the medial compartment can be reduced by lateral shift of the axial load. With this intervention, a high adduction moment can be also reduced to normal. Prodromos et al. (1985) reported that the preoperative adduction moment could predict surgical outcome for knee OA with varus deformity. When the adduction moment was higher preoperatively, it was still increased postoperatively and the leg significantly changed to varus alignment again while lower adduction moment did not. Therefore, the low adduction-moment group had substantially better clinical results than the high adduction-moment group at an average of 3.2 years after HTVO (Figure 24). Moreover, the adduction moment decreased significantly soon after HTVO but tended to increase gradually after one year (Schultz, Weber, Blumentritt, & Schmalz, 2003; Wada et al., 1998). Even after valgus alignment was obtained, the adduction tended to increase with time.

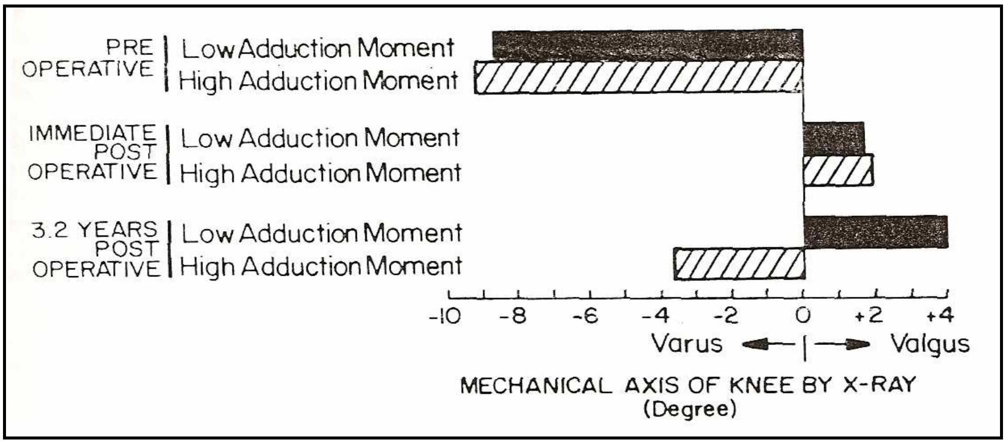


Figure 24. Preoperative, immediately postoperative and 3.2 years postoperative relationship between the external knee adduction moment and the static, radiographic mechanical axis of the knee. The high and low adduction-moment groups had nearly identical varus deformity before high tibial osteotomy and nearly identical valgus alignment immediately after the operation. At an average of 3.2 years after high tibial osteotomy, the patients in the low adduction-moment group had maintained the valgus correction, while those in the high adduction-moment group had a significant return to a varus position. Adapted from “A Relationship between Gait and Clinical Changes following High Tibial Osteotomy,” by C. C. Prodromos, T. P. Andriacchi, and J. O. Galante, 1985, *The Journal of Bone and Joint Surgery (American volume), 67*, p. 1192.

5 Research Deficit and Aim of the Thesis

Though the radiograph is an accurate measure of lower limb alignment during standing, it fails to capture any changes in alignment and joint loading that may occur when the limb is moving and weight-bearing. Therefore, the use of quantitative gait analysis as an adjunct to static radiographic measures of alignment has been investigated as a means in the study and treatment of knee OA and varus malalignment (Hunt et al., 2008; Mündermann et al., 2008).

Most studies investigating the gait of adult patients with established knee OA have focused on kinematics and kinetics at the knee in the sagittal and frontal plane (Al-Zahrani & Bakheit, 2002; Baliunas et al., 2002; Childs et al., 2004; Gok et al., 2002; Kaufman et al., 2001). However, there is a lack of research on gait data in the transverse plane (Astephen et al., 2008; Landry et al., 2007) and little attention has been paid to the changes in the mechanical environment of other joints of the affected limb. Moreover, the author of this study is unaware of previous three-dimensional gait analysis studies focused on the dynamic loading characteristics of the knee and hip joints as well as potential compensatory mechanisms in children and adolescents with pathological varus alignment of the knee but no signs of knee OA. Therefore, the main aim of this thesis was the investigation of three-dimensional knee and hip joint angles and moments as well as mechanisms of gait compensation in children and adolescents with pathological varus alignment of the knee compared to healthy control subjects and patients with established medial knee OA (Section 7). The author hypothesized that joint moments in all three planes would be different between patients and controls. The author also expected that mechanisms of gait compensation in the present patient group would be different from those reported in adult patients with medial knee OA.

A precondition to perform gait analysis on this patient group is the application of a useful model defining marker positioning, joint coordinate systems, procedures for data collection and the calculation of skeletal motion. The standard PiG model used by a vast majority of clinical gait laboratories is prone to errors arising from inconsistent anatomical landmark identification and knee axis malalignment (Leardini et al., 2005; Piazza & Cavanagh, 2000). As a result, the development and evaluation of a custom made lower body model for clinical gait analysis is shown in Section 6.

Consequently, the following concrete aims will be answered in the present thesis:

1. Is the reliability and accuracy of the custom made gait analysis model for determining three-dimensional kinematic and kinetic parameters increased compared to the standard PiG model (Section 6.4.2)?
   - The answer of this question is the basis for using the custom made model to analyse patients with varus malalignment in Section 7.

2. Does a relationship exist between static alignment obtained from radiographs and based on reflective markers (Section 7.2.1)?
   - This is important to estimate the suitability of the marker-based gait analysis system to determine frontal plane alignment of the leg.

3. Does a relationship exist between static varus malalignment obtained from radiographs and dynamic knee adduction moment (Section 7.2.2)?
   - The processing of this question will help to identify the potentially influence of compensatory gait characteristics that may explain differences between static and dynamic measures.

4. Are three-dimensional differences in joint angles present between the patient group with varus malalignment, healthy control subjects as well as patients with established medial knee OA (Section 7.2.3)?
   - The influence of static varus malalignment on gait characteristics in this young patient group without signs of knee OA and differences to those reported in patients with established medial knee OA will be shown.

5. Are three-dimensional differences in joint moments present between the patient group with varus malalignment, healthy control subjects as well as patients with established medial knee OA (Section 7.2.4)?
   - In which plane and to what extent does the static varus malalignment leads to increased joint moments and how is the dynamic joint loading situation in this patient group compared to patients with established medial knee OA? This may have important implications for the development or progression of degenerative joint disease in young patients without signs of knee OA.

6. Do potential mechanisms of gait compensation exist in patients with varus malalignment of the knee and are they different to those reported in adult patients with established knee OA (Section 7.2.5)?
   - Section 4.3 has been shown that individual compensatory mechanisms are used to reduce the dynamic loading on the medial knee compartment. This may have influence on clinical prognosis for the onset or progression of articular cartilage degeneration in this patient group and potential interventions can be tailored to each patient.

6 Development and Evaluation of a Lower Body Model for Clinical Gait Analysis

The standard PiG model used in clinical gait analysis is prone to errors arising from inconsistent anatomical landmark identification and knee axis malalignment (Ferrari et al., 2008). In this section the relevance of the development of a lower body model will be demonstrate. In addition to the detailed characterization of the model, it will be evaluated and compared with the standard PiG model (Section 6.4). The development and evaluation of this lower body model was performed between October 2006 and January 2009. Parts of this study were accepted for publication in the Journal of Applied Biomechanics on November 30th 2011 (Stief, Böhm, Michel et al., 2011).

6.1 Introduction

Clinical gait analysis is used as a tool to assist with clinical investigation and surgical planning and to assess the outcome of interventions (Baker, 2006). Skeletal movements during gait are typically recorded using markers placed on the surface of the skin. Different models exist in clinical applications defining marker positioning, joint coordinate systems, procedures for data collection and the calculation of skeletal motion (Grood & Suntay, 1983; Kadaba, Ramakrishnan, & Wooten, 1990; Miyazaki et al., 2002). The placement of the markers has considerable influence on the accuracy of gait studies (Gorton, Hebert, & Goode, 2001). Accuracy, in this context, refers to an accurate knee axis alignment according to the knee varus/valgus and flexion/extension range of motion (ROM) during gait (Schwartz & Rozumalsky, 2005). Moreover, any model for clinical movement analysis will only prove useful if it displays adequate reliability (Cappozzo, 1984).

One of the first models proposed by Davis, Ounpuu, Tyburski, and Gage (1991) and known as PiG, is used by a vast majority of clinical gait laboratories. Fifteen retro-reflective markers were placed in this lower body PiG model (Figure 25).

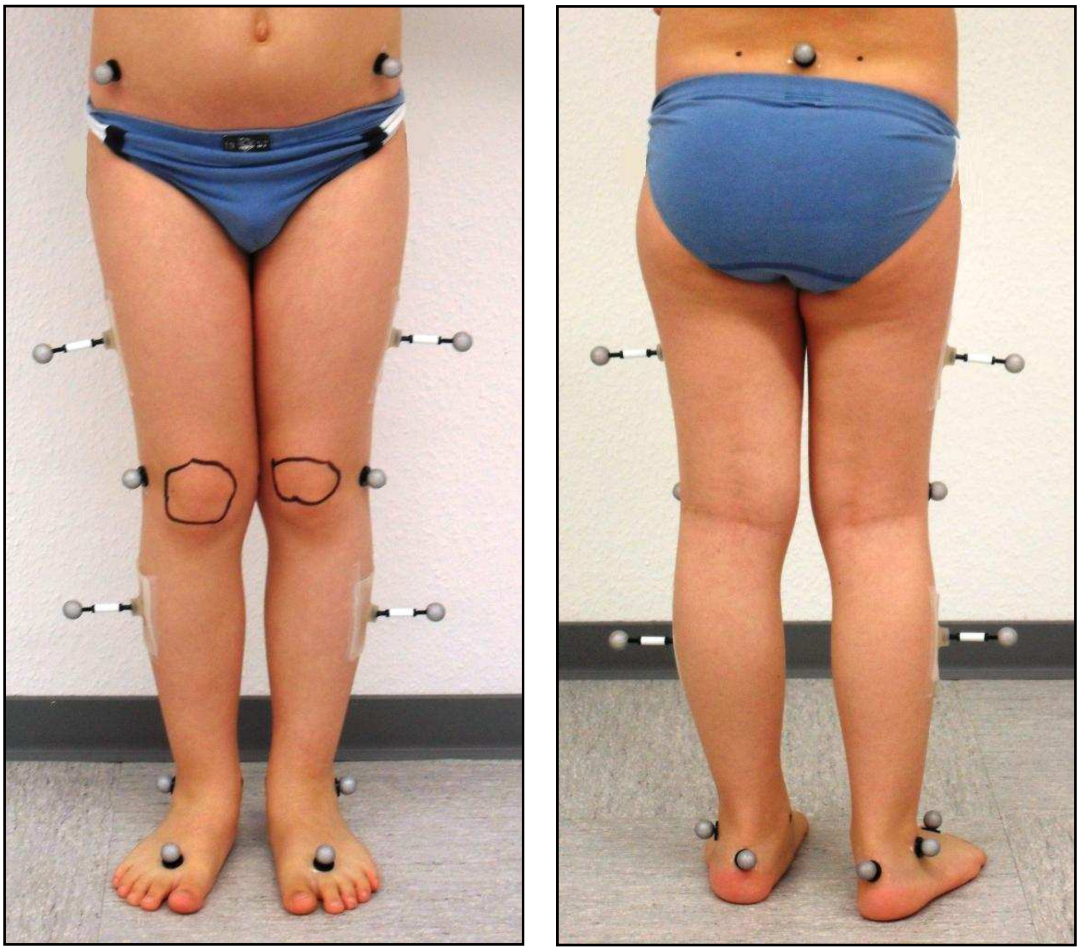


Figure 25. Placement of the markers for the PiG model.

It has been shown that inter-session and inter-assessor reliability are low for this model, especially at the hip and knee joint in the frontal and transverse plane (McGinley, Baker, Wolfe, & Morris, 2009). The errors in the PiG model, for example knee varus/valgus ROM up to 30° (Ferrari et al., 2008), are very likely caused by inconsistent anatomical landmark identification and marker positioning by the assessor. This leads to well documented errors of skin movement (Leardini et al., 2005) and kinematic cross-talk, that is, one joint rotation (e.g., flexion) being interpreted as another (e.g., adduction) due to axis malalignment (Piazza & Cavanagh, 2000). Moreover, internal/external hip rotation measurements using a PiG thigh wand marker have been shown to have large variability (Gorton, Hebert, & Goode, 2002). If a thigh wand marker is to be used in clinical practice, it is necessary that patients stand in a hip rotation posture that is equivalent to the hip rotation position used in gait (McMulkin, Gordon, Walter, & Griffin, 2009). This requires clinicians to have a prior knowledge of the hip rotation in gait before analysing, which can be very difficult because patients may use different strategies in standing and in walking posture.

Several non-invasive methods can be used to determine functional joint centers and axes of rotation relative to marker clusters without the need to accurately locate anatomical landmarks (Cappozzo, Catani, Della Croce, & Leardini, 1995; Charlton, Tate, Smyth, & Roren, 2002; Schwartz & Rozumalski, 2005). Additionally, the increase in the number of markers (Point Cluster protocol) can reduce the error of skin movement during locomotion (Alexander & Andriacchi, 2001). However, these procedures necessarily result in long data collection sessions with a multitude of movement repetitions for the functional approach and might not be applicable in daily clinical routine. Patients, particularly children, hardly can stand still for longer periods, wear and walk with a large number of markers because of disability or perform necessary movement amplitudes for a functional approach. Regarding functional method considerations, the physiological ROM of varus/valgus in the knee joint is known to be small, varying between 5° and 10° (Reinschmidt et al., 1997) due to the restrictions imposed by the joint anatomy. However, given this assumption, one must be cautious when applying the functional method in circumstances where varus/valgus laxity of the knee exists and actual varus/valgus movement is physiologically possible (Schache, Baker, & Lamoreux, 2006). Such circumstances do not suggest the application of the dynamic method. Furthermore, it has been shown that a full amplitude movement during a functional calibration of the hip joint center is better than restricted one (Sangeux, Peters, & Baker, 2009), which could be a problem for patients with reduced hip ROM or joint contractures of the lower body.

The Knee Alignment Device (KAD) method (Davis & DeLuca, 1996) or a Knee Center Device (KCD) was introduced to better define rotation axes (Schache et al., 2006), which reduces cross-talk error from axis malalignment and slightly improves these measurements. It does this by placing a constant angle offset on the axis throughout a dynamic trial, based on a static trial, which accounts for marker misplacement on the thigh and shank. However, with this approach the tester has to subjectively estimate of where the knee joint flexion/extension axis is perceived to lie based on visualisation alone. In this light, a major disadvantage of the KAD is that it is highly dependent upon the precise identification of axes of rotation by the tester. The consistency of the measures may therefore be influenced by the assessor's experience, expertise, professional background and additional training. Furthermore, it has been shown that a KCD is difficult to handle and less reliable within or between therapists than manual palpation (Serfling, Hooke, Bernhardt, & Kaufman, 2009).

Several recently published studies showed that adding a few extra markers to the standard PiG model enhance the reliability and accuracy of joint rotations (Biagi et al., 2008; Ferrari et al., 2008; Leardini et al., 2007). Although these studies are unique with respect to the choice and number of compared protocols, the overall procedures were repeated in only a small number of subjects varying between 1 and 10 basically healthy subjects. The custom made model (MA) in this study uses additional medial malleolus, medial femoral condyle and trochanter major markers to determine joint centers. This eliminates the reliance on the difficult, subjective palpation of the thigh and tibia wand markers necessary for the PiG model. Furthermore, the MA enables users to detect significant differences in the tibiofemoral angles and moments in the frontal and transverse plane between Nordic walking, walking and running (Stief et al., 2008) and reduces the measurement error of frontal knee angles and moments (Stief et al., 2009). In addition, the new model produces a more accurate and reliable knee joint axis compared to the PiG model, which reduces the measurement error for determining frontal and transverse plane gait data (Stief, Böhm, Michel et al., 2011).

This model was designed to produce more precise and objective joint parameters. Precision, in this context, refers to the reliability and the magnitude of differences stemming from trial-to-trial, or day-to-day variations. The inter-trial reliability measures the sensitivity of the model to subtle differences in motions. Inter-session reliability measures the combined effects of palpation and marker placement. Objectivity refers to a standardized marker position protocol with solely bony orientated anatomical landmarks. Therefore, the purpose of this experimental study was to estimate the reliability and accuracy of two different models for clinical gait analysis. This was achieved by analysing exactly the same gait acquisition for both models. Expected result would be an optimized model for the determination of kinematic and kinetic parameters, being more reliable and accurate than the standard PiG model.

6.2 The Custom Made Lower Body Model – Definitions and Kinematic Procedures

The MA was programmed in Vicon-Bodybuilder (version 3.6) and can be used with the commercial software Vicon-Workstation (version 4.6) or Vicon-Nexus (version 1.4.1). The marker set for the MA included 17 retro-reflective markers (14 mm diameter) placed on standardized, well defined locations as indicated in Figure 26.

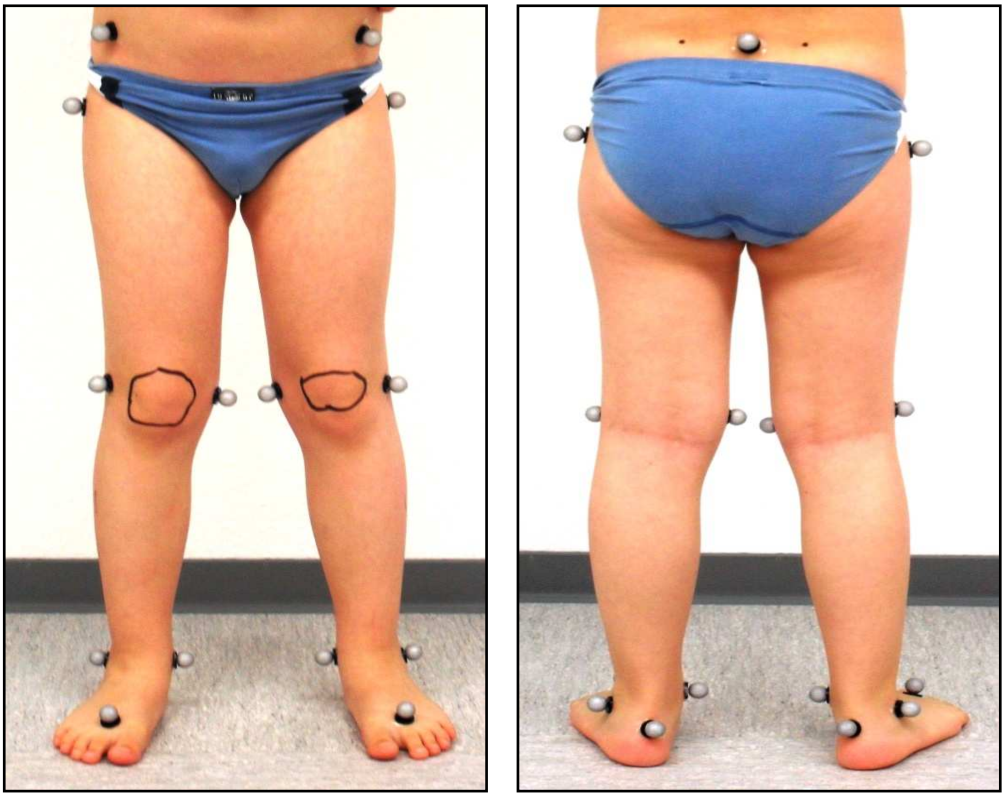


Figure 26. Placement of the markers for the MA. The medial femoral condyle markers were removed for the dynamic trials and the offset from the lateral femoral condyle markers collected during the static trial was used to recalculate the position of the medial markers.

To avoid using thigh and tibia wand markers, additional medial malleolus, medial femoral condyle and trochanter major markers were placed on the subjects to determine joint centers. The trochanter major marker was used to improve the prediction of the hip joint center by immediately calculate the distance between the anterior superior iliac spine and the trochanter major using anatomical landmarks. In the MA, relevant anthropometric measurements on the subject's body were therefore be determined by anatomical landmarks, instead of manual defective measurement of the anthropometric data in the PiG model, which seems to have poor reliability (Alderink, Cobabe, Foster, & Marchinda, 2000).

The location of hip, knee, and ankle joint centers were calculated relative to the associated embedded coordinate system origin. In both models, the center of the hip joint was calculated using a geometrical prediction method (Davis et al., 1991). The location of the hip joint center was calculated relative to the marker-based origin of the pelvic embedded coordinate system. An offset vector was computed from an anthropometric regression equation that was scaled by manually measured distances on the subject's body (PiG) and accordingly by marker-based locations (MA):

$$X_H = (-x_{dis} - r_{marker})\cos(\beta) + C\cos(\theta)\sin(\beta)$$
$$Y_H = S\left[C\sin(\theta) - \frac{d_{ASIS}}{2}\right]$$
$$Z_H = (-x_{dis} - r_{marker})\sin(\beta) - C\cos(\theta)\cos(\beta)$$

where: d_{ASIS} = distance (in meters) between the left and right pelvic anterior superior iliac spine,

x_{dis} = anteroposterior component of the distance (in meters) between the supe– rior iliac spine and the hip joint center in the sagittal plane of the pelvis,

r_{marker} = marker radius (in meters), and

S = +1 for the right side, and -1 for the left side.

The following anatomical parameters were established according to the radiographic examination of 25 hip studies (Davis et al., 1991):

$$\theta = 28.4(\pm 6.6)^\circ$$
$$\beta = 18(\pm 4)^\circ$$
$$C = 0.115 * \text{ leg length} - 0.0153$$

The basis for the hip joint centering algorithm is shown in Figure 27.

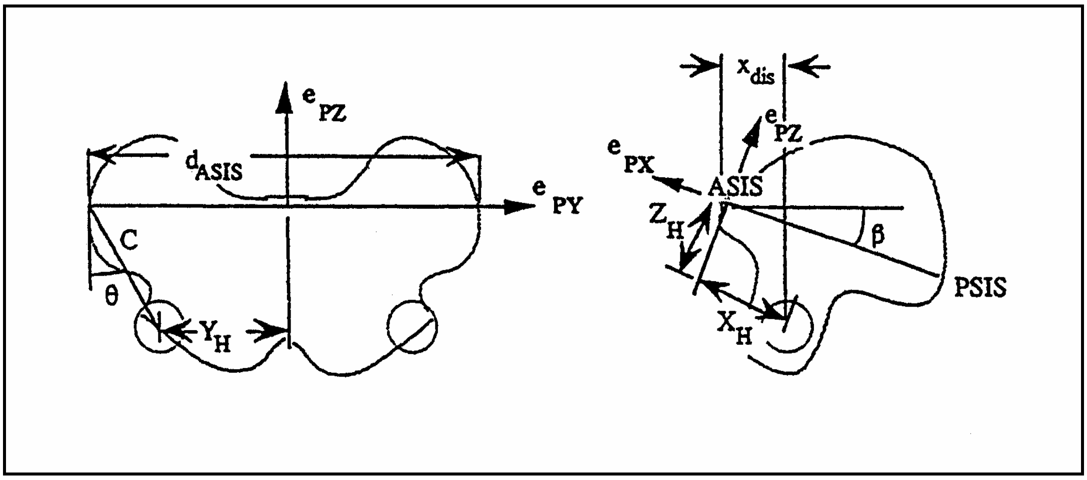


Figure 27. Hip joint centering geometry. Coronal plane view on the left side, sagittal plane view on the right side. ASIS = Distance between right and left pelvic anterior superior iliac spine; PSIS = Posterior superior iliac spine. Adapted from "A gait analysis data collection and reduction technique," by R. B. Davis, S. Ounpuu, D. Tyburski, and J. R. Gage, 1991, Human Movement Science, 10, p. 583.

The PiG model derived the rotational axis of the knee joint from the position of the pelvic, knee and thigh markers and the rotational axis of the ankle joint from the position of the knee, ankle and tibia markers. The location of the knee joint center was calculated relative to the origin of the marker-based thigh embedded coordinate system (located at the knee marker) in thigh coordinates, and the ankle joint center location relative to the origin of the marker-based shank embedded coordinate system (located at the ankle marker) in shank coordinates.

The knee joint center location is based on the coronal plane knee width measurement, obtained during the patient examination:

$$X_K = 0$$
$$Y_K = S(r_{marker} + 0.5\, w_{Knee})$$
$$Z_K = 0$$

where: r_{marker} = marker radius (in meters),

S = +1 for the right side, and -1 for the left side, and

w_{Knee} = knee width measurement (in meters).

The location of the ankle center employs the same strategy that is used for the knee center location.

In contrast to the PiG model, the centers of the knee and ankle joints using the MA were defined as the midpoint between the medial and lateral femoral condyle and malleolus markers rather than calculated using the lateral thigh and tibia markers. It has been shown that the location of the knee center using two condyle markers vary by only a few millimetres from the computer tomography knee center (Kornaropoulos et al., 2010). The medial femoral condyle marker was removed for the dynamic trials and the offset from the lateral femoral condyle collected during the static trial was used to recalculate its position.

Intersegmental angles were calculated from the position of the markers. Each limb segment (thigh, shank and foot) was idealized as a rigid body with a local coordinate system that was defined to coincide with a set of anatomic axes (Figure 28).

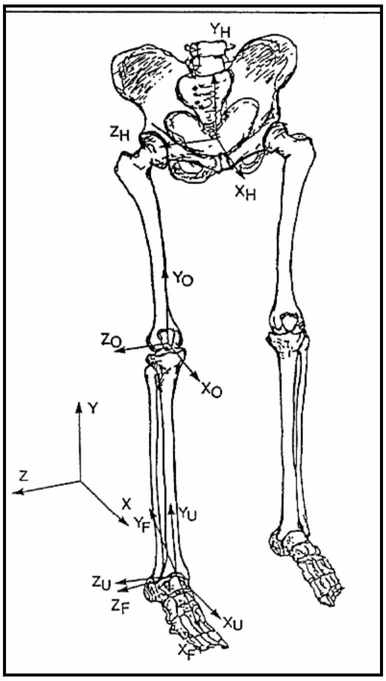


Figure 28. Local coordinate systems of each segment. Adapted from “Zur Belastung des Bewegungsapparates beim Laufen. Einfluss von Laufschuh und Lauftechnik,“ by B. Krabbe, 1994, Aachen: Shaker, p. 55.

The angles at the knee were resolved into a coordinate system fixed in a tibial reference system, with axes defining flexion-extension, abduction-adduction, and internal-external rotation. Similarly, the angles at the hip were resolved into a coordinate system fixed in a thigh reference system, with axes defining flexion-extension, abduction-adduction, and internal-external rotation. The angles at the ankle were resolved into a coordinate system fixed in a foot reference system, with axes defining dorsiflexion-plantarflexion.

6.3 The Custom Made Lower Body Model – Kinetic Procedures

The ground reaction force is basically the reaction to the force the body exerts on the ground and the precondition for the calculation of joint moments. It is measured at 1000 Hz by two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA). The schematic representation of measured variables are given in Figure 29.

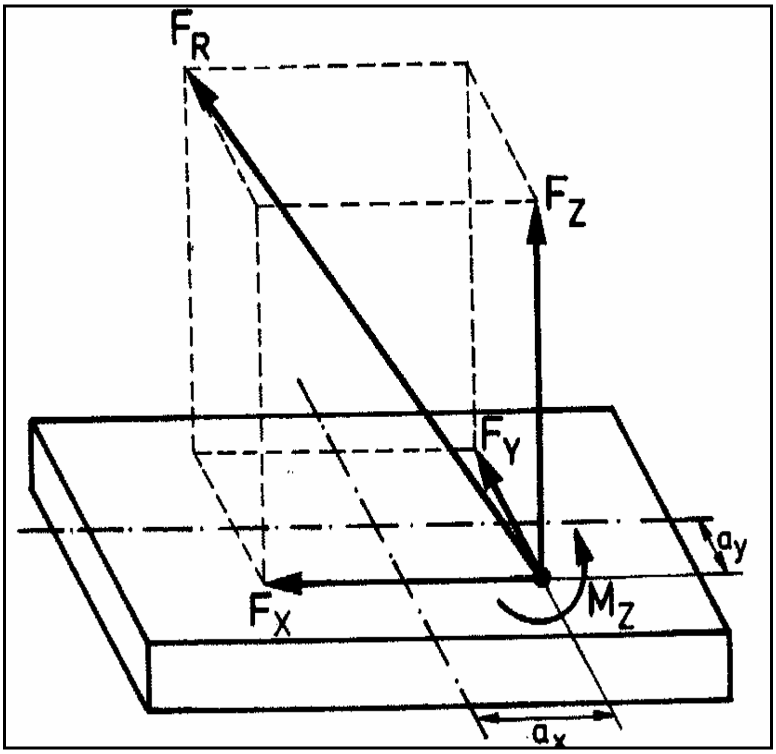


Figure 29. Force plate with schematic representation of measured variables. F_X, F_Y = reaction force in horizontal direction; F_Z = reaction force in vertical direction; M_Z = moment in vertical direction; a_x, a_y = location of force vector relative to the middle of the force plate. Adapted from "Biomechanische Meßverfahren," by W. Baumann and R. Preiß, 1996, In: R. Ballreich & W. Baumann (Eds.), *Grundlagen der Biomechanik des Sports* (p. 98). Stuttgart: Enke.

Three-dimensional external net joint moments at the hip, knee, and ankle were calculated through inverse dynamics as vector product of the position vector of the joint center and the collected ground reaction force along with the inertial properties of the limb and the angular and linear velocities of the segments of the lower extremity. The mass of each segment was calculated as a percentage of body weight. The parameters necessary for joint moment calculations for a particular body segment during walking are listed in Table 3, while the global parameters necessary for the limb as a whole are given in Table 4.

Table 3
Parameters Necessary to Calculate Joint Moments for a Particular Segment

1. Mass
2. Principal inertial tensor components
3. Position of mass center
4. Linear velocity of mass center
5. Linear acceleration of mass center
6. Angular position
7. Angular velocity
8. Angular acceleration

Note. Adapted from "Lower extremity joint moments and ground reaction torque in adult gait," by K. K. Ramakrishnan, M. P. Kadaba, and M. E. Wootten, 1987, In: J. L. Stein (Ed.), *Biomechanics of Normal and Prosthetic Gait* (p. 88). West Haverstraw: Helen Hayes Hospital.

Table 4
Global Parameters Necessary for the Limb as a Whole

1. The position of joint centers
2. Principal embedded coordinates
3. External ground reaction forces
4. Center of pressure
5. Screw torque on the forceplate

Note. Adapted from "Lower extremity joint moments and ground reaction torque in adult gait," by K. K. Ramakrishnan, M. P. Kadaba, and M. E. Wootten, 1987, In: J. L. Stein (Ed.), *Biomechanics of Normal and Prosthetic Gait* (p. 88). West Haverstraw: Helen Hayes Hospital.

The basis for the calculation of joint moments is the Newton's Second Law and Euler's equations of motion (Greenwood, 1965):

$$M_x = I_x \alpha_x + (I_z - I_y)\omega_y \omega_z$$
$$M_y = I_y \alpha_y + (I_x - I_z)\omega_z \omega_x$$
$$M_z = I_z \alpha_z + (I_y - I_x)\omega_x \omega_y$$

where: M_x, M_y, M_z = components of the sum of the external moments (about the center of gravity of the segment) applied to the limb segment,

$\alpha_x, \alpha_y, \alpha_z$ = components of the absolute segmental angular acceleration,

I_x, I_y, I_z = principal mass moments of inertia of the segment,

$\omega_x, \omega_y, \omega_z$ = components of the absolute segmental angular velocity, and

x, y, z = body-fixed coordinate axes, defined as the principal axes and located at the center of mass of the segment.

These moments in the principal directions are converted into the global x-y-z directions using appropriate coordinate transformations.

For example, the ankle reaction moment vector, M_A, and the force reaction vector, F_A, (Figure 30), may be determined with the appropriate kinematic information, e.g., joint center location (point A), center of pressure coordinates (point G), and center of gravity location (point CG), as well as the external load applied to the foot, i.e., the weight of the foot, *mg*, the ground force reaction, F_r and the vertical torque, *T* (Davis et al., 1991).

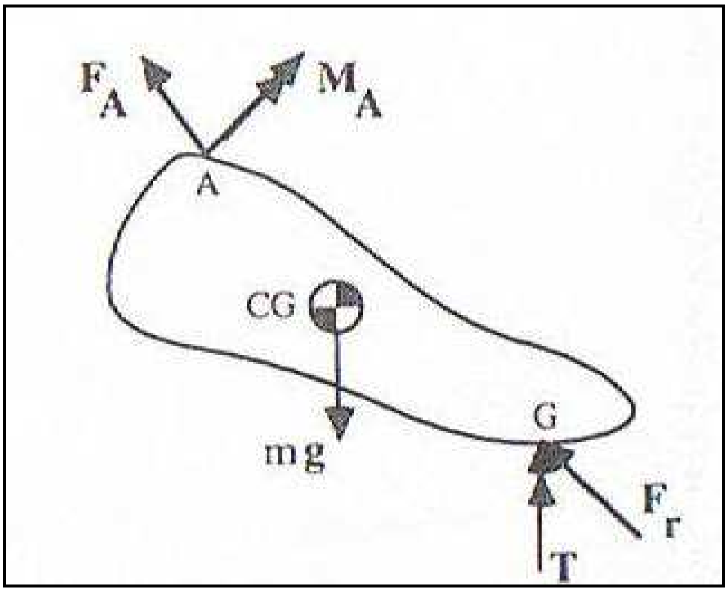


Figure 30. Free-body diagram of the foot segment used to determine force and moment reactions at the ankle joint. Adapted from “A gait analysis data collection and reduction technique,” by R. B. Davis, S. Ounpuu, D. Tyburski, and J. R. Gage, 1991, *Human Movement Science, 10*, p. 586.

In this way, the proximal joint reactions of each segment may be determined using the distal reaction results. Knee joint moments are determined from the forces and moments at the ankle joint plus the inertial and gravitational forces at the mass center of the shank segment. The hip joint calculations are performed in a similar way (Ramakrishnan, Kadaba, & Wootten, 1987). A flow chart of calculation sequence is shown in Figure 31.

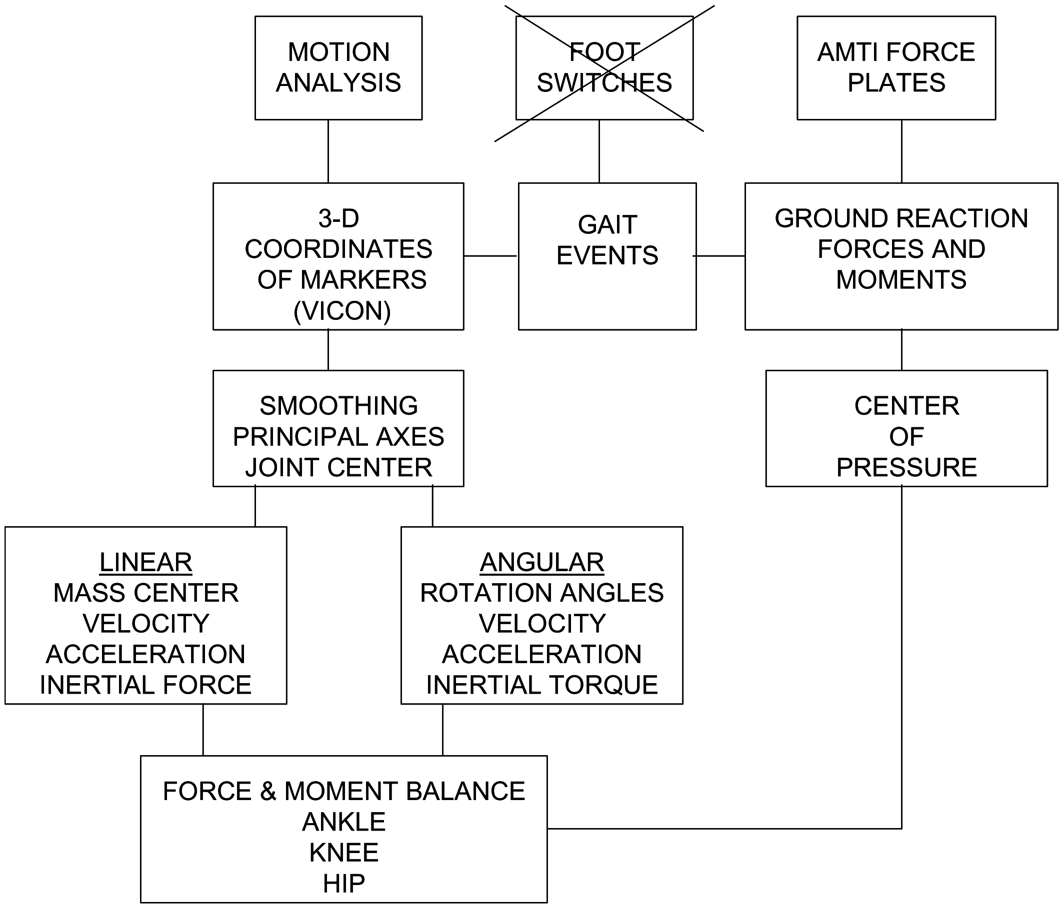


Figure 31. Calculation sequence of joint moment determination. Adapted from “Lower extremity joint moments and ground reaction torque in adult gait,” by K. K. Ramakrishnan, M. P. Kadaba, and M. E. Wootten, 1987, In: J. L. Stein (Ed.), *Biomechanics of Normal and Prosthetic Gait* (p. 88). West Haverstraw: Helen Hayes Hospital.

6.4 Reliability and Accuracy of the Custom Made Lower Body Model – Comparison With the Standard Plug-in-Gait Model

A single comprehensive marker set was defined allowing the use of exactly the same gait cycles for the PiG model (Davis et al., 1991) without the KAD and the MA.

6.4.1 Material and methods

6.4.1.1 Subjects

Twenty-five subjects with a mean age (with standard deviation in parenthesis) of 14.9 (3.8) years, a mean height of 1.63 (0.16) m and a mean weight of 52.4 (13.7) kg were gait analyzed by the same experienced examiner to test the inter-trial reliability. This subject group consisted of 14 healthy volunteers and 11 patients with pathological varus alignment based on a full length standing AP radiograph (Paley, 2002). Anthropometric data for both groups are shown in Table 5. No significant differences regarding age, height and weight were detected between the two groups (Table 5). The healthy control group had normal strength, full range of motion of the lower extremities, and no knee instability or neurologic deficits. The patient group with pathological varus alignment had no other pathologies or orthopedic problems and none of the patients had knee laxity. Therefore, patients with isolated varus malalignment were deliberately included in the study to test the suitability of both models for this patient group.

Table 5
Mean (With Standard Deviation in Parenthesis) and p-Values for Anthropometric Parameters of Patients With Isolated Pathological Varus Alignment and Healthy Volunteers

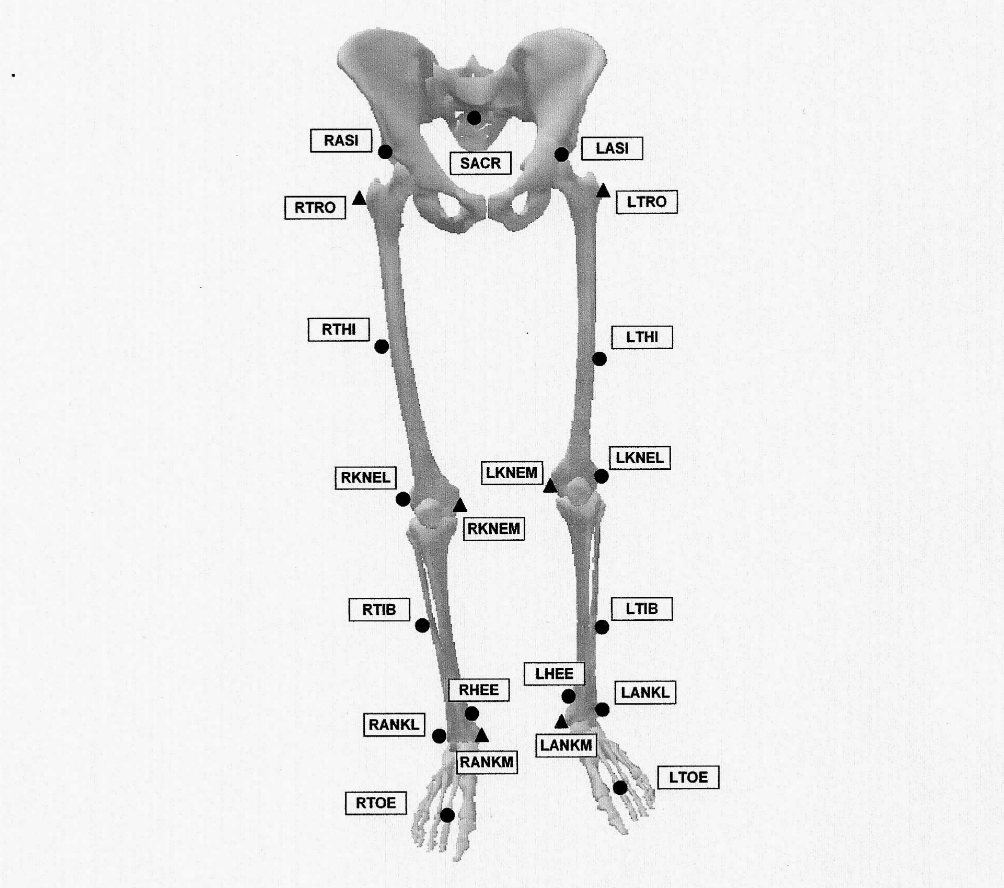
Anthropometric Parameter	Healthy Volunteers (*n* = 14)	Patients (*n* = 11)	*p*-Value
Sex, no. female/no. male	9/5	8/3	.73
Age, years	15.5 (4.2)	14.2 (3.1)	.40
Height, m	1.65 (0.15)	1.62 (0.19)	.67
Weight, kg	53.8 (13.3)	50.6 (14.6)	.58

Note. *p*-values (significance level $p < .05$) are based on parametric independent Student's *t*-test, except for the sex distribution, which is compared by non-parametric Mann-Whitney U test.

On two different sessions separated by at least three days and within two weeks, inter-session reliability within clinician was examined on ten healthy volunteers with a mean age of 20.4 (10.1) years, a mean height of 1.67 (0.17) m and a mean weight of 58.5 (17.4) kg. All subjects and/or their parents were thoroughly familiarized with the gait analysis protocol before giving informed consent to participate in this study.

6.4.1.2 Experimental procedure and data analysis

Kinematic data were collected using an 8-camera Vicon motion capture system operating at a sampling rate of 200 Hz (VICON Motion Systems, Oxford, UK). Two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to collect kinetic data at 1000 Hz. Kinematic and ground reaction force data were collected simultaneously. Data acquisitions were carried out in the presence of the same expert, who performed anthropometric measurements, landmark identification, and marker placement. A single comprehensive marker set allowed the use of the same gait cycles for both models. This marker set included 21 retro-reflective markers (14 mm diameter) on the pelvis, thigh, shank and foot as outlined in Figure 32.



Abbreviation	Placement	Required for protocol
SACR	On the skin mid-way between the posteior superior iliac spines	PiG / MA
LASI (RASI)	On the left (right) anterior superior iliac spine	PiG / MA
LTRO (RTRO)	On the prominent point of the left (right) trochanter major	MA
LTHI (RTHI)	Rigid wand marker mounted on the skin over the distal and lateral aspect of the left (right) thigh aligned in the plane that contains the hip and knee joint centers and the knee flexion/extension axis	PiG
LKNEL (RKNEL)	On the left (right) lateral femoral condyle	PiG / MA
LKNEM (RKNEM)	On the left (right) medial femoral condyle	MA
LTIB (RTIB)	Rigid wand marker mounted on the skin over the distal and lateral aspect of the left (right) shank aligned in the plane that contains the knee and ankle joint centers and the ankle flexion/extension axis	PiG
LANKL (RANKL)	On the left (right) lateral malleolus aligned with the bimalleolar axis	PiG / MA
LANKM (RANKM)	On the left (right) medial malleolus aligned with the bimalleolar axis	MA
LTOE (RTOE)	Ont the left (right) second metatarsal head, on the mid-foot sides of the equinus break between fore-foot and mid-foot	PiG / MA
LHEE (RHEE)	On the left (right) aspect of the Achilles tendon insertion, on the calcaneous at the same height above the plantar surface of the foot as the LTOE (RTOE) marker	PiG / MA

Figure 32. Marker set MA and PiG model. The markers indicated by circles are part of the standard PiG marker set; those indicated by triangles are the additional markers used in the MA.

After the subjects completed a standardized questionnaire (Figure A1 and A2 in the appendix) and a clinical examination (Figure A3 in the appendix) to test the exclusion criteria, they were asked to walk barefoot at their self-selected speed without targeting the force plates. Eight walking trials were recorded across a 15 m gait laboratory walkway for each subject (Figure 33).

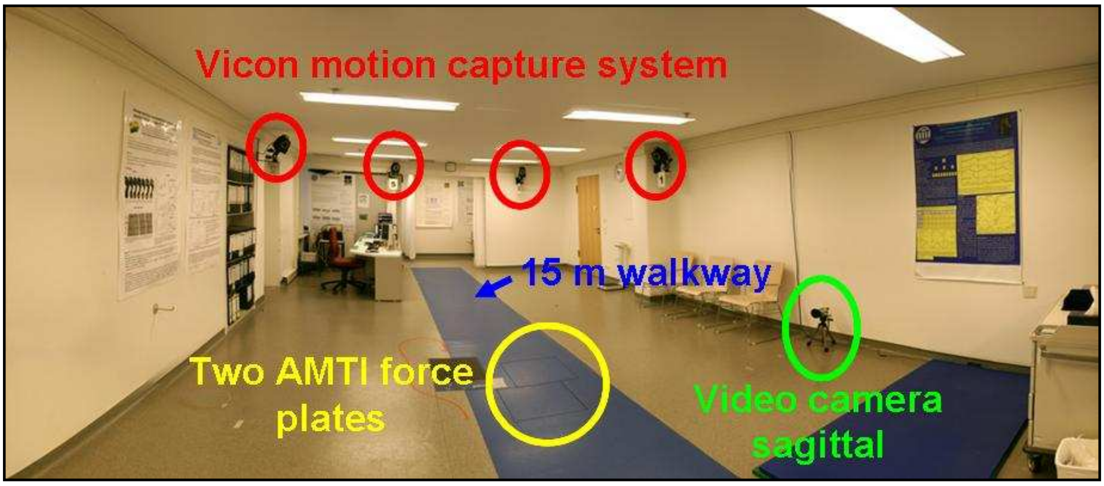


Figure 33. Gait laboratory.

After each acquisition session, 3D marker trajectories were reconstructed and missing frames were handled with a fill-gap procedure. The average values from five trials were selected on the basis of good quality of the marker trajectories and ground reaction forces. The data were smoothed with a Woltring filter and using a spline smoothing (Woltring, 1991). Both models were filtered identically. This ensured that any differences could be attributed to the models. Forty-five selected values of high clinical relevance (Table 6), for example maximum knee adduction in stance phase, were automatically determined from each trial by a custom made algorithm in Matlab 7.3.0 (The MathWorks, Inc., Natick, MA, USA). Data were normalized to time (0-100% gait cycle). Regarding joint moments, the convention adopted was that of an external moment, i.e. the resultant moment of the external forces.

6.4.1.3 Accuracy and statistical analysis

The accuracy of the models could not be assessed since the true joint parameters were not measured. Instead, the following indirect indicators of accuracy were computed:

- Knee varus/valgus ROM during gait: An accurate knee flexion axis minimizes the varus/valgus ROM resulting from cross-talk (Piazza & Cavanagh, 2000).
- Knee flexion/extension ROM during gait: An accurate knee flexion axis maximizes knee flexion/extension ROM by reducing cross-talk (Schwartz & Rozumalsky, 2005).

These assumptions are supported by a previous in vivo bone pin study (Ramsey & Wretenberg, 1999), which has measured the normal 3D angular rotations at the knee joint. It has been shown that for the stable knee joint, the physiological ROM of knee varus/valgus only varies between 5° and 10° (Reinschmidt et al., 1997). Since in this study no knee laxity exists in patients with varus malalignment, this assumption is also effective for the patient group. Minimization of the knee joint angle cross-talk was therefore considered to be a valid criterion to evaluate the relative merits of the two models.

Two forms of intra-observer difference procedures were undertaken: 1) Intertrial reliability, i.e. 25 subjects were analyzed with both models by the same examiner over 5 valid walking repetitions; 2) inter-session reliability, i.e. 10 healthy subjects were analyzed with both models by the same examiner over 5 valid walking repetitions on two different dates.

Inter-trial reliability was assessed using the Intraclass Correlation Coefficient (ICC; model 2,1). To provide an absolute measure of reliability, the root mean square error (RMSE) was used for the inter-trial SD and the technical error of measurement (TEM; Perini, de Oliveira, G. L., Ornellas, & de Oliveira, F. P., 2005) for the inter-session SD. Regarding the RMSE and the TEM, the relative values normalized by the ROM of the corresponding mean curve (RMSE%, TEM%) were reported for each variable. The inter-trial SD measures the sensitivity of the method to subtle differences in motions. Inter-session SD measures the combined effect of the new method, palpation, and marker placement.

The shape of distribution of the present sample was checked using the Kolmogorov-Smirnov test. Because a normal distribution was confirmed in the present study, differences in knee varus/valgus and flexion/extension ROM between the models were tested for significance using a paired-sample *t*-test. The effect size (Cohen's *d*) of the results was interpreted according to Cohen (1988). The significance level adopted in this study was set at $p < .05$ and the statistical calculations were carried out using the SPSS version 12.0.1 (Chicago, IL, USA) and Matlab 7.3.0 (The MathWorks, Inc., Natick, MA, USA).

6.4.2 Results

6.4.2.1 Inter-trial reliability

Table 6 shows a small inter-trial variability for both models with absolute and relative RMSE values of less than 3.3° and 16%, respectively for ankle, knee, hip and pelvic kinematic and of less than 0.27 Nm/kg and 14%, respectively for ankle, knee and hip kinetic results.

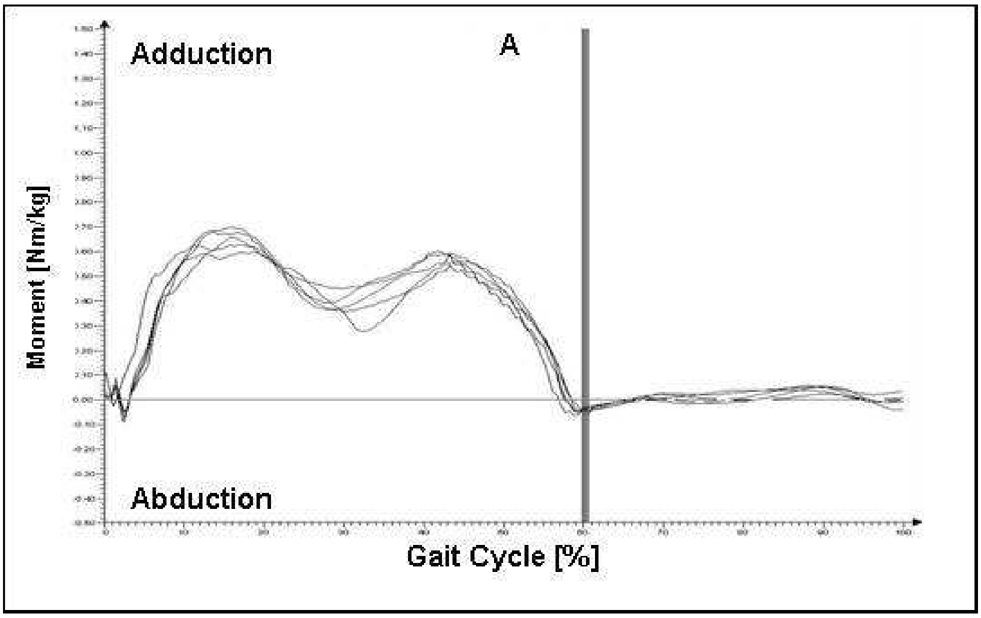
Table 6

Inter-Trial Variability (n = 25) Across Each Joint Rotation and Moment Variables for Both Models

			RMSE		RMSE%		ICC	
Plane	Segment	Rotations (°)	MA	PiG	MA	PiG	MA	PiG
Sagittal	Ankle	Min. plantarflexion loading response	1.21	1.35	4.48	4.26	.908	.896
		Max. dorsiflexion terminal stance	1.06	1.14	3.93	3.60	.963	.948
		Min. plantarflexion pre-swing/initial swing	2.39	2.53	8.83	8.00	.924	.881
		Max. dorsiflexion mid swing	1.16	1.31	4.28	4.14	.963	.936
		Range of Motion dorsi/plantarflexion gait cycle	2.23	2.43	8.26	7.68	.740	.764
	Knee	Max. flexion loading response/mid stance	2.00	2.00	3.13	3.43	.913	.908
		Min. extension terminal stance	1.76	1.51	2.79	2.60	.944	.950
		Max. flexion initial swing/mid swing	1.76	1.80	2.80	3.09	.940	.935
		Range of Motion flexion/extension gait cycle	1.96	2.05	3.11	3.52	.847	.906
	Hip	Min. extension terminal stance/pre-swing	1.24	1.18	2.80	2.53	.985	.978
		Max. flexion mid swing/terminal swing	1.19	1.35	2.70	2.91	.981	.968
		Range of Motion flexion/extension gait cycle	1.80	1.77	4.07	3.81	.874	.884
	Pelvic	Max. anterior tilt contact phase	1.01	0.90	15.10	10.46	.969	.977
		Max. anterior tilt swing phase	0.89	0.92	13.25	10.76	.974	.974
		Range of Motion anterior/posterior tilt gait cycle	1.02	0.89	15.19	10.37	**.420**	**.469**
Frontal	Knee	Max. adduction stance phase	1.21	1.02	8.20	4.80	.954	.963
		Max. adduction swing phase	1.34	1.64	9.07	7.69	.946	.965
		Range of Motion adduction/abduction gait cycle	1.45	2.49	9.78	11.70	.926	.946
	Hip	Max. adduction stance phase	1.01	1.10	7.20	7.55	.944	.936
		Min. abduction swing phase	1.24	1.44	8.87	9.87	.875	.860
		Range of Motion adduction/abduction gait cycle	1.55	1.74	11.06	11.98	.852	.825
	Pelvic	Max. obliquity up stance phase	0.70	0.79	6.77	7.77	.937	.917
		Min. obliquity down swing phase	0.78	0.87	7.54	8.49	.856	.841
		Range of Motion obliquity up/down gait cycle	1.15	1.30	11.17	12.76	.898	.869
Transverse	Knee	Max. internal rotation stance phase	2.12	1.77	6.05	7.44	.970	.973
		Min. external rotation swing phase	2.93	2.00	8.38	8.24	.946	.931
		Range of Motion internal/external rotation gait cycle	2.99	2.28	8.55	9.58	.861	.838
	Hip	Max. internal rotation stance phase	1.30	2.48	6.24	7.45	.981	.961
		Min. external rotation stance phase	1.59	2.51	7.64	7.54	.952	.973
		Range of Motion internal/external rotation gait cycle	2.01	3.23	9.64	9.68	.844	.852
	Pelvic	Max. internal rotation contact phase	1.98	1.62	9.39	11.83	.861	.906
		Min. external rotation contact phase	2.14	2.01	10.11	14.68	.800	.837
		Range of Motion internal/external rotation gait cycle	2.57	2.14	12.15	15.64	.892	.931
		Moments (Nm/kg)						
Sagittal	Ankle	Max. dorsiflexion terminal stance	0.05	0.04	3.48	3.15	.932	.960
	Knee	Max. flexion loading response/mid stance	0.08	0.08	10.95	10.38	.768	.877
		Max. extension terminal stance	0.05	0.05	6.39	6.58	.865	.962
	Hip	Max. flexion loading response	0.20	0.27	10.32	13.16	**.639**	**.442**
		Max. extension terminal stance	0.06	0.12	3.19	5.97	.931	.724
Frontal	Knee	Max. adduction mid stance	0.03	0.06	6.68	11.45	.973	.893
		Max. adduction terminal stance	0.03	0.03	5.98	6.62	.978	.964
	Hip	Max. adduction mid stance	0.05	0.07	4.48	8.37	.838	**.653**
		Max. adduction terminal stance	0.04	0.05	3.31	5.64	.933	.939
Transverse	Knee	Max. internal rotation terminal stance	0.02	0.02	9.39	9.07	.935	.910
	Hip	Max. external rotation loading response/mid stance	0.02	0.02	6.38	7.61	.912	.806
		Max. internal rotation terminal stance	0.01	0.02	5.23	5.41	.904	.897

Note. Low correlations (ICC < .70) are in boldface. RMSE = absolute root mean square error; RMSE% = relative root mean square error normalized by the range of motion of the corresponding mean curve; ICC = Intraclass Correlation Coefficient; MA = custom made model; PiG = Plug-in-Gait marker set.

However, the relative RMSE exhibited increased variability in 9 of 11 knee and hip kinetic parameters apart from maximum knee flexion and internal rotation moment measured with the standard PiG model (Table 6). Figure 34 shows typical graphs of inter-trial variability on the basis of the knee moment in the frontal plane in one subject for both models.



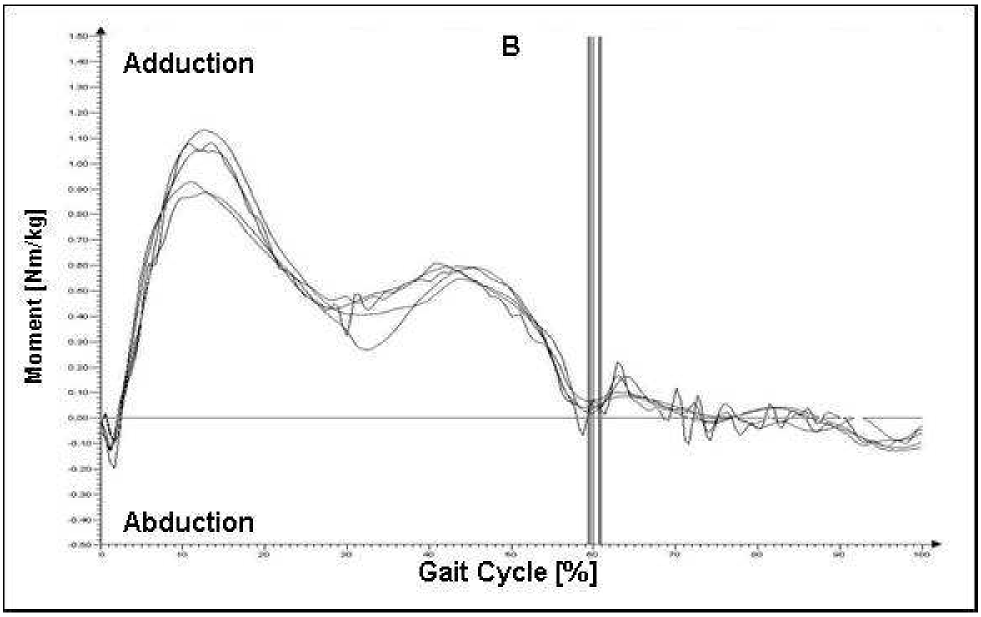


Figure 34. Typical inter-trial variability (5 trials) of the knee moment in the frontal plane in one subject for the custom made model (MA; A) and the Plug-in-Gait (PiG) model (B). The vertical lines each identify the end of the stance phase. The same gait cycles were analyzed for both models.

The corresponding ICC values were evaluated according to Vincent (1999). For 42 of 45 parameters they were between .985 and .724 in both models (Table 6) indicating excellent to moderate reliability apart from pelvic ROM in sagittal plane (ICC MA = .420; ICC PiG = .469), hip maximum flexion moment (ICC MA = .639; ICC PiG = .442) and hip maximum adduction moment during mid stance (ICC PiG = .653) indicating poor reliability.

6.4.2.2 Inter-session reliability

Both models showed similarly good inter-session reliability for all ankle, knee and hip joint flexion/extension angles and hip angles in the frontal plane (TEM < 2.6°; TEM% < 8.2%; Table 7). Regarding frontal plane knee motion, the mean (with standard deviation in parenthesis) absolute and relative TEM showed higher values for the PiG (TEM = 2.5° (0.3); TEM% = 15% (0.4)) compared to the MA (TEM = 1.6° (0.2); TEM% = 11% (0.3)). The knee, hip and pelvic joint angles in the transverse plane also revealed higher mean measurement errors for the PiG (TEM = 3.7° (2.0); TEM% = 17% (5.3)) compared to the MA (TEM = 2.8° (1.4); TEM% = 12% (5.4)). The sagittal plane ankle, knee and hip joint moments showed higher mean absolute and relative TEM values for the PiG (TEM = 0.10 Nm/kg (0.07); TEM% = 7.4% (3.26)) compared to the MA (TEM = 0.08 Nm/kg (0.05); TEM% = 5.8% (3.09)). Furthermore, the mean inter-session reliability in frontal plane knee and hip joint moments was better using the MA (TEM = 0.048 Nm/kg (0.02); TEM% = 6.3% (1.76)) compared to the PiG (TEM = 0.063 Nm/kg (0.02); TEM% = 8.9% (2.93)). In the transverse plane the measurement errors for knee and hip joint moments were similar for both models.

Table 7
Inter-Session Variability (n = 10) Across Each Joint Rotation and Moment Variables for Both Models

			TEM		TEM%	
Plane	Segment	Rotations (°)	MA	PiG	MA	PiG
Sagittal	Ankle	Min. plantarflexion loading response	1.59	2.17	5.12	6.33
		Max. dorsiflexion terminal stance	1.93	1.03	6.22	3.00
		Min. plantarflexion pre-swing/initial swing	2.13	2.43	6.84	7.11
		Max. dorsiflexion mid swing	1.87	1.28	6.00	3.73
		Range of Motion dorsi/plantarflexion gait cycle	1.69	2.17	5.43	6.35
	Knee	Max. flexion loading response/mid stance	1.73	1.30	2.78	2.17
		Min. extension terminal stance	1.85	1.66	2.98	2.77
		Max. flexion initial swing/mid swing	1.70	1.94	2.73	3.25
		Range of Motion flexion/extension gait cycle	1.41	1.68	2.27	2.81
	Hip	Min. extension terminal stance/pre-swing	2.19	1.72	5.03	3.79
		Max. flexion mid swing/terminal swing	1.97	1.86	4.53	4.10
		Range of Motion flexion/extension gait cycle	2.60	2.28	5.98	5.02
	Pelvic	Max. anterior tilt contact phase	0.97	0.93	17.28	12.96
		Max. anterior tilt swing phase	1.17	1.18	20.86	16.43
		Range of Motion anterior/posterior tilt gait cycle	0.41	0.47	7.23	6.50
Frontal	Knee	Max. adduction stance phase	0.49	1.49	3.49	8.95
		Max. adduction swing phase	2.60	3.88	18.65	23.28
		Range of Motion adduction/abduction gait cycle	1.60	2.21	11.52	13.24
	Hip	Max. adduction stance phase	0.70	0.80	5.00	5.60
		Min. abduction swing phase	1.14	0.97	8.17	6.83
		Range of Motion adduction/abduction gait cycle	1.10	0.97	7.88	6.83
	Pelvic	Max. obliquity up stance phase	0.86	0.86	8.34	8.61
		Min. obliquity down swing phase	1.42	1.42	13.83	14.29
		Range of Motion obliquity up/down gait cycle	1.36	1.47	13.51	14.80
Transverse	Knee	Max. internal rotation stance phase	4.24	5.59	12.79	25.39
		Min. external rotation swing phase	2.89	5.16	8.72	23.41
		Range of Motion internal/external rotation gait cycle	3.67	2.55	11.05	11.56
	Hip	Max. internal rotation stance phase	3.81	5.94	18.27	20.20
		Min. external rotation stance phase	4.02	5.40	19.24	18.34
		Range of Motion internal/external rotation gait cycle	3.41	4.50	16.33	15.28
	Pelvic	Max. internal rotation contact phase	1.07	1.33	5.76	12.14
		Min. external rotation contact phase	0.80	1.19	4.29	10.90
		Range of Motion internal/external rotation gait cycle	1.46	1.53	7.82	14.02
		Moments (Nm/kg)				
Sagittal	Ankle	Max. dorsiflexion terminal stance	0.04	0.06	2.60	4.38
	Knee	Max. flexion loading response/mid stance	0.08	0.08	10.49	9.47
		Max. extension terminal stance	0.04	0.09	4.75	10.00
	Hip	Max. flexion loading response	0.15	0.22	7.18	9.69
		Max. extension terminal stance	0.09	0.07	4.07	3.25
Frontal	Knee	Max. adduction mid stance	0.04	0.08	8.58	13.17
		Max. adduction terminal stance	0.03	0.03	6.68	6.53
	Hip	Max. adduction mid stance	0.06	0.07	4.92	8.06
		Max. adduction terminal stance	0.06	0.07	4.86	7.78
Transverse	Knee	Max. internal rotation terminal stance	0.02	0.02	9.35	8.49
	Hip	Max. external rotation loading response/mid stance	0.01	0.02	4.68	5.61
		Max. internal rotation terminal stance	0.02	0.02	7.01	7.00

Note. TEM = absolute technical error of measurement; TEM% = relative technical error of measurement normalized by the range of motion of the corresponding mean curve; MA = custom made model; PiG = Plug-in-Gait marker set.

Figure 35 shows typical graphs of the mean inter-session variability in one subject for both models on the basis of the knee moment in the frontal plane.

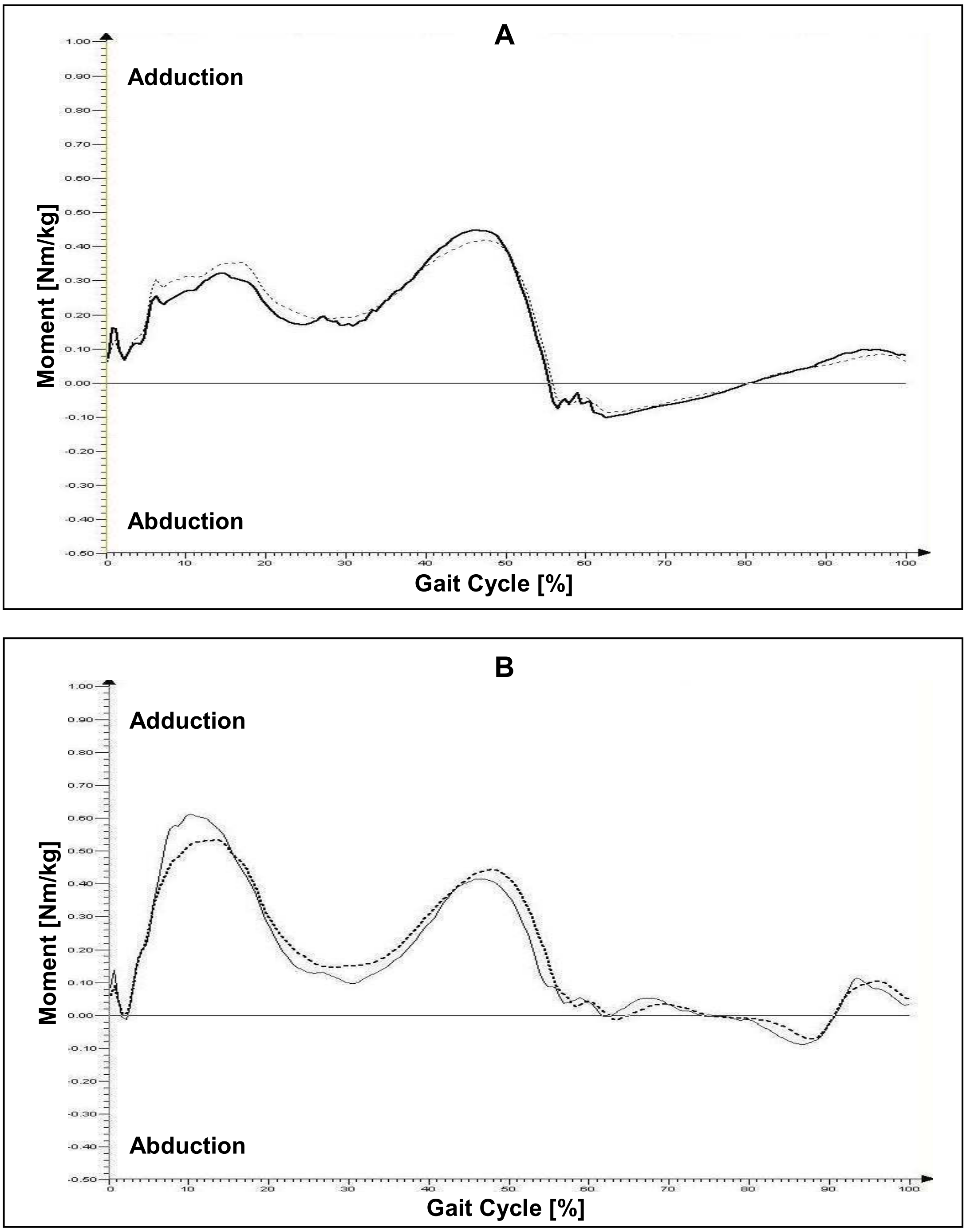


Figure 35. Typical inter-session variability of the knee moment in the frontal plane in one subject for the custom made model (MA; A) and the Plug-in-Gait (PiG) model (B). The continuous graphs indicate the first session mean of 5 trials. The dotted graphs indicate the second session mean of 5 trials. The same gait cycles were analyzed for both models.

6.4.2.3 Accuracy

Knee varus/valgus and flexion/extension ROM during total gait cycle were computed for 125 walking trials (25 subjects x 5 walking trials) using both the MA and standard PiG model. The MA revealed an average knee varus/valgus ROM of 14.8° (5.1) and the PiG model revealed an average knee varus/valgus ROM of 21.3° (10.5). This resulted in 6.5° less knee varus/valgus ROM for the MA (Figure 36). The difference was significant ($p = .002$; Cohen's $d = 0.69$).

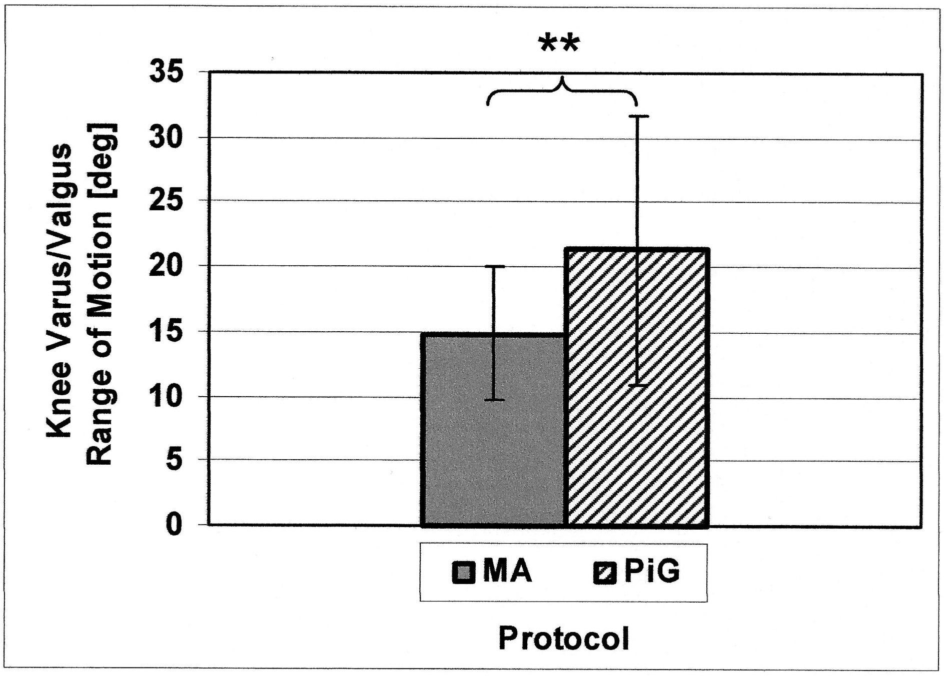


Figure 36. Knee varus/valgus range of motion during total gait cycle. The mean ($n = 25$) and standard deviation is shown.
** $p < .01$.

Regarding knee flexion/extension ROM, the MA revealed an average knee flexion/extension ROM of 62.9° (4.7) and the PiG model revealed an average knee flexion/extension ROM of 58.2° (6.4). This resulted in 4.7° less knee flexion/extension ROM for the PiG model (Figure 37). The difference was also significant ($p < .001$; Cohen's $d = 0.85$).

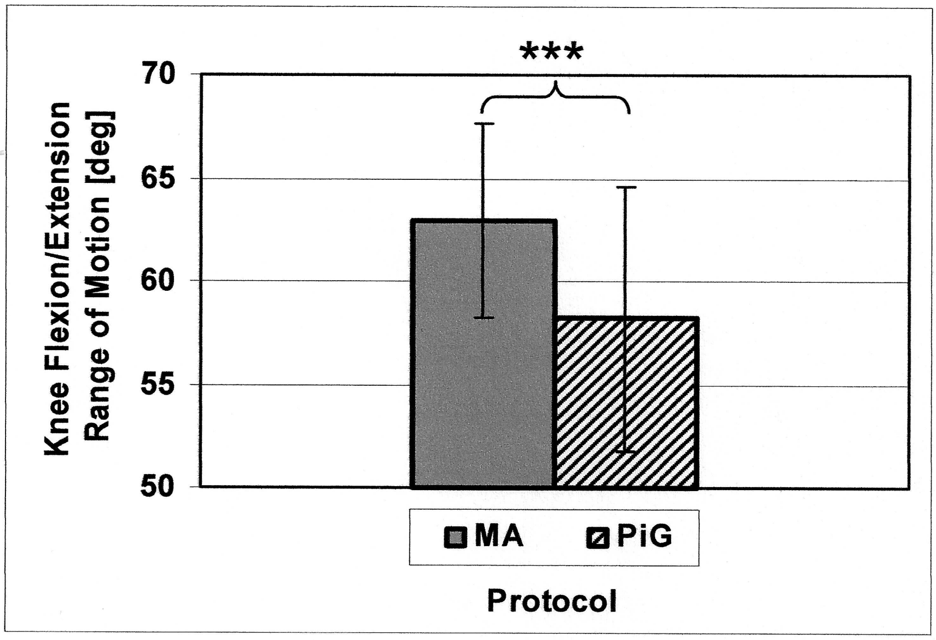


Figure 37. Knee flexion/extension range of motion during total gait cycle. The mean (n = 25) and standard deviation is shown.
*** $p < .001$.

6.4.3 Discussion

In the present study, single gait cycles were analyzed simultaneously by using two different models. These models were evaluated with respect to two main criteria: (1) Within-subject variability with repeated measurements and (2) knee joint angle cross-talk. Previous studies have used one or both criteria for a similar purpose (Gorton et al., 2001; Schache et al., 2006; Schwartz & Rozumalsky, 2005).

The data from the repeated measures experiment indicate that the MA is objective and precise. The inter-trial variability for both models was slightly better than in other studies that reported measurement errors of less than 5°, excluding hip and knee rotation showing even higher errors (McGinley et al., 2009). The increased inter-trial variability of knee and hip kinetic parameters measured with the standard PiG model is not valid for the kinematic parameters.

This might be due to the inverse dynamic process involving the segmental calculation of acceleration, which amplifies the effect of skin motion artifacts of the thigh and tibia wand markers used in the PiG model.

The lower inter-session TEM values for the MA compared to the PiG model regarding frontal plane knee angles and moments and transverse plane motion in the knee and hip joint suggest that the error in repeated palpation of the landmarks is lower using the MA. In the present study, joint kinematics show larger inter-session differences than joint kinetics. This was also observed in other studies (Kirtley, Whittle, & Jefferson, 1985; Lelas, Merriman, Riley, & Kerrigan, 2003). One reason for the reduced TEM for joint moments might be the normalization with body weight leading to a smaller inter-session difference in peak moments compared to the absolute joint angles. The comparison of values in Tables 6 and 7 shows that inter-session errors were only in 51% of the overall parameters slightly greater than inter-trial errors for both models. Nevertheless, the size of the inter-session variability in this experiment could well be substantially worse if tests are performed using different testers or in different laboratories (McGinley et al., 2009). Therefore, further studies also will have to concentrate on inter-laboratory reliability of gait analysis models in typical gait disorder populations.

Indirect measures also indicate that the MA is more accurate compared to the PiG model. In the present experiment, knee joint angle cross-talk referred specifically to the relationship between the knee flexion/extension and varus/valgus kinematic profiles. The MA significantly reduced the knee axis cross-talk phenomenon compared to the PiG model. The effect size of these results was medium to large. These results are comparable to those reported by Schwartz and Rozumalski (2005) using a functional approach in comparison with the standard PiG model. With an understanding of the knee cross-talk phenomenon, these findings demonstrate that the MA produces a more accurate knee joint axis than that derived using the PiG model with wand markers. In the PiG model the alignment of the knee and ankle axes depends on the difficult, non standardized subjective palpation of the thigh and tibia wand markers, which has been shown to have large variability (Gorton et al., 2002) and to enlarge skin motion artifact effects (Manal, McClay, Stanhope, Richards, & Galinat, 2002). As a result, the mean hip rotation is often unacceptably imprecise because the distal thigh wand marker does not fully track the quantity of hip rotation. These results are especially important regarding joint rotations in transverse plane since hip rotation during gait is largely determined by the transverse plane alignment of the knee axis. Moreover, internally rotated gait is a common gait disorder in spastic cerebral palsy patients (Wren, Rethlefsen, & Kay, 2005) and it is an important kinematic variable used in clinical decision making and particularly in planning femoral derotational osteotomies (Ounpuu, DeLuca, Davis, & Romness, 2002). Using the PiG model may therefore lead to hypo- or hypercorrection.

The MA eliminates the reliance on the subjective palpation of the thigh and tibia wand markers and the application of a KAD or KCD. Additionally, this method may reduce errors due to skin motion artifacts, when compared to the thigh and shank markers, by improving bone tracking. The MA marker set appears to be applicable to a wide range of patient populations and to be appealing to clinicians because of the easier skeletal marker definitions. Being based on the identification of anatomical landmarks, examiner training depends on instructions for landmark access by palpation. This would make learning and training of the examiners advantageous and could probably lead to a reduction of intra- and inter-examiner variability. Another noteworthy feature is the immediate calculation of relevant anthropometric measurements on the subject’s body by the anatomical landmarks, instead of manual defective measurement of the anthropometric data, which seems to have poor reliability (Alderink et al., 2000).

6.4.4 Conclusion

The calculation of the knee and ankle joint centers with additional medial femoral condyle and medial malleolus markers are the primary aspects of the MA. The main aim of the MA, based on anatomical references and international recommendations for human movement analysis, was to minimize the effects of the experimental errors associated with anatomical landmarks identification.

The comparison of the results from the two models on the same gait cycles revealed good correlations for 93% of the gait variables. However, the use of the MA instead of the PiG model is recommended when analysing frontal and transverse plane gait data. This should lead to lower measurement errors for most of the gait variables and to a more accurate determination of the knee joint axis. Especially the documented lower errors in determining frontal plane knee joint moments are an important issue in regard to gait analysis data in patients with varus malalignment and medial compartment knee OA. It has been frequently shown that the adduction moment at the knee during gait is found to be the best predictor for the determination of the medial compartment loading of the knee (Andriacchi, 1994) and influences the outcome of proximal tibial osteotomy for knee OA (Wang et al., 1990). Using the standard PiG model may therefore lead to an erroneous clinical interpretation of gait parameters, especially in the frontal and transverse plane.

The experimental study in this section shows that the custom made model – referred to as MA – is well suited to determine three-dimensional joint angles and moments. It is therefore applied to analyse patients with varus malalignemt of the knee and healthy control subjects in the following study of Section 7.

7 Application of the Lower Body Model for Gait Analysis in Children and Adolescents With Varus Malalignment of the Knee

In this section, the custom made lower body model will be applied to patients with varus malalignment of the knee and healthy control subjects. The second experimental study in this thesis works on and answers the concrete aims 2 – 6 illustrated in Section 5.

7.1 Material and Methods

7.1.1 Subjects

Fourteen, otherwise healthy children and adolescent with varus malalignment of the knee 12-19 years of age were consecutively selected between January 2008 and March 2010 (Table 8). They had pathological varus alignment of the knee according to the mechanical axis angle based on a full length standing anteroposterior radiograph (Moreland et al., 1987). The clinical assessment has been carried out according to Figure A3 in the appendix Section. To ensure that differences in gait patterns in the patient group cannot be attributed to accompanying disease patterns, the following exclusion criteria were used:

- signs of OA or rheumatoid arthritis
- knee laxity
- anterior cruciate ligament deficiency
- neuromuscular dysfunction
- achondroplasia

- sagittal or transverse plane deformities of the leg
- flexion contractures in the knee or hip joint
- leg length discrepancy of more than 1 cm
- avascular necrosis
- history of major trauma or a sports injury of the knee
- knee surgery within the last 6 months
- chronic joint infection
- intraarticular corticosteroid injection
- morbid obesity according to the body mass index (Cole, Bellizzi, Flegal, & Dietz, 2000).

Fifteen healthy subjects between 14 and 21 years were recruited as control group (Table 8).

Table 8
Study Population Characteristics (*Mean With Standard Deviation in Parenthesis) and p-Values*

Variable	Controls (*n* = 15)	Patients (*n* = 14)	*p*-Value
Sex, no. female/no. male	9/6	9/5	.82
Age, years	15.1 (4.3)	15.1 (1.9)	.96
Height, m	1.64 (0.15)	1.68 (0.13)	.45
Weight, kg	52.8 (12.8)	55.4 (10.8)	.56
Body mass index, kg/m^2	19.2 (2.2)	19.5 (1.9)	.77
Mechanical axis angle,°		8.86 (7.38)	

Note. *p*-values (significance level $p < .05$) are based on parametric independent Student's *t*-test, except for the sex distribution, which is compared by non-parametric Mann-Whitney U test.

The anthropometric parameters did not significantly differ between the patient and control group (Table 8). None of the control subjects had previously been treated for any clinical lower back or lower extremity conditions and none had any activity-restricting medical or musculoskeletal conditions. In reference to the clinical assessment (Figure A3 in the appendix), these individuals had normal strength, full range of motion of the lower extremities and no knee instability or neurologic deficits. These controls did not undergo any radiographic examination. All subjects and/or their parents were thoroughly familiarized with the gait analysis protocol before giving informed consent to participate in this study.

7.1.2 Gait analysis and experimental design

Kinematic data were collected using an 8-camera Vicon motion capture system operating at a sampling rate of 200 Hz (VICON Motion Systems, Oxford, UK). Two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to collect kinetic data at 1000 Hz. The data from the eight cameras and the forces were recorded synchronically. The modified Helen Hayes marker set (Davis et al., 1991) referred to as MA and elaborately described in Section 6.2 was applied to determine joint centers. The joint moments were calculated using inverse dynamics approach. The mass of each segment was calculated as a percentage of body weight.

Data acquisitions were carried out in the presence of the same expert, who performed anthropometric measurements, landmark identification, marker placement as described (Figure 26) and a clinical examination (Figure A3 in the appendix). After the subjects completed a standardized questionnaire (Figure A1 and A2 in the appendix), they were asked to walk barefoot at their self-selected speed without targeting the force plates. Walking speed for each trial was calculated as the average velocity of the superior iliac spine marker in walking direction. Eight walking trials were recorded across a 15 m gait laboratory walkway (Figure 33) for each subject.

After each acquisition session, 3D marker trajectories were reconstructed and missing frames were handled with a fill-gap procedure. The data were smoothed with a Woltring filter and using a spline smoothing (Woltring, 1991). The average values from five trials were selected on the basis of good quality of the marker trajectories and ground reaction forces. To eliminate the confounding variable of bilateral involvement, all measurements were performed only on the limb with greater malalignment. Discrete variables of high clinical relevance describing peak values of kinematic (Table 9) and kinetic data (Figure 44, 47, 50) during stance phase were automatically determined by a custom made algorithm in Matlab 7.3.0 (The MathWorks, Inc., Natick, MA, USA). Moments were normalized to body weight and the convention adopted was that of an external moment, i.e. the resultant moment of the external forces.

7.1.3 Radiographic measurement

Radiographic assessment of the lower limb alignment was conducted on the same day as the gait analysis. Patients stood in a forward knee position with the patella centered over the femoral condyles and feet straight ahead to control for any foot rotation effects (Hunt, Fowler, Birmingham, Jenkyn, & Giffin, 2006) and to attain a true full-length weightbearing AP radiograph (Paley, 2002).

In accordance with Hunt et al., (2008), Miyazaki et al. (2002), Specogna et al. (2007) and Weidenhielm et al. (1994) using the same method for a similar purpose, the mechanical axis angle (MAA) of the lower extremity formed between a line drawn from the center of the hip to the center of the knee and a line drawn from the center of the knee to the center of the ankle (Figure 38) was used to quantify alignment in the frontal plane and obtained with the analysis software DiagnostiX-32 (Gemed) by the same investigator. The center of the hip was found as the geometric center of the femoral head, the center of the knee was identified as the midpoint of the tibial spines extrapolated inferiorly to the surface of the intercondylar eminence (Moreland et al., 1987), and the center of the ankle was defined as the mid-width of the tibia and fibula at the level of the tibial plafond (Paley, 2002).

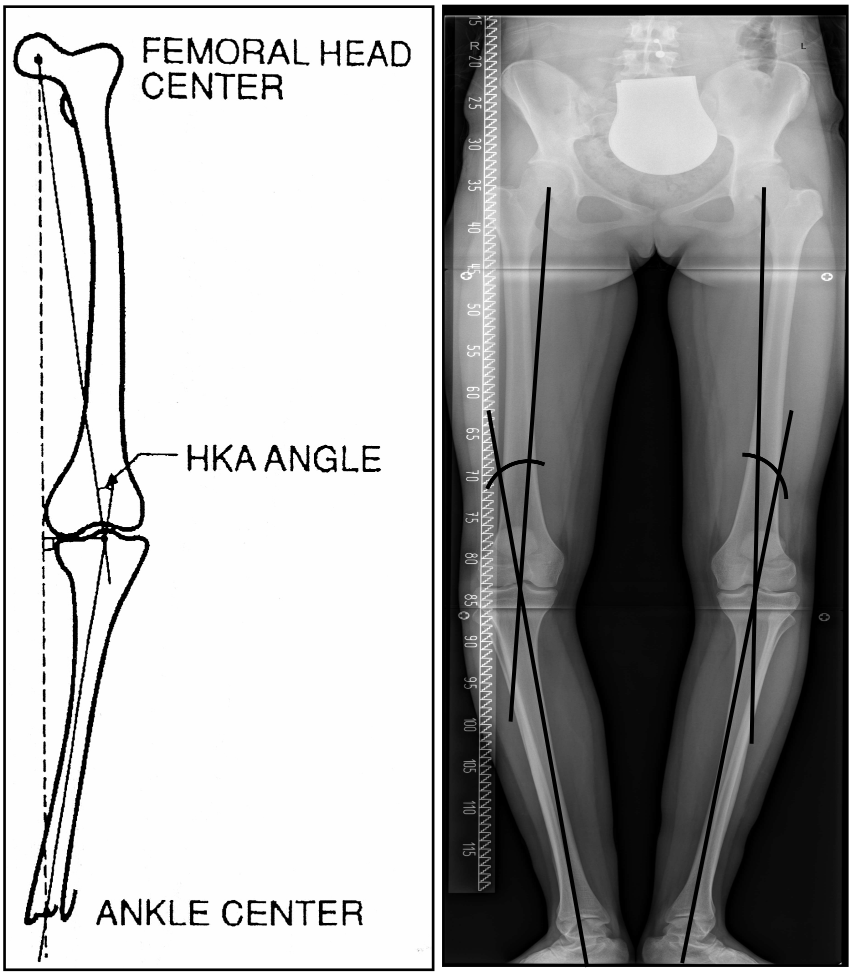


Figure 38. Hip-knee-ankle angle (HKA angle).

The MAA uses therefore the same landmarks and definitions like the three-dimensional gait analysis to measure lower limb alignment in the frontal plane to enable comparison between these two measurements. Alignment was defined as pathological varus when the angle was more than 1.3° (Moreland et al., 1987). In accordance with Specogna et al. (2004), reliability of MAA measurements was high ($ICC_{2,1}$ = 0.97). Table 8 in the Subjects Section shows the mean varus alignment of the present patient group.

7.1.4 Statistical analysis

The shape of distribution of the sample data collected was checked using the Kolmogorov-Smirnov test. Because a normal distribution was confirmed in the present study, differences between groups were tested for significance using a Student's *t*-test. The significance level adopted in this study was set at $p \leq .05$ and the statistical calculations were carried out using the SPSS version 12.0.1 (Chicago, IL, USA).

Simple linear regression (Pearson product-moment correlation coefficient; *r*) was used to examine the relationship between the MAA obtained from static radiographs and the static lower limb alignment measurements based on reflective markers as well as the maximum external knee adduction moment during stance phase.

7.2 Results

7.2.1 Relationship between static alignment obtained from radiographs and based on reflective markers

The scatter diagram (Figure 39) shows a strong linear relationship ($r = .93$) between the MAA measured from standing radiographs and the static lower limb alignment measurements in the frontal plane based on reflective markers and the gait analysis system. The correlation is significant at the .01 level.

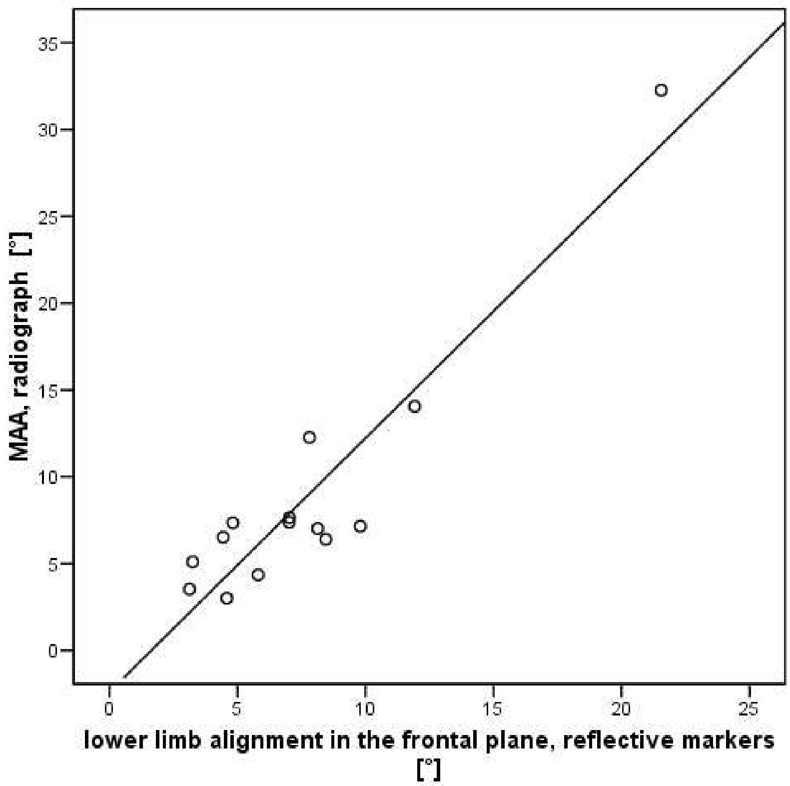


Figure 39. Scatter diagram of the relationship between the radiographic mechanical axis angle (MAA) and the static lower limb alignment measurements in the frontal plane based on reflective markers and the gait analysis system.

7.2.2 Relationship between static varus malalignment obtained from radiographs and dynamic knee adduction moment

The scatter diagram (Figure 40) shows a linear relationship (r = .79) between the MAA measured from standing radiographs and the maximum external knee adduction moment during the stance phase. The correlation is significant at the .01 level.

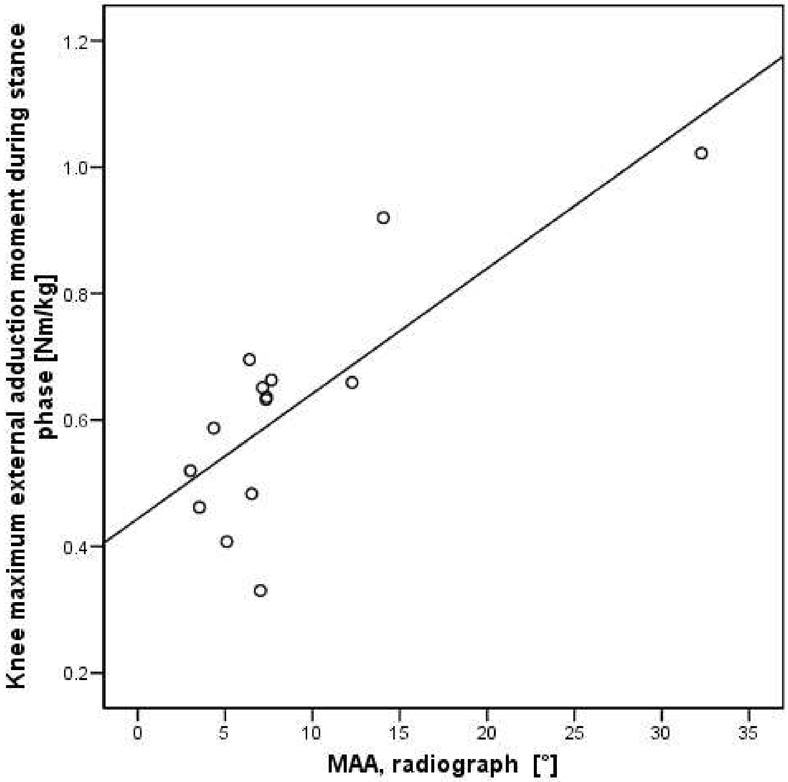


Figure 40. Scatter diagram of the relationship between the radiographic mechanical axis angle (MAA) and the maximum external knee adduction moment during the stance phase.

7.2.3 Kinematic differences

Average values and the significance level for the kinematic parameters of both groups are presented in Table 9.

Table 9
Mean, Standard Deviation and p-Values for Kinematic Parameters During Stance Phase

Plane	Joint	Kinematic parameter	Controls (*n* = 15) Mean	SD	Patients (*n* = 14) Mean	SD	*p*-Value
Sagittal	Foot	Sole angle initial contact	-16.13	3.10	-16.29	4.55	.91
	Knee	Max. flexion stance phase	14.55	4.65	17.23	5.12	.15
		Max. extension terminal stance	3.58	3.69	8.19	6.20	.02*
	Hip	Max. extension stance phase	-13.47	8.35	-9.97	10.17	.32
Frontal	Knee	Max. adduction stance phase	3.55	2.45	10.91	5.13	<.001***
		Range of motion adduction/abduction stance phase	6.44	2.43	8.97	5.61	.12
	Hip	Max. adduction stance phase	7.45	2.71	4.14	4.68	.03*
		Max. abduction stance phase	-5.48	2.09	-7.65	3.55	.05*
Transverse	Foot	Max. progression stance phase	-5.93	5.40	-4.26	6.68	.46
	Knee	Max. internal rotation stance phase	3.66	6.73	4.86	16.26	.79
	Hip	Max. internal rotation stance phase	14.14	7.67	15.69	9.74	.64

Note. Kinematic parameters in degrees. SD = standard deviation; flexion = positive, extension = negative; adduction = positive, abduction = negative; internal rotation = positive, external rotation = negative.
* $p \leq .05$. *** $p \leq .001$.

In the sagittal plane, the patient group walked with a significantly reduced maximum knee extension during terminal stance (Figure 41).

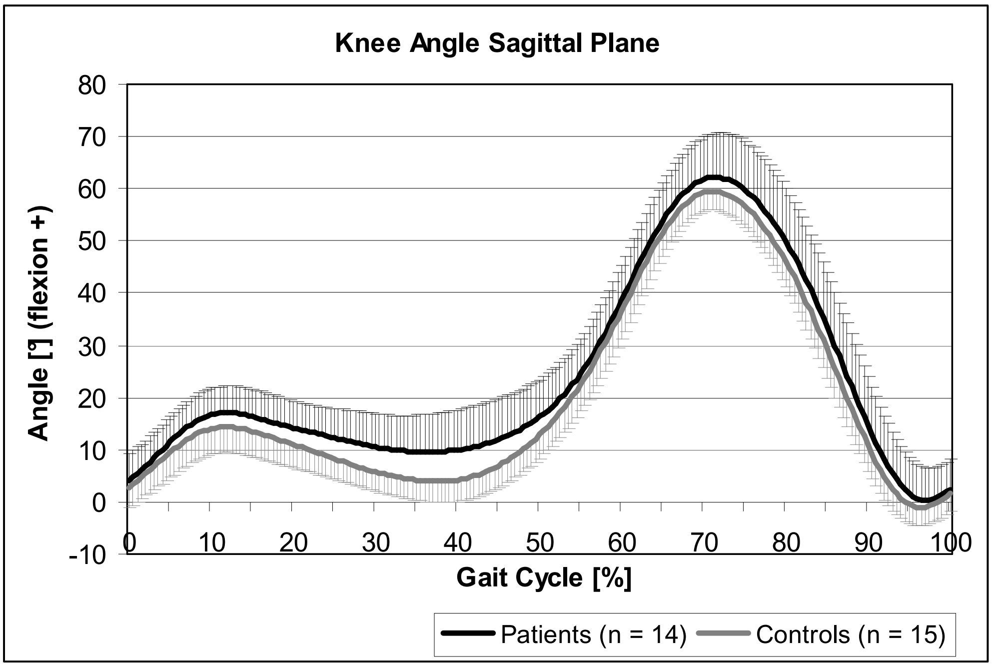


Figure 41. Average group curves of the knee sagittal plane angle. The error bars above and below the mean at each time point represent the standard deviation.

Furthermore, significant differences between patients with varus malalignment of the knee and healthy controls were detected for all peak knee and hip angles in the frontal plane (Figure 42, Figure 43). Kinematic parameters in the transverse plane were similar for both groups.

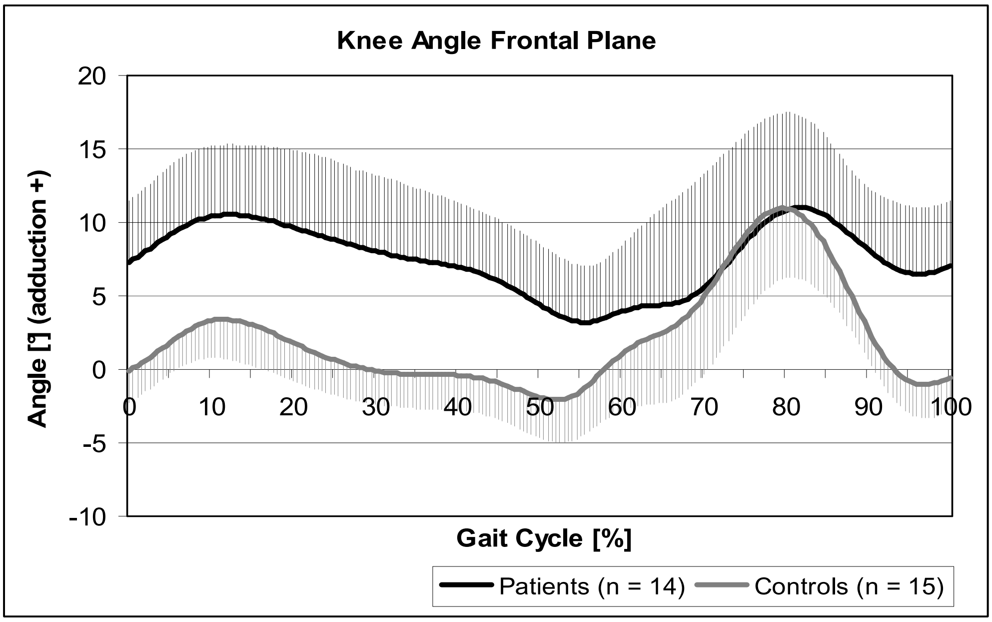


Figure 42. Average group curves of the knee frontal plane angle. The error bars above and below the mean at each time point represent the standard deviation.

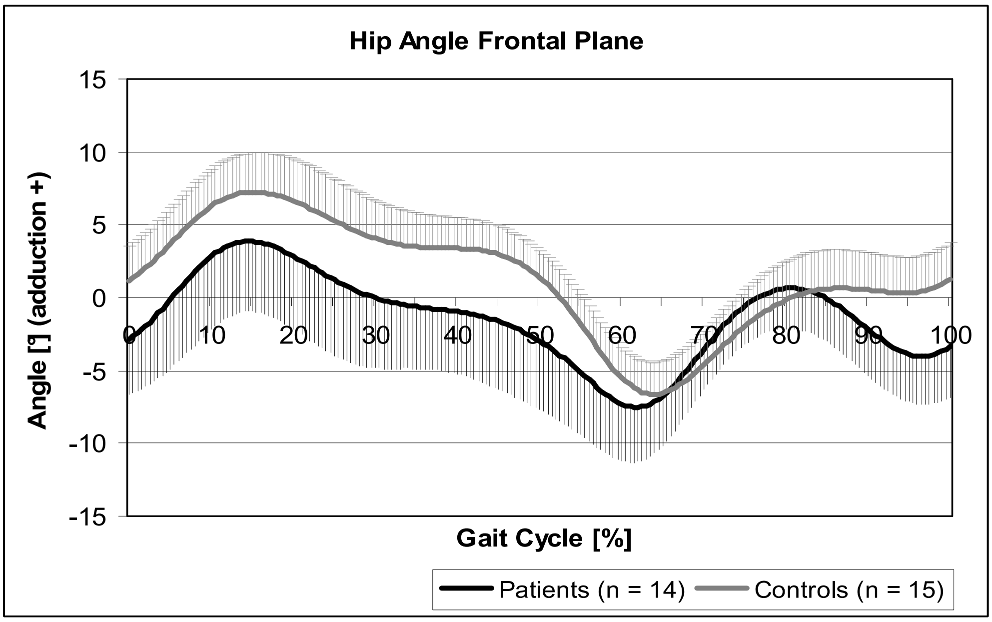


Figure 43. Average group curves of the hip frontal plane angle. The error bars above and below the mean at each time point represent the standard deviation.

7.2.4 Kinetic differences

7.2.4.1 Sagittal plane

Figure 44 shows the knee and hip joint moments in the sagittal plane for both groups.

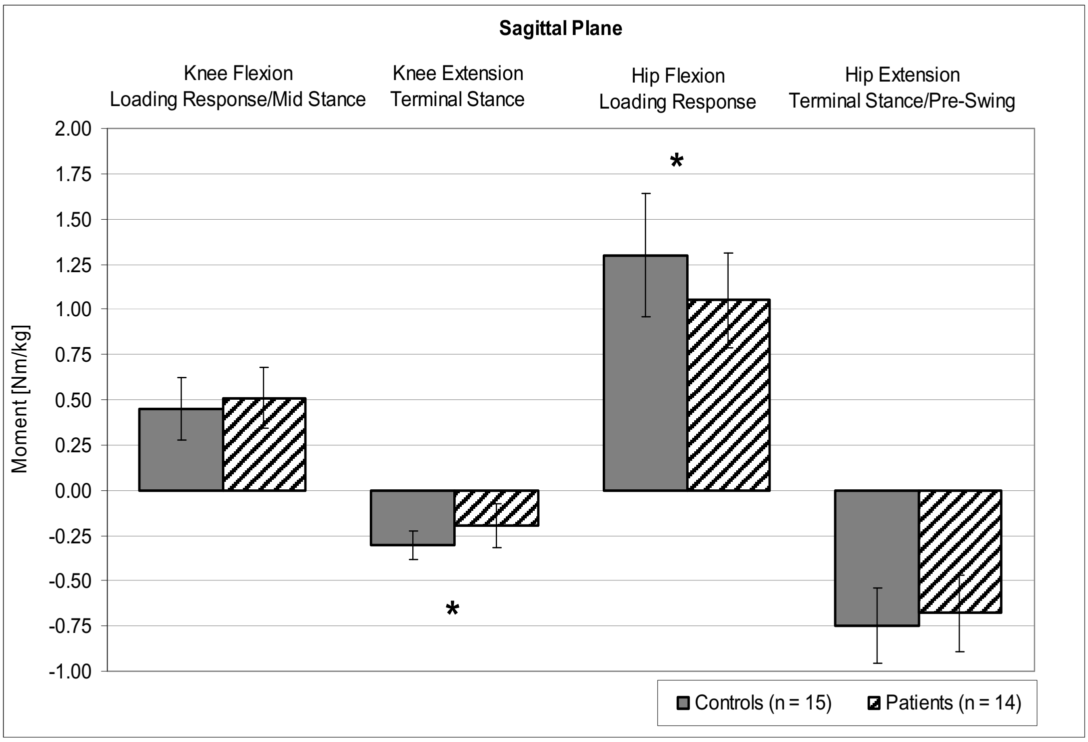


Figure 44. Average maximum external knee and hip joint moments during stance phase in the sagittal plane.
* $p \leq .05$.

Children and adolescents with varus malalignment of the knee exhibited a significantly lower maximum knee extension moment in terminal stance (Figure 45) and maximum hip flexion moment in loading response (Figure 46) compared to the control group.

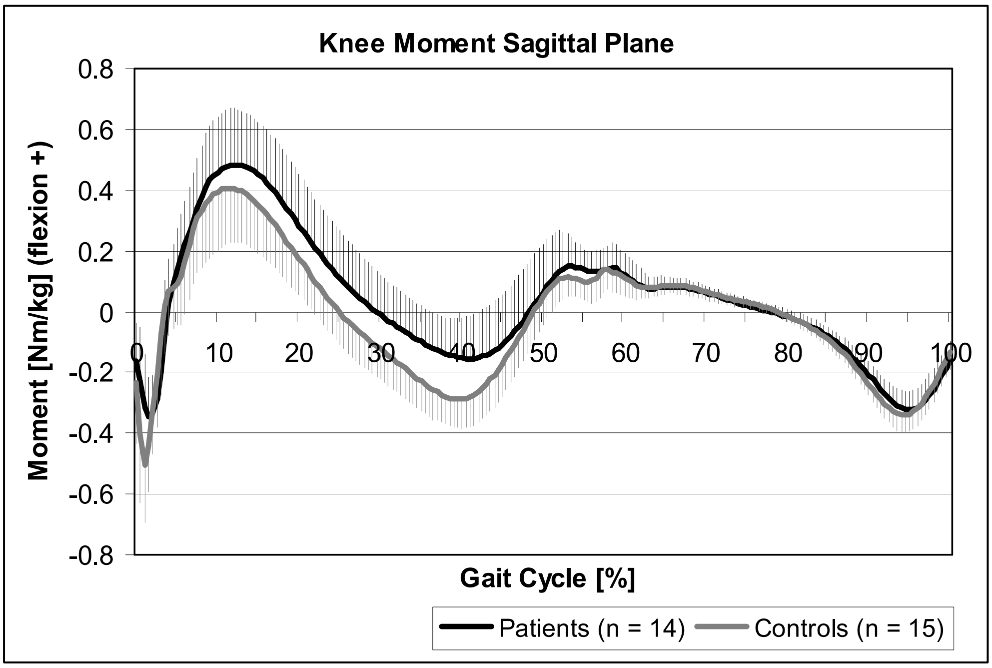


Figure 45. Average group curves of the knee sagittal plane moment. The error bars above and below the mean at each time point represent the standard deviation.

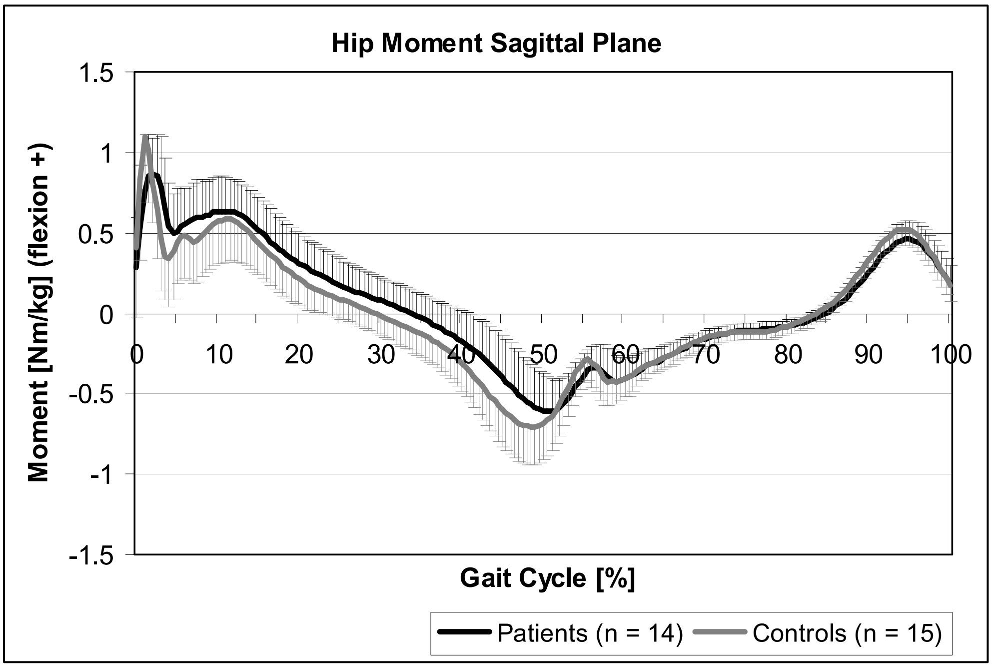


Figure 46. Average group curves of the hip sagittal plane moment. The error bars above and below the mean at each time point represent the standard deviation.

7.2.4.2 Frontal plane

As expected, the maximum knee adduction moments in mid and terminal stance in the frontal plane were significantly higher in the patient group (Figure 47, Figure 48).

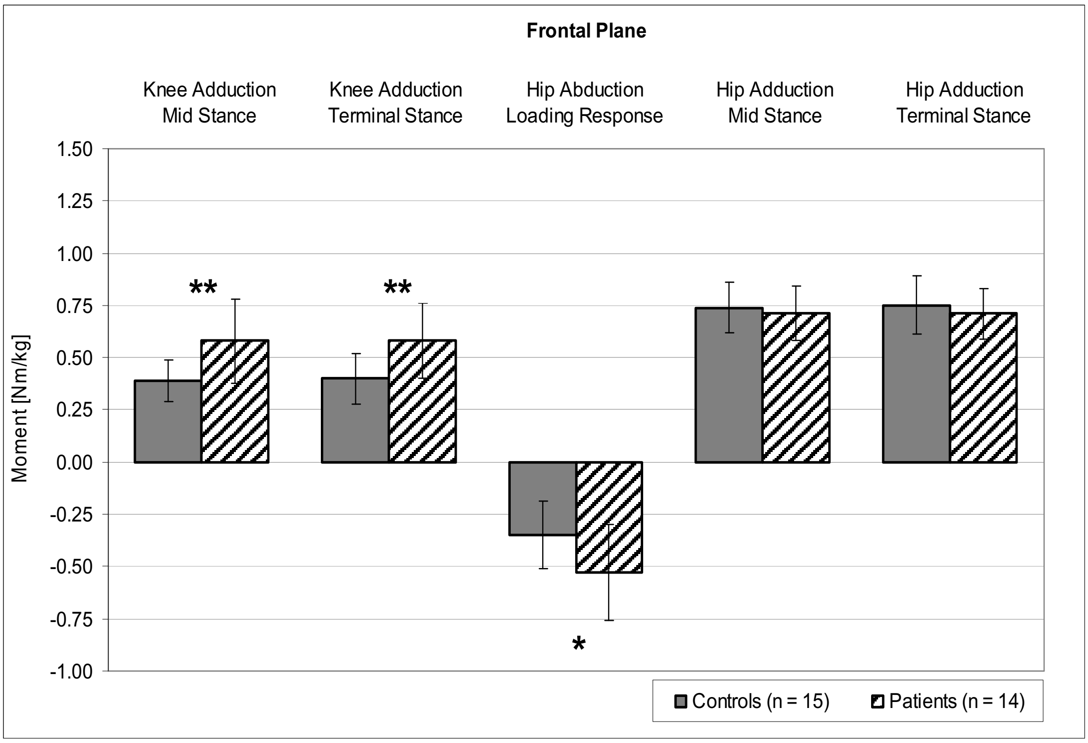


Figure 47. Average maximum external knee and hip joint moments during stance phase in the frontal plane.
* $p \leq .05$. ** $p \leq .01$.

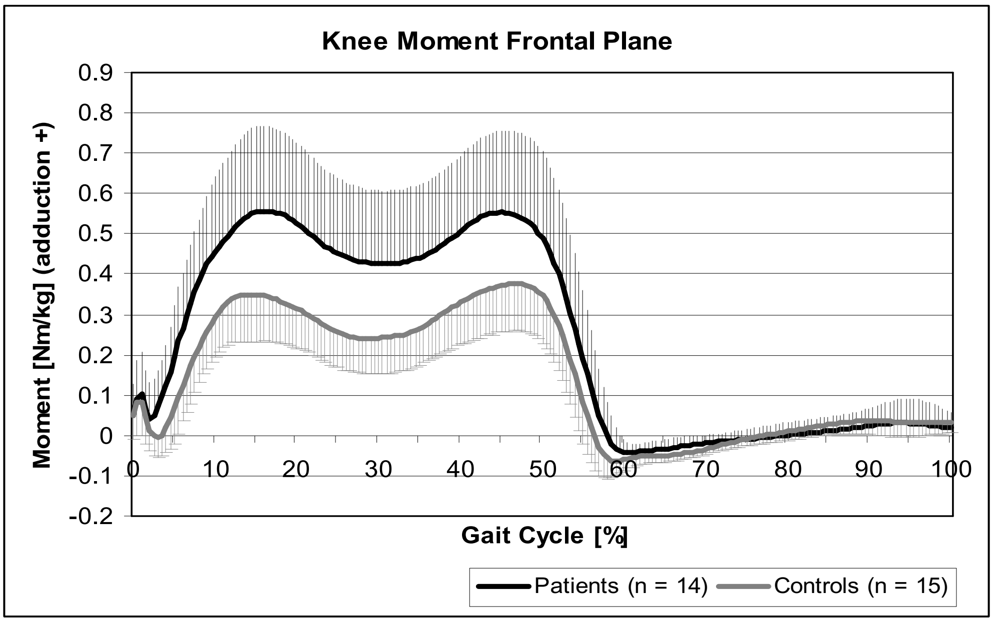


Figure 48. Average group curves of the knee frontal plane moment. The error bars above and below the mean at each time point represent the standard deviation.

Moreover, the maximum hip abduction moment in loading response was significantly increased in patients with varus malalignment of the knee. The maximum hip adduction moments in mid and terminal stance were not significantly different in both groups (Figure 47, Figure 49).

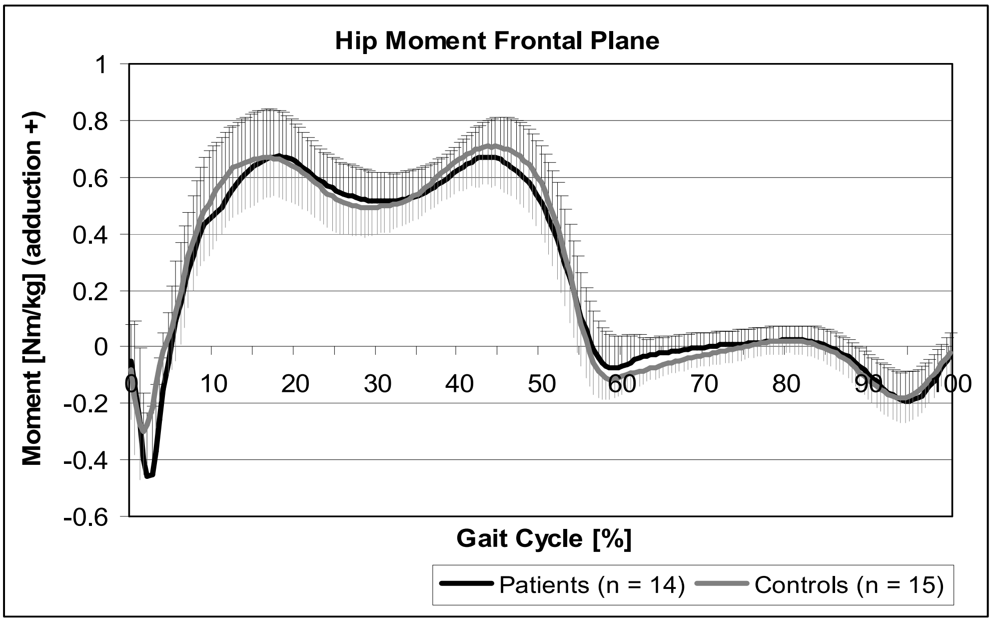


Figure 49. Average group curves of the hip frontal plane moment. The error bars above and below the mean at each time point represent the standard deviation.

7.2.4.3 Transverse plane

In the transverse plane, the maximum knee internal rotation moment in terminal stance (Figure 50, Figure 51) and the maximum hip external rotation moment in loading response/mid stance (Figure 50, Figure 52) were significantly higher in children and adolescents with varus malalignment of the knee.

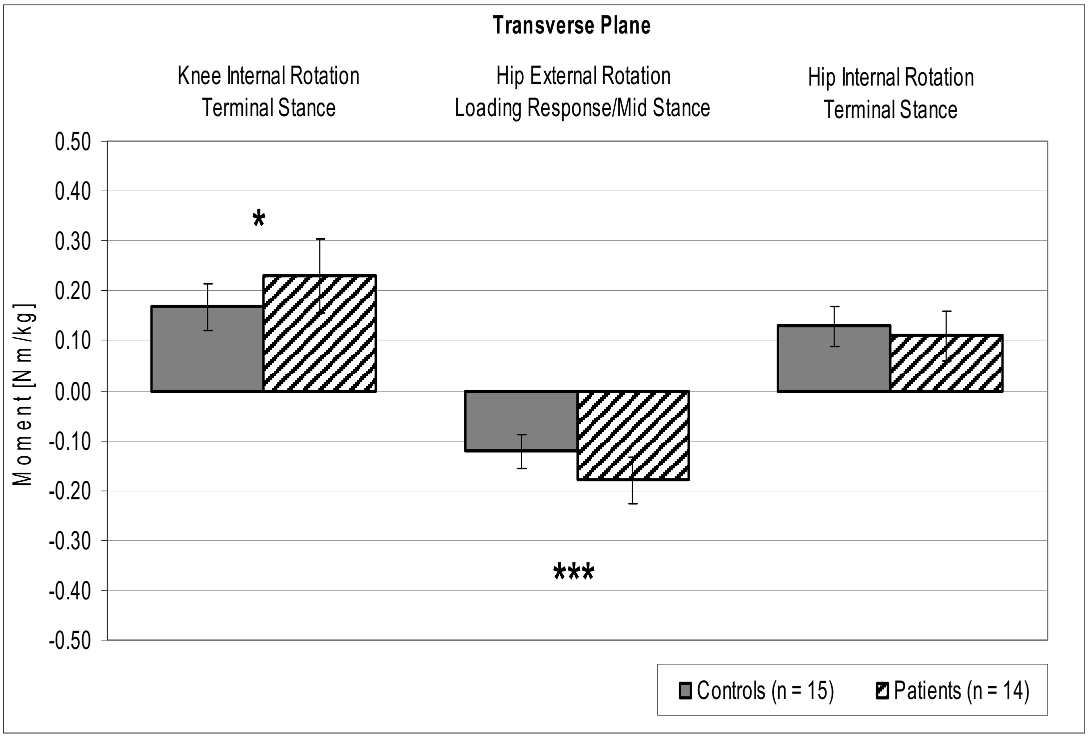


Figure 50. Average maximum external knee and hip joint moments during stance phase in the transverse plane.
* $p \leq .05$. *** $p \leq .001$.

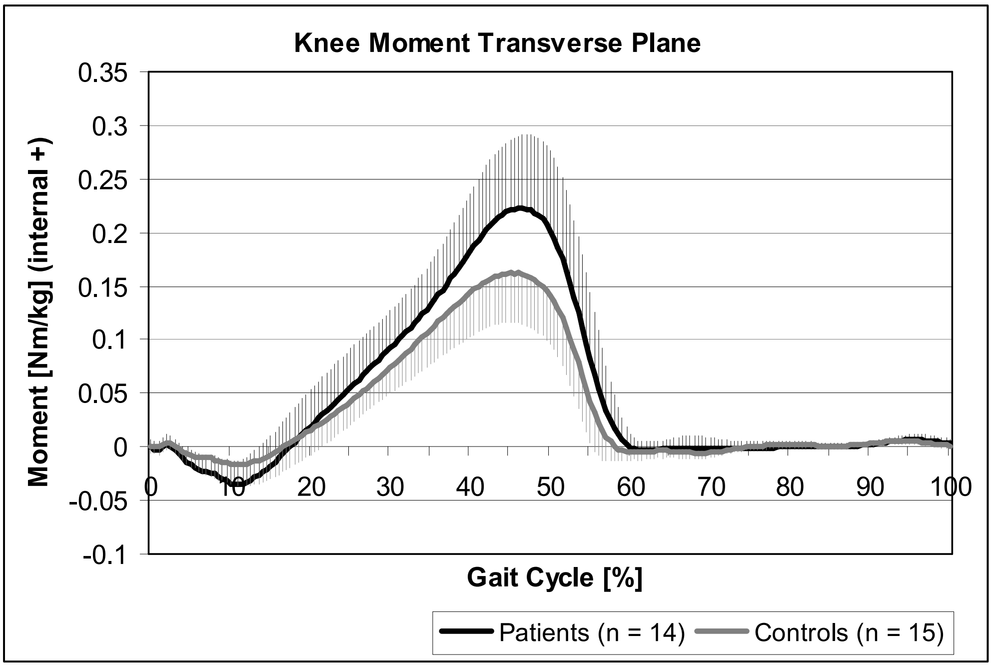


Figure 51. Average group curves of the knee transverse plane moment. The error bars above and below the mean at each time point represent the standard deviation.

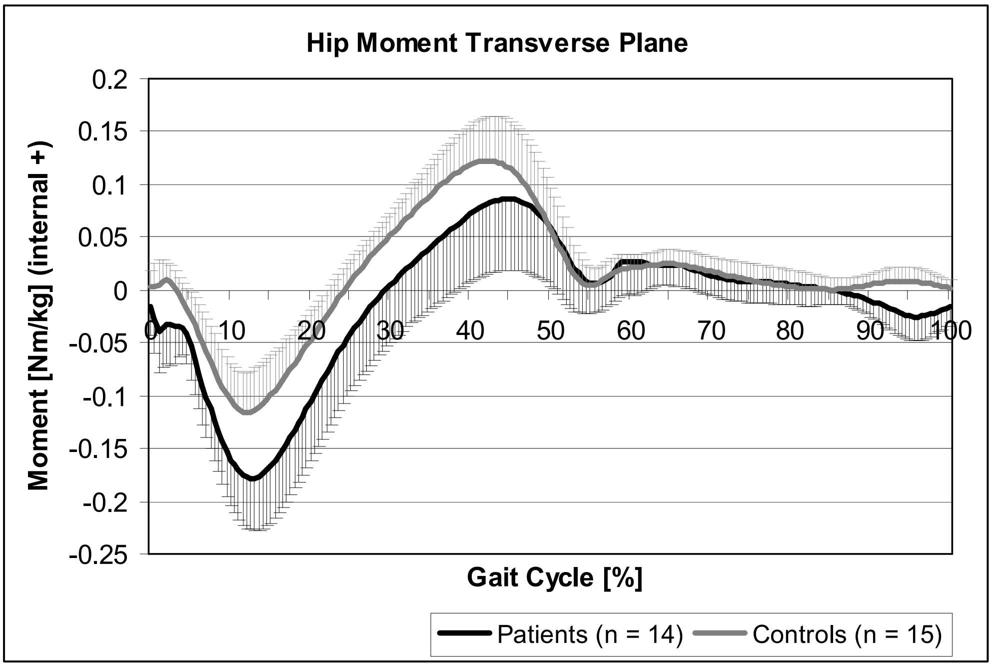


Figure 52. Average group curves of the hip transverse plane moment. The error bars above and below the mean at each time point represent the standard deviation.

7.2.5 Biomechanical compensatory mechanisms in patients with varus malalignment of the knee

The spatio-temporal gait parameters normalized according to Hof (1996) did not show significant differences between the two groups (Table 10).

Table 10
Mean, Standard Deviation and p-Values for Spatio-Temporal Gait Parameters

	Controls (n = 15)		Patients (n = 14)		
Spatio-temporal parameter	Mean	SD	Mean	SD	p-Value
Walking speed (m/s)	1.28	0.11	1.26	0.11	.60
Cadence $\left(\frac{cadence}{\sqrt{g\ /\ l_0}}\right)$	34.62	1.87	31.39	8.72	.17
Step length (step length / l_0)	0.77	0.06	0.78	0.08	.85
Stride length (stride length / l_0)	1.53	0.11	1.46	0.43	.55
Step width (m)	0.10	0.03	0.10	0.02	.77

Note. SD = standard deviation; l_0 = leg length; g = acceleration of gravity.

7.3 Discussion

Due to the present study is the first to actually report data pertaining to gait analysis in this young patient population without radiographic disease progression, the results will be discussed with healthy control subjects and gait analysis data in patients with established medial knee OA.

7.3.1 Relationship between static alignment obtained from radiographs and based on reflective markers

The strong linear relationship (r = .93) between the MAA measured from standing radiographs and the static lower limb alignment measurements in the frontal plane based on reflective markers and the gait analysis system suggests the suitability of the marker-based gait analysis system to determine frontal plane alignment of the leg. A Pearson product-moment correlation squared (r^2) of .86 indicates that 86% of the total variation in the MAA measured from standing radiographs can be explained by the linear relationship. Therefore, the static lower limb alignment measurements in the frontal plane based on reflective markers and the gait analysis system is a good non-invasive, alternative approach to analyse patients with varus malalignment of the knee.

7.3.2 Relationship between static varus malalignment obtained from radiographs and dynamic knee adduction moment

The linear relationship (r = .79) between the MAA measured from standing radiographs and the maximum knee external adduction moment during the stance phase shows that there is a positive correlation between these parameters. An r^2 of .62 indicates that 62% of the total variation in the maximum external knee adduction moment during the stance phase can be explained by the linear relationship. This suggests that the potentially influence of compensatory gait characteristics that may explain differences between static varus malalignment and dynamic knee adduction moment plays a secondary role in this patient group.

7.3.3 Kinematic and kinetic differences

7.3.3.1 Sagittal plane

The results of the present study generally implies that varus malalignment of the knee should not be viewed as an isolated problem in the frontal plane. It is largely responsible for kinematic and kinetic differences in the sagittal and transverse plane between the groups. In the sagittal plane, the reduction of the maximum knee extension moment in terminal stance can be explained by the significantly reduced maximum knee extension in terminal stance in the patient group. The maximum knee flexion in stance phase was not significantly different between groups. These results differ from previous studies reporting that patients with symptomatic medial knee OA stiffen their knees to reduce the demands on the quadriceps muscles and diminish pain (Al-Zahrani & Bakheit, 2002; Kaufman et al., 2001; Mündermann et al., 2005). This more extended gait pattern is not necessarily present in young patients with varus malalignment of the knee. A potential factor associated with the difference in knee flexion seen in the present study is the role of the quadriceps (Barios et al., 2009). Perhaps greater quadriceps strength or activation would allow for more knee flexion during weight acceptance. Barrios et al. (2009) actually reported that young individuals with asymptomatic varus knee alignment ambulated with greater knee flexion. Future studies might consider investigating the role of quadriceps muscle strength and activation in young patients with varus malalignment. Nevertheless, such differences in sagittal plane knee kinematics and kinetics could also be caused by slower walking speeds (Kirtley et al., 1985), as has been reported in several gait studies with patients with medial knee OA (Al-Zahrani & Bakheit, 2002; Gok et al., 2002; Kaufman et al., 2001).

At the hip, patients with medial knee OA walk with reduced peak extension angles and moments in late stance phase compared to healthy controls (Al-

Zahrani & Bakheit, 2002; Mündermann et al., 2005). Healthy older adults also demonstrate this pattern compared to younger individuals (Devita & Hortobagyi, 2000). The children and adolescents with varus malalignment of the knee in the present study demonstrated no significant differences in sagittal plane maximum hip angles and moments in late stance phase in comparison to the control group. This suggests that alterations in hip extension are secondary to either the aging process or the onset of OA.

7.3.3.2 Frontal plane

The maximum knee adduction moments in mid and terminal stance in the frontal plane were approximately 32% greater in the patient group. This is primarily caused by the lower extremity malalignment and the higher maximum knee adduction angle during the stance phase. This provides indirect evidence that knee adduction moments are significantly influenced by frontal plane lower extremity structure, regardless of the presence of medial knee OA. This also suggests that increased knee adduction moments equate to an increase in load and may have important implications for the progression of degenerative joint disease in this patient group (Miyazaki et al., 2002). Figure 53 shows the typical force vector characteristics for a patient with varus malalignment of the knee and a healthy subject. The calculation of the knee adduction moment is based on the cross product of the ground reaction force and its vertical distance to the knee joint center.

A

B

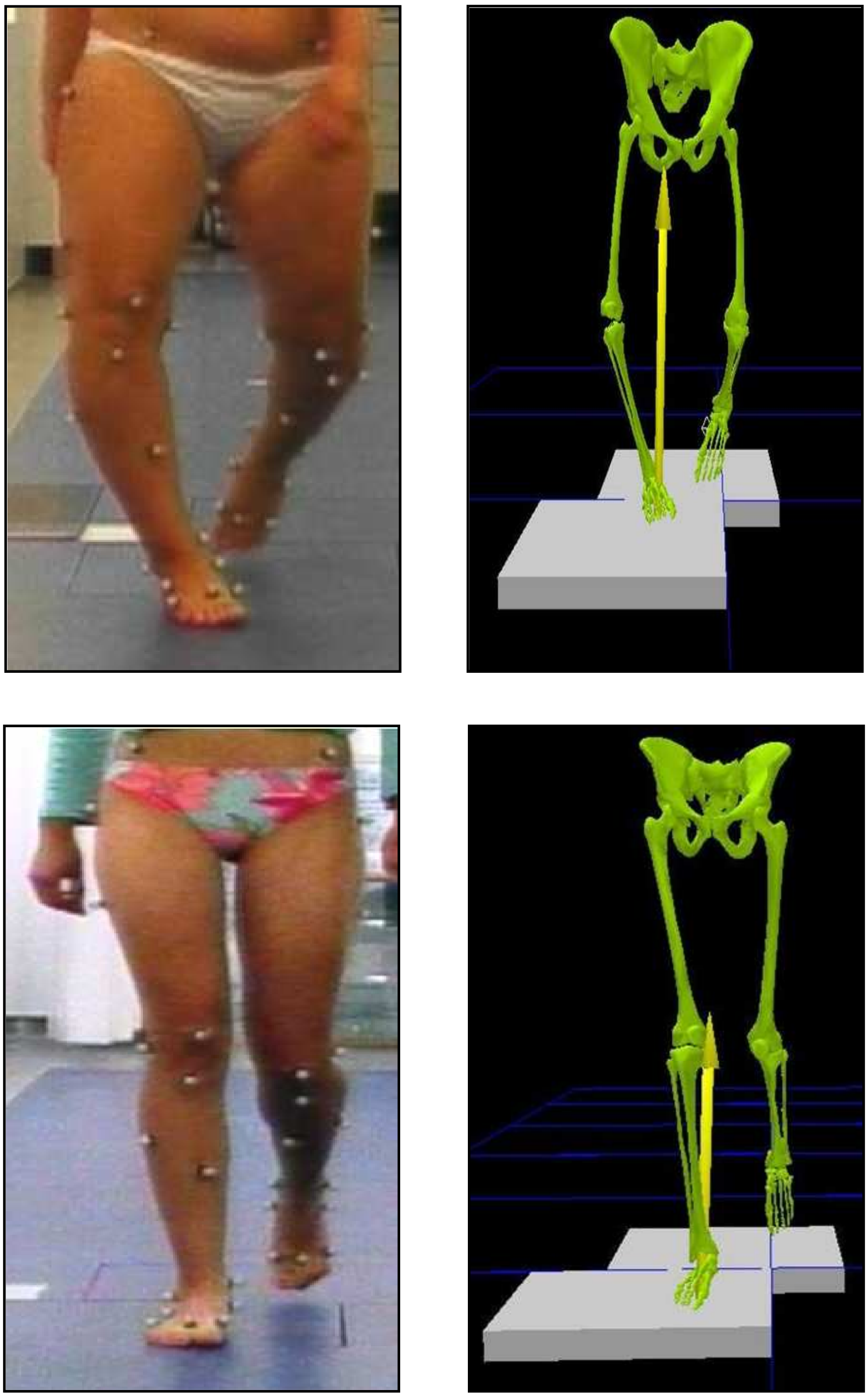


Figure 53. Typical force vector characteristics for a patient with varus malalignment of the knee (A) and a healthy subject (B) selected from each subgroup.

The higher maximum hip abduction moment immediately following heel strike in patients with varus malalignment of the knee indicates that these individuals exert greater hip adductor muscle forces during loading response in order to move their trunk laterally (Mündermann et al., 2005). It has been suggested that hip adductor muscles may have the capability to stabilize knees with medial compartment OA or knees with varus malalignment (Yamada, Koshino, Sakai, & Saito, 2001). This change in loading pattern is a potential mechanism of gait compensation used by patients with varus malalignment of the knee to reduce the mediolateral distance between the center of mass and the knee joint center. These results are in accordance with Mündermann et al. (2005) showing a higher maximum hip abduction moment in patients with medial knee OA. In accordance with Barrios et al. (2009), the patient group in the present study exhibited maximum hip adduction moments in mid and terminal stance that were similar to controls. The external hip adduction moment is balanced by the internal hip abductor muscle moment, suggesting sufficiently strong hip abductor muscles in this young patient group.

7.3.3.3 Transverse plane

Regarding the transverse plane, abnormally increased maximum knee internal rotation and hip external rotation moments were present in children and adolescents with varus malalignment of the knee. Due to the fact that kinematic parameters in the transverse plane were not significantly different between the two groups, the observed differences in joint moments are not directly related to transverse plane rotations. Astephen et al. (2008) also reported that patients with knee OA walk with a higher late stance phase knee internal rotation moment. While transverse plane mechanics have been implicated in the progression of knee OA (Andriacchi & Mündermann, 2006), only a few studies have quantified differences in transverse plane kinetics at the knee (Astephen et al., 2008; Gok et al., 2002; Landry et al., 2007) and none at the hip in patients with knee OA. It has been suggested that changes in transverse plane mechanics at the knee can initiate degenerative changes by placing new loads on regions of the articular cartilage that were previously conditioned for different load levels (Andriacchi & Mündermann, 2006). The author only can speculate that these higher moments in patients with varus malalignment of the knee later lead to pain, increased ligament forces or degenerative changes in the knee joint.

7.3.4 Biomechanical compensatory mechanisms in patients with varus malalignment of the knee

Different authors have reported that adult patients with established medial knee OA may be able to alter their walking mechanics and to reduce their maximum knee adduction moment (Andrews et al., 1996; Hurwitz et al., 2002; Mündermann et al., 2005). It is believed that patients with medial knee OA walk at slower speeds to reduce loading in the medial compartment of the knee (Robon, Perell, Fang, & Guererro, 2000). However, in contrast to previous studies in patients with knee OA (Al-Zahrani & Bakheit, 2002; Gok et al., 2002; Kaufman et al., 2001; Weidow et al., 2006), patients with varus malalignment of the knee and healthy controls in the present study walked at similar speeds. Thus, differences in gait patterns could not be attributed to differences in walking speed. Moreover, there were no significant differences in other spatio-temporal gait parameters between groups. Young patients with varus malalignment of the knee, with no signs of knee OA, probably do not need to reduce their walking speed or alter their spatio-temporal gait parameters in order to decrease knee joint loading. Specifically, the foot progression angle did not differ between the two groups indicating that patients with varus malalignment of the knee do not reduce the knee adduction moment by compensating with an increased foot progression angle.

7.4 Conclusion

Lower limb alignment has received considerable attention in the literature pertaining to knee OA (Section 2.2). Despite numerous authors highlighting limitations of relying solely on static measurements of lower limb alignment in the study and treatment of knee OA and the potential benefits of measures of dynamic lower limb alignment and joint moments, the present study is the first to actually report data pertaining to gait analysis in this young patient population without radiographic disease progression. The purpose was to compare selected gait mechanics between young individuals with varus knees and those with normal knee alignment. The mechanics of interest were those previously reported to be altered in people with medial knee OA (Section 4).

The results of this study show that the peak external knee adduction moment is higher than normal in subjects with radiographic varus malalignment. This implies that higher medial compartment knee joint loads are present in this population and that varus knee alignment is largely responsible for the altered frontal plane mechanics associated with medial knee OA. As knee adduction moments relate to disease progression in patients with medial knee OA, this variable might also relate to disease development (Miyazaki et al., 2002). Sagittal plane-oriented knee variables were similar between the patient and control group, with the exception of the reduced maximum knee extension moment and angle in terminal stance in the patient group. These findings suggest that the stiffer, more extended gait pattern associated with medial knee OA is not necessarily present in individuals with healthy, varus knees who may eventually develop disease. In contrast to patients with established knee OA, young patients with varus malalignment of the knee probably do not need to alter their spatio-temporal gait parameters in order to decrease the knee joint loading. Overall, these data suggest that individuals with healthy varus knees exhibit some, but not all, of the altered mechanics seen in patients with medial knee OA.

8 Clinical Significance

Full-length standing X-rays are the current gold standard for assessing frontal plane limb alignment and for surgical planning. However, this examination fails to capture any changes in joint loading that may occur when the limb is moving and weight-bearing. Due to the fact that the adduction moment is found to be a significant risk factor for knee OA (Section 4), the three-dimensional gait analysis could be used as a diagnostic tool to gain a better understanding of the relationship between loading and the onset or progression of articular cartilage degeneration in patients with varus malalignment of the knee and it provides a means of detecting, which patients develop compensatory mechanisms to reduce the dynamic loading on the medial knee compartment. In young subjects with varus malalignment of the knee without signs of knee OA, three-dimensional gait analysis helps to establish, before the onset of irreversible degenerative changes, which people are at a high risk for developing knee OA. Consequently, three-dimensional gait analysis could be used for clinical prognoses regarding the onset or progression of medial knee OA.

The results of this study indicate that children and adolescents with varus malalignment of the knee do not show the typical load reducing compensatory mechanisms, for example an increased foot progression angle (Wang et al., 1990) or reduced walking speed (Kaufman et al., 2001) demonstrated in adult patients with established medial knee OA. Current non-invasive treatments of pathological medial compartment knee loading and their potential for lowering the external knee adduction moment were shown in Section 2.3.2. Based on a regression equation of a previous investigation (Andrews et al., 1996) involving healthy subjects, a reduction in maximum knee adduction moment by 10% with a 10° greater foot progression angle (more toe-out foot placement) can be expected. This suggests that gait training and the correction of biomechanical variables in children and adolescents with varus malalignment of the knee may help to reduce their maximum knee adduction moments at this early stage. Some patients may be able to alter their walking mechanics and this could be a potential strategy to delay the evolution of later medial knee OA (Miyazaki et al., 2002). However, all non-invasive load-modifying techniques reduce the maximum knee adduction moment only by approximately 10%. The maximum knee adduction moments in the present study were approximately 32% greater in children and adolescents with varus malalignment of the knee compared to healthy subjects. This suggests that non-invasive treatments alone can not change pathological medial compartment knee loading to normal.

Even if a surgical intervention is the essential treatment for varus knees with or without OA, dynamic gait parameters could be assistant for a postoperative prognosis. Prodromos et al. (1985) reported that the preoperative adduction moment could predict surgical outcome for knee OA with varus deformity. When the adduction moment was higher preoperatively, it was still increased postoperatively and the leg significantly changed to varus alignment again while lower adduction moment did not. Therefore, the preoperative adduction moment could be also used as a decision-making aid for the dimension of surgical correction of the varus-aligned knee.

9 Case Study

In contrast to the results of the experimental study in Section 7, the following case study indicates a patient with static varus malalignment of the knee (Figure 54) showing compensatory mechanisms to reduce the dynamic loading on the medial knee compartment.

Figure 54. Patient with varus malalignment of the knee (right > left).

The static radiographic assessment of the lower limb alignment based on a full length standing AP radiograph is visualized in Figure 55.

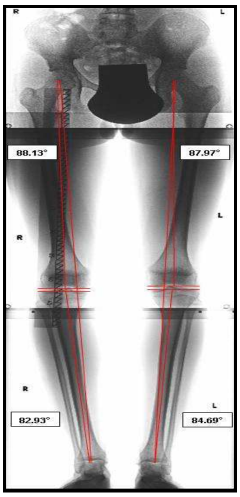


Figure 55. Radiographic assessment of the patient.

The quantification of the alignment in the frontal plane demonstrates the following results:

- no relevant malalignment on the thigh
- varus malalignment on the shank: ca. 4° (right), ca. 2.5° (left)
- MAA: 5.3° (right), 3.9° (left)

In contrast to the static measurement, the force vector characteristics during gait (Figure 56) showed a physiological distribution for the right leg. The force vector for the left leg passed slightly medial the knee joint center.

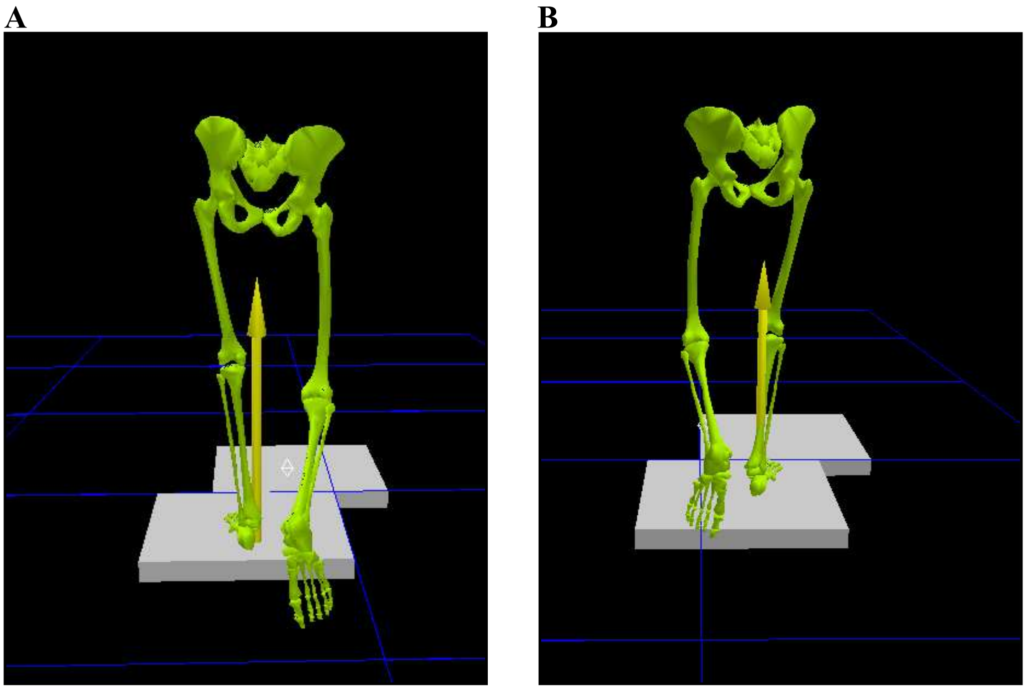


Figure 56. Force vector characteristics during gait for the left (A) and right leg (B).

The corresponding knee adduction moment in the frontal plane (Figure 57) was within a normal range for the right leg and abnormally increased in mid stance for the left leg.

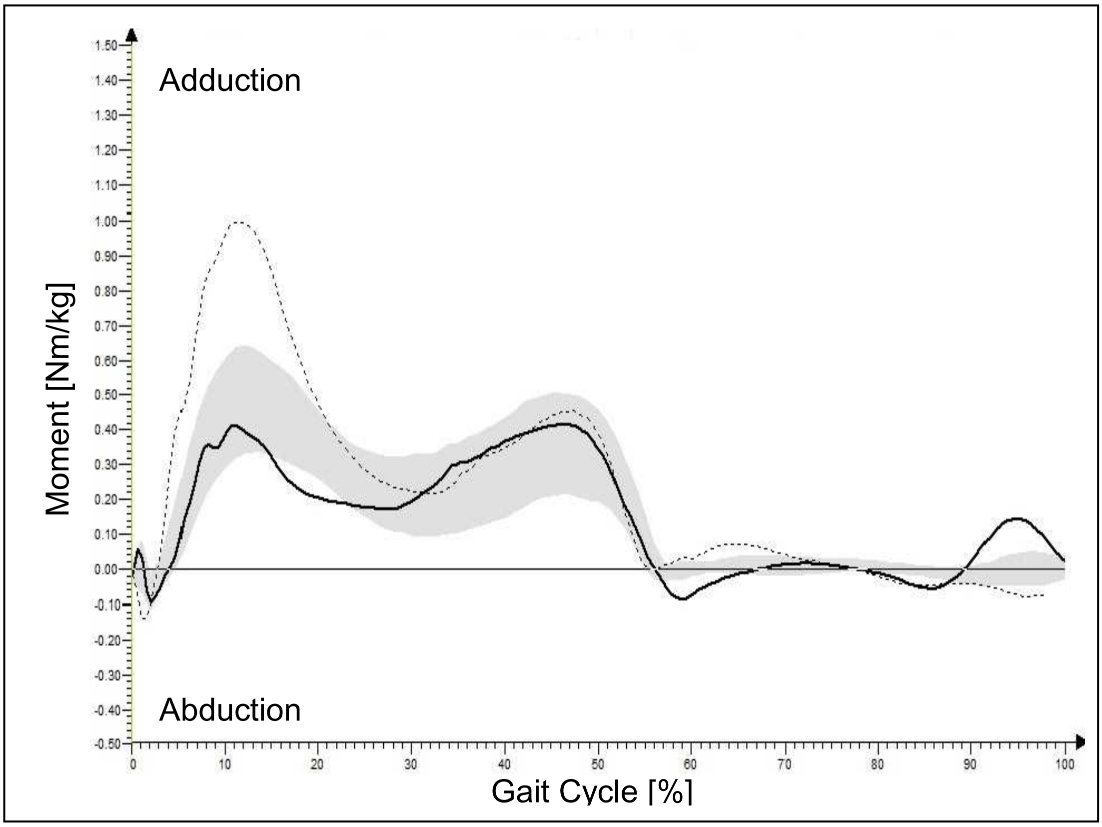


Figure 57. Curves of the knee adduction moment in the frontal plane. The gray graph indicates the mean with standard deviation for the healthy subject group (n = 15). The continuous graph indicates the right knee. The dotted graph indicates the left knee.

These dynamic joint loading characteristics are quite surprisingly in regard to the static radiographic assessment showing a more varus-aligned knee on the right side in this patient with varus malalignment of the knee. Individual compensatory mechanisms, which were previously proved to reduce the knee adduction moment during gait (Andrews et al., 1996; Hurwitz et al., 2002; Mündermann et al., 2005) can be used to explain this discrepancy between static measurement and dynamic loading situation. Figure 58 shows an increased foot progression angle (more toe-out foot placement or external rotation) for this patient on the right side.

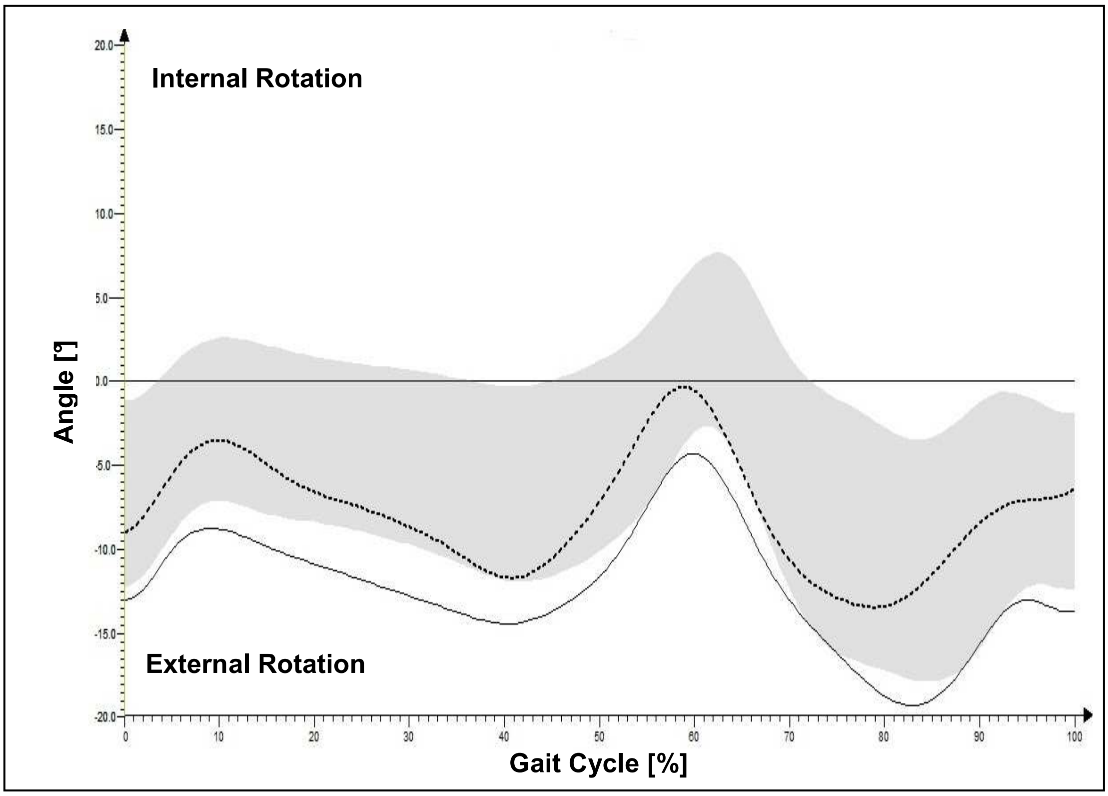


Figure 58. Foot progression angle in the transverse plane. The gray graph indicates the mean with standard deviation for the healthy subject group (n = 15). The continuous graph indicates the right foot. The dotted graph indicates the left foot.

This compensating strategy was elaborately described in Section 4.3 and can be considered as a mechanism used to lower the adduction moment at the knee (Wang et al., 1990).

This case report impressively shows the discrepancy between static measurement and dynamic loading situation. Since these altered kinematic data are not captured in a static radiograph, the three-dimensional gait analysis is a powerful tool to detect compensatory mechanisms and abnormal gait patterns that may influence the outcomes of degenerative joint diseases such as OA. In addition, this dynamic measuring technique could be used as a decision-making aid for surgical interventions in borderline values and potential treatments can be tailored to each patient.

10 Limitations

There are some limitations of the present thesis. The experimental study in Section 6 has been shown that the custom made marker-based model indicates good reliability and accuracy and is therefore well suited to determine three-dimensional joint angles and moments in patients with varus malalignemt of the knee. However, the accuracy of marker-based measures of lower limb alignment in general is ultimately dictated by the ability to correctly identify the position of the hip joint center of rotation. Although the position of the hip joint center in the present study was based on well-documented equations (Davis et al., 1991), it fails to account for individual anatomical differences. Maybe a functional approach to determine the hip joint center would have been more accurate. Another factor that must be accounted for when using surface markers is the precision of placement of the markers on the skin as well as the relative movement of the markers during movement. Although current motion capture systems possess extremely good accuracy and every attempt is made to be as precise as possible during marker placement and to minimize marker movement, this remains an aspect of quantitative gait analysis that requires consideration when viewing any results from gait analyses.

Further limitations of the present thesis include the absence of dynamic surface electromyography on certain muscle groups. A potential factor associated with the difference in knee flexion is the role of the quadriceps. Due to not measuring the activity of the quadriceps, the hypothesis from Barrios et al. (2009) that greater quadriceps strength or activation would allow for more knee flexion during weight acceptance in young individuals with asymptomatic varus knee alignment compared to patients with established knee OA could not be tested. Whether quadriceps strength is an effective treatment for maintaining knee joint stability and reducing knee joint loads (Mikesky et al., 2000) in patients with varus malalignment of the knee without signs of knee OA could also not be answered in the present examination. Similarly, the reason for the higher maximum hip abduction moment immediately following heel strike in patients with varus malalignment of the knee (Figure 47) remains unclear. It has been suggested that the activation of hip adductor muscles during loading response may have the capability to stabilize knees with medial compartment OA or knees with varus malalignment and to reduce the mediolateral distance between the center of mass and the knee joint center (Yamada et al., 2001).

Lastly, although the use of quantitative gait analysis for a variety of patient populations has increased in recent years, it remains a method of measurement, which is not available to all clinicians.

11 Outlook

The author believe that by identifying measures of gait pattern and dynamic loading situation in young patients with varus malalignment of the knee, potential interventions can be tailored to each patient. For example, some patients may be able to alter their walking mechanics and this could be a potential strategy to delay the evolution of later medial knee OA (Miyazaki et al., 2002).

This thesis indicates that several biomechanical factors, such as lower limb alignment and the knee adduction moment, contribute toward the pathogenesis of knee OA. To understand the complexity of knee OA and to develop earlier and more appropriate treatment strategies for the disease will require continued study on the interrelationships between risk factors for the disease and appropriate models of disease progression. A longitudinal study of individuals with healthy, varus knees would help to establish aberrant gait mechanics as risk factors for disease development and have the power to truly elucidate the role of dynamic loading in the progression of knee OA.

Future studies might also consider investigating the role of muscle strength and activation in patients with varus malalignment or knee OA. Continued research is needed to identify whether increased co-contraction leads to higher muscle stiffness or more joint loading, and to understand the long-term effect of altered muscle activity on the progression of OA.

As shown in Section 2.3.1, HTVO is an effective treatment for medial compartment knee OA and varus malalignment. High eccentric load concentration of the medial compartment can be reduced by lateral shift of the axial load. With this intervention, a high adduction moment in the frontal plane can be reduced to normal. Future studies should also investigate the late effects of HTVO on joint angles and moments in the sagittal and transverse plane during gait and in activities of daily life in young patients with varus malalignment of the knee.

12 Summary

Knee OA is the most common and disabling medical condition of the elderly (Felson et al., 1987). Of the joint compartments, the medial tibiofemoral compartment is most commonly affected (Dearborn et al., 1996). Many studies have suggested that varus alignment is associated with increased medial knee loading, which, in turn, contribute to the development of OA (McNicholas et al., 2000; Miyazaki et al., 2002; Moreland et al., 1987; Sharma et al., 1998). The use of quantitative gait analysis in addition to static radiographic measures of alignment has been suggested as a means in the study and treatment of knee OA (Hunt et al., 2008; Mündermann et al., 2008). Therefore, the present thesis aimed at determining the clinical relevance of three-dimensional gait analysis for the treatment of children and adolescents with varus malalignment of the knee as an adjunct to static radiographic measures.

The calculation of lower limb kinematics and joint kinetics during gait already requires accurate identifications of the joint centers. Thus, the present thesis consists of two experimental studies. First, it is shown that the custom made lower body model for clinical gait analysis is well suited to determine three-dimensional joint angles and moments (Section 6). In this experimental study, the reliability and accuracy of this model compared to the standard PiG model were estimated. Twenty-five subjects were gait analyzed to test the inter-trial reliability. Inter-session reliability was examined in ten healthy subjects. Moreover, the knee flexion/extension and varus/valgus kinematic profiles referred to the knee joint angle cross-talk were computed to test the accuracy of the models. Regarding frontal plane knee angles and moments as well as transverse plane motions in the knee and hip joint, the inter-session errors were lower for the custom made model compared to the standard clinical approach. Additionally, the new model produced a more accurate and reliable knee joint axis compared to the PiG model. These results are especially important for measuring frontal and transverse plane gait parameters. Therefore, the new model – evaluated in this study – seems to be suitable to analyse patients with frontal plane leg malalignment. Similarly, the strong linear relationship (r = .93) between the measured mechanical axis angle from standing radiographs and the static lower limb alignment measurements in the frontal plane based on reflective markers suggests the suitability of the marker-based gait analysis system to determine frontal plane alignment of the leg. Consequently, the static lower limb alignment measurements in the frontal plane based on the gait analysis system is a good non-invasive, alternative approach to analyse patients with varus malalignment of the knee.

In the second experimental study of this thesis, the tested lower body model was applied to patients with varus malalignment of the knee and healthy control subjects (Section 7). To the author's knowledge, this is the first study, which documents the biomechanical effect of varus malalignment during gait in a young patient population without radiographic disease progression. In this study, knee and hip joint angles and moments were obtained during walking. Moreover, the author wanted to know if typical compensatory mechanisms – shown in patients with established knee OA – are also present in this young patient group. Fourteen, otherwise healthy children and adolescents with varus malalignment of the knee and 15 healthy control subjects were analyzed. In summary, differences were observed between normally aligned and varus-aligned knees in the frontal plane. The external knee adduction moment was substantially higher in patients with varus knees and similar to previously reported values for individuals with medial knee OA. This provides indirect evidence that knee adduction moments are significantly influenced by frontal plane lower extremity alignment, regardless of the presence of medial knee OA. These high values support the notion that this moment equates to an increase in load and may have important implications for the progression of degenerative joint disease in this patient group (Miyazaki et al., 2002). Interestingly, the varus group ambulated not with a stiffer knee. A somewhat surprising finding as patients with established OA typically ambulate with more knee extension. In the transverse plane, abnormally increased knee internal rotation and hip external rotation moments were present in children and adolescents with varus malalignment of the knee. These findings imply that varus malalignment of the knee should not be viewed as an isolated problem in the frontal plane. In contrast to adult patients with established medial knee OA, the young patients assessed in the present study did not show typical compensatory mechanisms such as an increased foot progression angle or reduced walking speed. This suggests that targeted gait training and the correction of biomechanical variables in young patients with varus malalignment of the knee may reduce their knee adduction moment. This could be a potential strategy to delay the progression of articular cartilage degeneration in the knee joint. Overall, these data suggest that individuals with healthy varus knees exhibit some, but not all, of the altered mechanics seen in patients with medial knee OA. Clinical gait analysis is therefore a powerful tool to detect compensatory mechanisms and to assess abnormal gait patterns that may influence the outcomes of degenerative joint diseases such as OA.

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List of Abbreviations

AP	anteroposterior
FT	femoral torsion
g	acceleration of gravity (= 9.81 m/s² on earth)
HKA	hip-knee-ankle
HTO	high tibial osteotomy
HTVO	high tibial valgus osteotomy
Hz	hertz
ICC	Intraclass Correlation Coefficient
i.e.	that is
KAD	Knee Alignment Device
KCD	Knee Center Device
$\boldsymbol{l}_0$	leg length
MA	custom made model
MAA	mechanical axis angle
Nm	Newton meter
OA	osteoarthritis

PiG	Plug-in-Gait
r	Pearson product-moment correlation coefficient
r^2	Pearson product-moment correlation squared
RMSE	root mean square error
RMSE%	relative root mean square error
ROM	range of motion
SD	standard deviation
TEM	technical error of measurement
TEM%	relative technical error of measurement
TFA	tibiofemoral alignment

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Appendix

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- Clinical Assessment

Orthopedic Questionnaire

Persönliche Daten

Name: ______________________________ Vorname: ______________________________

Straße: ______________________________ PLZ / Ort: ______________________________

Telefon (privat): ________________________ E-Mail: ______________________________

Telefon (mobil): ________________________

Geschlecht: ❑ w ❑ m Geburtsdatum: ________________ Alter: __________ Jahre

Ärztliche Diagnose:

____

____

____

Aktuelle Beschwerden/Schmerzen und funktionelle Einschränkungen in den unteren Extremitäten: ❑ nein ❑ ja

Wenn ja, wo und welche?:

____

____

____

Sport möglich? ❑ nein ❑ ja

Wenn ja, was für Sportarten?:

1. ____________________ seit __Jahren __mal pro Woche/Monat Dauer pro Einheit ca. __min.

2. ____________________ seit __Jahren __mal pro Woche/Monat Dauer pro Einheit ca. __min.

3. ____________________ seit __Jahren __mal pro Woche/Monat Dauer pro Einheit ca. __min.

Welche der angegebenen Sportarten werden im Verein ausgeübt? ❑ 1 ❑ 2 ❑ 3

Welches ist das bevorzugtes Bein (Sprungbein, Spielbein, Standbein)? ❑ re ❑ li

Figure A1. Orthopedic questionnaire – Page 1.

Wie absolvieren Sie überwiegend Ihre Alltagsaktivitäten (Schule, Ausbildung, Beruf, Freizeit)?

1. ❑ sitzend ❑ stehend ❑ gehend
2. ❑ wenig Bewegung ❑ viel Bewegung
3. ❑ keine körperliche Arbeit ❑ geringe körpeliche Arbeit ❑ schwere körpeliche Arbeit

Gegenwärtige Therapie: ❑ nein ❑ ja

Medikation:
____

Physiotherapie (Behandlungsschwerpunkte, Häufigkeit):
____

____

____

Werden selbständig noch weitere Übungen durchgeführt? ❑ nein ❑ ja

Wenn ja, was und wie häufig?
____

____

Bisherige Operationen an den unteren Extremitäten: ❑ nein ❑ ja

Wenn ja, bitte Art und Zeitpunkt der Operation angeben:
1. ____
2. ____
3. ____

Tragen Sie orthopädische Einlagen: ❑ nein ❑ ja

Wenn ja, Art der Einlage (z. B. gegen Spreizfuß, Senkfuß):

1. ______________________ seit: ______________
2. ______________________ seit: ______________

Figure A2. Orthopedic questionnaire – Page 2.

Clinical Assessment

Klinische Untersuchung

Untersucher(in)		ID	
Datum		Name	

	Passives Bewegungsausmaß	links			rechts			Muskelfunktionstest	links	rechts
Hüfte	Extension/Flexion							Flexion (Sitz)		
								Extension (Bauchlage)		
	Rectus femoris							Gluteus (Bauchlage)		
	Antetorsion							Abduktion (0°)		
	Abduktion/Adduktion (0°)							Abduktion (90°, Sitz)		
	Abduktion (90°)							Adduktion (0°)		
	Innenrotation/Außenrotation (0°, Bauchlage)							Adduktion (90°, Sitz)		
Rumpf								Rectus abdominis (beidseits)		
Knie	Extension/Flexion							Flexion (Sitz)		
	Ischiokrurale							Extension (Sitz)		
Sprunggelenk	Dorsalflexion/Plantarflexion (Knie 0°)							Dorsalflexion (Sitz)		
	Dorsalflexion (Knie 90°)							Plantarflexion (Stand, Wdh)		
	Tibiatorsion									

Figure A3. Clinical assessment.