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Effect of lower limb malalignment in the frontal plane on transverse plane mechanics during gait in young individuals with varus knee alignment

Felix Stief [a,*], Harald Böhm [b], Chakravarthy U. Dussa [b], Christel Multerer [b], Ansgar Schwirtz [c], Andreas B. Imhoff [d], Leonhard Döderlein [b]

[a] Orthopedic University Hospital Friedrichsheim gGmbH, Frankfurt/Main, Germany
[b] Orthopedic Hospital for Children, Aschau i. Chiemgau, Germany
[c] Department of Biomechanics in Sport, Technical University, Munich, Germany
[d] Department of Orthopedic Sports Medicine, Technical University, Munich, Germany

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ABSTRACT

Background: Varus knee alignment has been identified as a risk factor for the progression of medial knee osteoarthritis (OA). This study tested the hypothesis that not only frontal plane kinematics and kinetics but also transverse plane lower extremity mechanics during gait are affected by varus malalignment of the knee.

Methods: Eighteen, otherwise healthy children and adolescents with varus malalignment of the knee were studied to examine the association between static varus malalignment and functional gait parameters. Kinematic data were collected using a Vicon motion capture system (Vicon Motion Systems, Oxford, UK). Two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to collect kinetic data.

Results: The results indicated that changes in transverse plane mechanics occur concomitantly with changes in knee malalignment in the frontal plane. A mechanical consequence of varus knee malalignment is obviously an increased endorotation of the foot (internal foot placement) and an increased internal knee rotation (tibia rotation) during stance phase. The linear correlation between the maximum external knee adduction moment in terminal stance and the internal knee rotation in terminal stance ($r = 0.823$, $p < 0.001$) shows that this transverse plane gait mechanics is directly in conjunction with intrinsic compressive load on the medial compartment during gait.

Conclusions: Understanding factors that influence dynamic knee joint loading in healthy, varus malaligned knees may help us to identify risk factors that lead to OA. Thus, three-dimensional gait analysis could be used for clinical prognoses regarding the onset or progression of medial knee OA.

1. Introduction

Mechanical factors, such as varus malalignment of the knee in the frontal plane, have been implicated in the progression of medial knee OA [1–6]. A recent study by Hayashi et al. [6] has shown that varus knee alignment is associated with increased risk of incidence and enlarging bone marrow lesions. Quantitative gait analysis as an adjunct to static radiographic measures of alignment provides an estimate of tibiofemoral load through an inverse dynamic analysis and can be used to gain a better understanding of the biomechanical factors that contribute to the pathogenesis of knee OA [7,8]. The external knee adduction moment is used as the primary parameter for characterizing intrinsic compressive load and articular cartilage degeneration in the medial knee compartment during gait [4,5,9,10].

Previous studies have investigated biomechanical changes during gait in the sagittal and frontal planes in patients with knee OA compared to healthy controls [5,11,12]. For example, patients with medial knee OA make initial contact with the ground with the knee in a more extended position and have a smaller range of knee flexion during the stance phase of walking [11]. However, these studies on patients with established knee OA have made it difficult to determine if biomechanical changes are involved in the development of disease, are in response to degenerative changes in the joint, or are compensatory mechanisms in response to these degenerative changes or to joint pain.

In children and adolescents with varus malalignment of the knee without signs of knee OA abnormally increased knee internal rotation and hip external rotation moments were detected [13]. While transverse

* Corresponding author at: Orthopedic University Hospital Friedrichsheim gGmbH, Marienburgstraße 2, 60528 Frankfurt/Main, Germany. Tel.: +49 69 6705 862.
E-mail addresses: felix.stief@gmx.de, f.stief@friedrichsheim.de (F. Stief).

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plane mechanics can initiate degenerative changes to the articular cartilage and have been implicated in the progression of knee OA [10], there is a lack of research on the relationship between static varus malalignment of the knee and transverse plane lower extremity mechanics during gait. In particular, it is still not known which role the rotational gait profile plays in the progression of knee OA. The purpose of the present study was therefore to investigate the effect of pathological varus alignment of the knee on transverse plane gait parameters of the lower extremity in young individuals without signs of knee OA. The following question should be answered and may have important implications for the progression of knee OA: Does a relationship exist between static lower extremity malalignment in the frontal plane and transverse plane lower extremity mechanics during gait? We hypothesized that (1) the static varus alignment of the knee correlates with internal knee rotation (tibial rotation) and endorotation of the foot (internal foot placement) during the stance phase of gait, and (2) the rotational gait profile has an influence on the external knee adduction moment and thus also leads to degenerative changes in the knee joint in young patients with varus malalignment of the knee.

2. Methods

2.1. Subjects

Eighteen, otherwise healthy children and adolescents with varus malalignment of the knee 12–19 years of age were consecutively selected during clinical visits between January 2008 and October 2012 (Table 1). Solely patients were included with a clinical indication for a full-length standing anteroposterior radiograph and a pathological varus alignment of at least one knee according to the mechanical axis angle (MAA) of the lower limb [14]. Alignment was defined as pathological varus when the angle was more than 1.3° [14]. Patients were excluded if they had signs of OA or rheumatoid arthritis, anterior cruciate ligament (ACL) 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 lower extremity, knee surgery within the last 12 months, chronic joint infection, intraarticular corticosteroid injection, or morbid obesity according to the body mass index [15].

Fifteen healthy subjects were recruited as control group (Table 1). All subjects had undergone a clinical examination, including passive hip, knee and ankle range of motion (Table 1). The patients and the healthy subjects had normal strength and full range of motion in the lower extremities.

All subjects, and their parents in the case of subjects under age 18, were thoroughly familiarized with the gait analysis protocol. Subjects 18 and over gave written informed consent. Subjects younger than 18 gave verbal assent to study participation, and their parents provided written informed consent to participate in this study, as approved by the local ethics committee and in accordance with the Helsinki Declaration.

Table 1
Study population characteristics (mean with standard deviation in parenthesis) and *p*-values.

Variable	Patients	Controls	*p* -value
n	18	15	
Sex, no. female/no. male	11/7	9/6	
Age, years	15.2 (1.8)	15.1 (4.3)	0.941
Height, m	1.64 (0.18)	1.64 (0.15)	0.973
Body mass, kg	55.7 (11.2)	52.8 (12.8)	0.502
Body mass index, kg/m^2	20.6 (3.0)	19.2 (2.2)	0.144
Max. passive internal hip rotation (prone, hip neutral, knee flexed 90°), °	44.3 (13.7)	45.8 (3.6)	0.690
Max. passive external hip rotation (prone, hip neutral, knee flexed 90°), °	37.0 (14.6)	41.3 (3.1)	0.290
Max. passive tibial torsion (prone, hip neutral, knee flexed 90°), °	−12.0 (10.5)	−12.9 (4.5)	0.763
Self-selected walking speed, m/s	1.25 (0.10)	1.28 (0.11)	0.431
Mechanical axis angle, °	10.3 (7.0)		

External foot rotation (tibial torsion) is negative.

2.2. Gait analysis methods

Three-dimensional gait analysis was carried out using a Vicon motion capture system (Vicon Motion Systems, Oxford, UK) operating at a sampling rate of 200 Hz (Fig. 1). The level walkway was 15 m long and viewed by eight infrared cameras. Two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were situated at the mid-point of the walkway to collect kinetic data at 1000 Hz.

To improve the reliability and accuracy when analyzing frontal and transverse plane gait data, a custom made lower body protocol described in a previous investigation [16] was used. In addition to the standardized Helen Hayes marker set [17], reflective markers on the medial malleolus, medial femoral condyle and major trochanter were applied to determine positions of joint centers of rotation for the hip, knee and ankle. In contrast to the Helen Hayes marker set [17], the centers of rotation for the knee and ankle joints were defined statically as the midpoint between the medial and lateral femoral condyle and malleolus markers. This eliminates the reliance on the subjective palpation of the thigh and tibia wand markers, which is difficult to handle and less reliable within or between therapists than manual palpation [18]. The center of the hip joint was calculated using a geometrical prediction method [17]. The major trochanter marker was used to improve the prediction of the hip joint center by immediately calculating the distance between the anterior superior iliac spine and the major trochanter by anatomical landmarks [16]. Varus/valgus alignment of all knee joints was determined by rotating the local shank coordinate system about the *y*-axis of the thigh (vector from the lateral knee to the medial knee) until the *x*-axis of the shank (vector from the knee center to the ankle center) fell in line with the *x*-axis of the thigh (vector from the knee center to the hip center). According to Hunt et al. [7] using the same marker set for a similar purpose, this varus angulation provided an accurate, marker-based measure of lower limb alignment in the frontal plane. It has been shown that adding these few extra markers to the standard Helen Hayes marker set increases the repeatability in three-dimensional gait analysis, reduces the measurement errors when analyzing frontal and transverse plane gait data, and improves the accuracy of the knee joint axis by reducing the knee axis cross-talk phenomenon [16,19,20]. Furthermore, our lower body protocol enables the detection of significant effects in the tibiofemoral angles and moments in the frontal and transverse plane between three different patterns of movement [21]. Further processing incorporating force plate data allowed external moments (normalized to body weight) about the mathematically derived joint centers to be calculated.

Discrete variables of interest (Table 2) were calculated for five barefoot trials at a self-selected speed and then averaged for further analysis on the basis of complete marker trajectories and a clear foot-forceplate-contact. In the case of bilateral involvement, measurements were performed only on the limb with greater malalignment. This ensures that the statistical calculations were independent from each other. No significant differences were found in the discrete variables of interest comparing the left and right sides in the control group. Therefore, only one side – the left side was chosen here – was used for further analysis of the controls.

After each acquisition session, missing frames were handled with a fill-gap procedure using the Vicon-Nexus software version 1.7.1 (Vicon Motion Systems, Oxford, UK). The data were smoothed with a Woltring filter and using a smoothing spline [22]. The external knee adduction moment as well as the transverse plane gait parameters during stance phase were automatically determined by a custom made algorithm in Matlab 7.12.0 (The MathWorks, Inc., Natick, MA).