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The impact of increased femoral antetorsion on gait deviations in healthy adolescents ☆

Nathalie Alexander [a,b,*], Kathrin Studer [c], Harald Lengnick [c], Erika Payne [c], Harry Klima [c], Regina Wegener [a,b]

[a] Laboratory for Motion Analysis, Department of Paediatric Orthopaedics, Children's Hospital of Eastern Switzerland, St. Gallen, Switzerland
[b] Department of Orthopaedics and Traumatology, Cantonal Hospital St. Gallen, Switzerland
[c] Department of Paediatric Orthopaedics, Children's Hospital of Eastern Switzerland, St. Gallen, Switzerland

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ABSTRACT

Increased femoral antetorsion leads to several gait deviations, and amongst others, an increased knee flexion was reported in mid and terminal stance. Therefore, the purpose of this retrospective study was to identify gait deviations caused by increased femoral antetorsion and to perform subgroup analyses based on sagittal knee kinematics. Patients with isolated, CT confirmed increased femoral antetorsion (n = 42) and age-matched typically developing children (TDC, n = 17) were included in this study. Patients were referred to gait analysis because of gait abnormalities going along with an increased femoral antetorsion ≥30°. Kinematic and kinetic data were recorded during 3D gait analysis and three valid gait cycles were analyzed. Principal component (PC) analysis was used to achieve data transformation. A linear mixed model was used to estimate the group effect of PC-scores of retained PCs explaining 90% of the cumulative variance. Group effects of PC-scores revealed that patients walked with more flexed hips and greater anterior pelvic tilt throughout the gait cycle. Knee flexion was increased in patients during mid and terminal stance. Increased frontal plane knee and hip joint moments were found for patients compared to TDC. Furthermore, dividing patients into two subgroups based on their sagittal knee kinematics showed that kinematic gait deviations were more pronounced in patients with higher femoral antetorsion, while deviations in joint moments were more pronounced in patients with lower femoral antetorsion. Increased femoral antetorsion showed alterations in all lower limb joints and may be not only a cosmetic problem. Therefore, 3D gait analysis should be used for clinical management and operative treatment should be considered depending on severity of gait deviations.

1. Introduction

Femoral antetorsion refers to the twist between proximal and distal parts of the femur on the transverse plane (Kaiser et al., 2016). The normal amount of antetorsion depends on age and sex (Hefti, 2000), and is between 15 and 20° in adulthood (Crane, 1959; Fabry et al., 1973). Increased femoral torsion is often considered a primarily cosmetic problem since it often spontaneously decreases during growth. The definition of a pathological antetorsion is inconsistent in the literature ranging from >30° to 50° (Cordier and Katthagen, 2000; Hefti, 2000; Jani et al., 1979). Most literature suggests surgery to be the only possible treatment (Fabry, 2010; Hefti, 2000; Sass and Hassan, 2003), which should be indicated cautiously (Sass and Hassan, 2003).

Studies have shown that there are significant correlations with increased femoral torsion and orthopedic/musculoskeletal disorders (Ejnisman et al., 2013; Powers, 2003; Stevens et al., 2014). Powers (2003) and Stevens et al. (2014) reported a correlation of increased femoral torsion with patellofemoral pain. It is also a known risk factor for patellofemoral instability (Dejour and Le Coultre, 2007). Furthermore, excessive femoral antetorsion has been reported to be associated with femoroacetabular impingement (Audenaert et al., 2012; Ejnisman et al., 2013; Ito et al., 2001). Ejnisman et al. (2013) investigated hips with femoroacetabular impingement and reported that patients with femoral

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* Corresponding author at: Laboratory for Motion Analysis, Department of Paediatric Orthopaedics, Children's Hospital of Eastern Switzerland, Claudiusstrasse 6, 9006 St. Gallen, Switzerland.

E-mail address: nathalie.alexander@kispisg.ch (N. Alexander).

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version greater than 15° were 2.2 times more likely to have labral tears.

Femoral torsion is commonly assessed using computer tomography (CT) or magnetic resonance imaging (MRI) (Cordier and Katthagen, 2000; Hefti, 2000; Radler et al., 2010), but femoral antetorsion measured by CT-scans correlate weakly with hip internal rotation during walking (Radler et al., 2010). Furthermore, static techniques such as CT-scans do not take dynamic components of gait, such as primary and secondary deviations, into account and therefore three-dimensional (3D) gait analyses is recommended when planning a surgical correction of torsional deformities (Bruderer-Hofstetter et al., 2015; Radler et al., 2010). Altered joint kinematics and kinetics during walking can be caused by skeletal deformities affecting the mechanical axis of the lower limb (Gugenheim et al., 2004; Sobczak et al., 2012). Torsional deformities may stress adjacent joints (Bretin et al., 2011; Eckhoff, 1994; Tonnis and Heinecke, 1999). In the process of evaluating the altered joint kinematics and kinetics, 3D gait analysis is well-established for objectively measuring gait deviations and is frequently used for clinical decision-making (Wren et al., 2011).

Hip internal rotation gait is often studied in children with cerebral palsy (Dohin, 2017; Wren et al., 2013). There is, however, limited literature available about hip internal rotation gait in typically developing adolescents with torsional deformities (Bruderer-Hofstetter et al., 2015; Radler et al., 2010). Bruderer-Hofstetter et al. (2015) reported that patients with increased femoral antetorsion walked with smaller external foot progression angle, greater knee adduction, more internally rotated and flexed hips and greater anterior pelvic tilt. Additionally, an increased knee flexion in mid and terminal stance was found for patients with increased femoral antetorsion (Alexander et al., 2018; Passmore et al., 2018a).

Besides the studies mentioned, gait deviations due to increased femoral antetorsion in typically developing adolescents with torsional deformities have received little attention as increased femoral antetorsion <50° is most of the time not considered for surgery (Cordier and Katthagen, 2000; Fabry, 2010; Sass and Hassan, 2003; Tonnis and Heinecke, 1999). For example, currently there is no surgical indication for patients with isolated increased femoral antetorsion <40° according to the German guidelines for idiopathic coxa antetorta (Leitlinien der Orthopädie, 2002). Gait deviations due to increased femoral antetorsion, however, may cause future complaints. Studies have shown an association between femoral antetorsion and anterior knee pain (Eckhoff et al., 1997) as rotational limb alignment has a major impact on patellar kinematics (Keshmiri et al., 2016). This is also supported by increased mediolateral patellofemoral joint contact forces (Passmore et al., 2018a). In addition, Bretin et al. (2011) could show that rotational disorder may lead to joint arthrosis. Therefore understanding gait patterns in this patient group is important.

The aim of this study was to investigate kinematic and kinetic gait deviations in adolescents with isolated increased femoral antetorsion. It was hypothesized that increased knee flexion in mid and terminal stance caused by increased femoral antetorsion might have an influence on joint loading. Furthermore, changes in kinematic and/or kinetic patterns are expected when dividing patients into subgroups based on their sagittal knee kinematics.

Table 1
Mean (standard deviation) anthropometric data, walking speed and clinical examinations for patients and TDC.

	Patients (n = 42)	TDC (n = 17)
Femoral torsion [°]	38.9 (6.7)	–
Age [years]	12.9 (1.9)	13.5 (2.3)
Side [left/right]	30/29	17/15
Height [m]	1.57 (0.09)	1.59 (0.14)
Mass [kg]	45.6 (9.7)	47.1 (14.3)
Walking speed [m/s]	1.25 (0.11)	1.31 (0.13)
External hip rotation [°]*	15.3 (12.5)	31.3 (11.5)
Internal hip rotation [°]*	65.6 (14.7)	54.4 (10.5)
Knee extension [°]	4.7 (4.0)	1.1 (2.1)

* Indicates significant difference between groups.

2. Methods

2.1. Participants

Forty-two patients and 17 age-matched typically developing children (TDC) were retrospectively included from our overall patient pool (Table 1). The study was approved by the regional ethics board (EKOS 12/088) and written informed consent was provided by all participants and their legal guardian. In case of clinically increased femoral torsion >30° in the trochanteric prominence angle test (Ruwe et al., 1992), patients were referred to CT for confirmation and to gait analysis by the pediatric orthopedic outpatient clinic. CT measurements (Somatom Definition AS 64 slices, Siemens Healthineers, Erlangen, Germany) were performed on the basis of Hernandez et al. (1981) using a superposition of two figures at the femoral head and neck. The mean radiation dose was 3.5 mGy. Normal values of femoral antetorsion were defined as 15° ± 10° (Jani et al., 1979). Patients with unilateral or bilateral femoral antetorsion >30° in CT-scans were included in the study. Exclusion criteria for patients are summarized in Table 2. Data for TDC were previously collected in our gait laboratory as normative data. Exclusion criteria for TDC were: clinical indications of torsional deformities of femur or tibia, adiposity, leg length discrepancy >0.5 cm, previous surgery or fractures on limbs or back, need for orthoses correcting flat feet.

2.2. Data collection

Clinical examination included amongst others evaluation of hip rotation and knee extension. Anthropometric data were collected to rescale the Vicon Plug-in-Gait (PiG)-model. Self-reflecting markers were attached according to the PiG-model (Kadaba et al., 1990). Participants walked barefoot at a self-selected normal speed. Kinematic data were collected using an eight-camera, marker based motion capture system (Vicon, Oxford, Oxford Metrics Ltd, UK;

Table 2
Exclusion criteria for patients.

• Time period between CT and gait analysis >6 months	• Scoliosis
	• Recurrent patella instability
• Neurological diseases	• Impaired vision
• Hip pathologies (e.g. Perthes' disease)	• Treated foot deformities
• Chromosomal abnormality	• Age < 10 and > 18 years
• Congenital diseases	• Adiposity
• Leg length discrepancy >1 cm	• Regular intake of analgesics
• Co-existing tibial torsion deformities <25°; >41° Waidelich et al. (1992)	

200 Hz) and kinetic data were recorded via two force plates embedded in a 10.5 m walkway (AMTI, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA; 1000 Hz). Proper marker placement was checked using a static and dynamic trial prior to the measurement with a primary focus on the correct placement of the knee and thigh markers. Knee ab-/adduction motion >15° during swing phase indicated imprecise marker placement (Reinschmidt et al., 1997), and led to exclusion of three patients (values presented are for patients included in the study). Five valid gait cycles were collected, where participants hit the force plates without any visual interruption of the gait cycle.

2.3. Data analysis

From the five valid gait cycles, three representative gait cycles were taken for further analysis. Data were processed using Vicon Nexus (Vicon, Oxford Metrics Ltd, Oxford, UK). Kinematic and kinetic data were filtered using the Woltring method (mean squared error 10) and a 200 Hz Butterworth low pass filter, respectively. Joint angles were calculated in sagittal, frontal (coronal) and transverse planes. External net joint moments were calculated in sagittal and frontal (coronal) planes using an inverse dynamics approach (Davis et al., 1991) and were normalized to body mass. All data were time normalized to gait cycle duration.

2.4. Principal component analysis

Principal component (PC) analysis was performed using MATLAB (MathWorks Inc., Natick, MA) and was used as exploratory step to achieve data reduction (Chau, 2001). A cut-off criterion of 90% of the cumulative explained variance was used to determine the number of retained PCs (Bruderer-Hofstetter et al., 2015; Chau, 2001). Separate reconstructions of the original data were plotted as means for each group. The maximum coefficients within a PC-vector defined the main effects of the PC-scores.

2.5. Statistical analysis

All statistical analyses (level of significance $\alpha = 0.05$) were conducted in SPSS 20.0 (IBM Corporation, Amonk, NY). Because of correlated data (three gait cycles; both legs – where applicable) a linear mixed model (LMM) (Kuss and Watzke, 2005) was used to estimate the effects of *group* and *side* on PC-scores of retained PCs. To account for the fact that both legs were included for some participants, *side* effect was explicitly modeled. We were particularly interested in the fixed effects of *group* adjusted for possible sampling variations between *participants*, *gait cycles* and *sides*. PCs explaining less than 5% of the variance were not included in further analysis in order to focus on clinically relevant differences. Differences in participant characteristics between both groups were identified using a LMM for variables being leg dependent

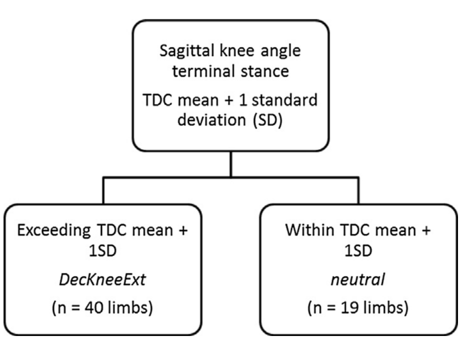

Fig. 1. Schematic presentation of sub group definition.

(e.g. hip internal rotation) and an independent Student's *t*-test or χ^2 test for other variables. Significances for clinical examinations were corrected for rounding error (Zdravkovic and Jost, 2018). Differences represented by PC reconstructions were categorized as magnitude operator, difference operator and phase shift operator (Wrigley et al., 2005).

Subsequently, patients were divided into two groups whether they showed decreased knee extension or not (Fig. 1). This was done by comparing waveforms in terminals stance and in case 80% of patients' sagittal knee angle waveform was above TDC mean + 1SD, it was defined as exceeding TDC data. Subgroup analyses in joint kinematics and kinetics between TDC and the two subgroups were performed using LMM with a Bonferroni corrected post hoc analysis.

3. Results

Age, height, mass and walking speed did not differ between patients and TDC ($p > 0.050$), but clinically determined external and internal hip rotations differed significantly ($p = 0.000$).

Significantly increased hip internal rotation, hip flexion, anterior pelvic tilt and decreased external foot progression angles were found in patients throughout the gait cycle. Furthermore, knee flexion was increased in mid and terminal stance (Table 3, Fig. 2). No differences were found for frontal hip kinematics (Table 3). No clinically relevant differences between patients and TDC were found for frontal knee kinematics since main differences were found in swing phase (Table 3), which is not interpreted.

Concerning joint loadings, an increased hip flexion moment in mid stance and decreased hip extension moment in terminal stance was found for patients (PC1 30.7%, $p = 0.000$). Hip adduction moment was increased for patients in terminal stance (PC1 48.8%, $p = 0.033$). Sagittal plane knee joint moments did not differ between patients and TDC (Fig. 2, PC1 47.5%, $p = 0.328$). Patients showed a higher knee adduction moment first peak and lower second peak (PC2 19.6%, $p = 0.000$, Fig. 2). Concerning magnitude, only medio-lateral ground reaction forces showed significant differences, with patients having higher and lower medio-lateral ground reaction forces at the first and second peak, respectively (PC2 12.1%, $p = 0.000$).

Subgroup analyses based on the sagittal knee kinematics revealed differences between all three groups (Table 5, Fig. 3). Femoral antetorsion based on CT scans was significantly higher ($p = 0.001$) for *DecKneeExt* compared to *neutral*, but no significant differences were found in clinical examination, walking speed and anthropometrics (Table 4). Increased hip flexion moment in mid stance (*DecKneeExt* > *neutral* > TDC; PC1 30.7%, $p \leq 0.001$) and decreased hip extension moment in terminal stance (*DecKneeExt* > *neutral* > TDC; PC1 30.7%, $p \leq 0.001$). Higher hip adduction moments were found for *neutral* in mid stance and especially terminal stance compared to *DecKneeExt* and TDC (PC1 48.8%, $p \leq 0.011$, Fig. 3). Lower knee flexion moments in mid stance and higher knee extension moments in terminal stance were found for *neutral* compared to *DecKneeExt* and TDC (PC1 47.5%, $p = 0.000$, Fig. 3). *Neutrals* showed a higher knee adduction moment (mid & terminal stance) compared to both other groups (PC1 56.7%, $p \leq 0.004$, Fig. 3). No magnitude differences between patient groups were found for ground reaction forces.

4. Discussion

The aim of this study was to investigate kinematic and kinetic gait deviations in adolescents with isolated increased femoral antetorsion. The hypothesis that increased knee flexion in mid and terminal stance might influence joint loading could be partly

Table 3
Significant principal components for joint kinematics and the patients' deviations in comparison to TDC (interpretation).

Principal component		Variance explained (%)	p-value	Operator	Interpretation
Sagittal pelvis angle	PC1	94.2	*0.000*	Magnitude	More anterior pelvic tilt
Sagittal hip angle	PC1	79.9	*0.000*	Magnitude	More hip flexion
	PC2	10.3	*0.022*	Difference	Relatively lower hip flexion (loading response) and lower hip extension (pre swing)
	PC3	6.1	*0.000*	Phase shift	Later hip flexion (terminal stance) and later hip extension (initial swing)
Transversal hip angle	PC1	72.1	*0.000*	Magnitude	Higher hip internal rotation
	PC2	9.9	*0.000*	Difference	Relatively more internal hip rotated hip in terminal stance and more external hip rotation during mid swing
Sagittal knee angle	PC1	47.8	*0.000*	Magnitude	Increased knee flexion in mid and terminal stance
	PC2	23.1	*0.016*	Phase shift	Earlier knee flexion (pre swing) and knee extension (mid swing)
	PC3	12.4	*0.000*	Magnitude	Lower knee flexion (initial swing)
	PC4	8.7	*0.000*	Difference	Relatively lower knee flexion (mid stance) and increased knee flexion (terminal stance)
Sagittal ankle angle	PC1	45.1	*0.018*	Magnitude	Lower plantar flexion in pre and initial swing
	PC2	21.4	*0.007*	Phase shift	Later plantar flexion in pre swing
Foot progression angle	PC1	77.6	*0.000*	Magnitude	More internal foot rotation
	PC3	5.8	*0.015*	Difference	Relatively more internal foot rotation in initial swing and more external foot rotation in terminal swing

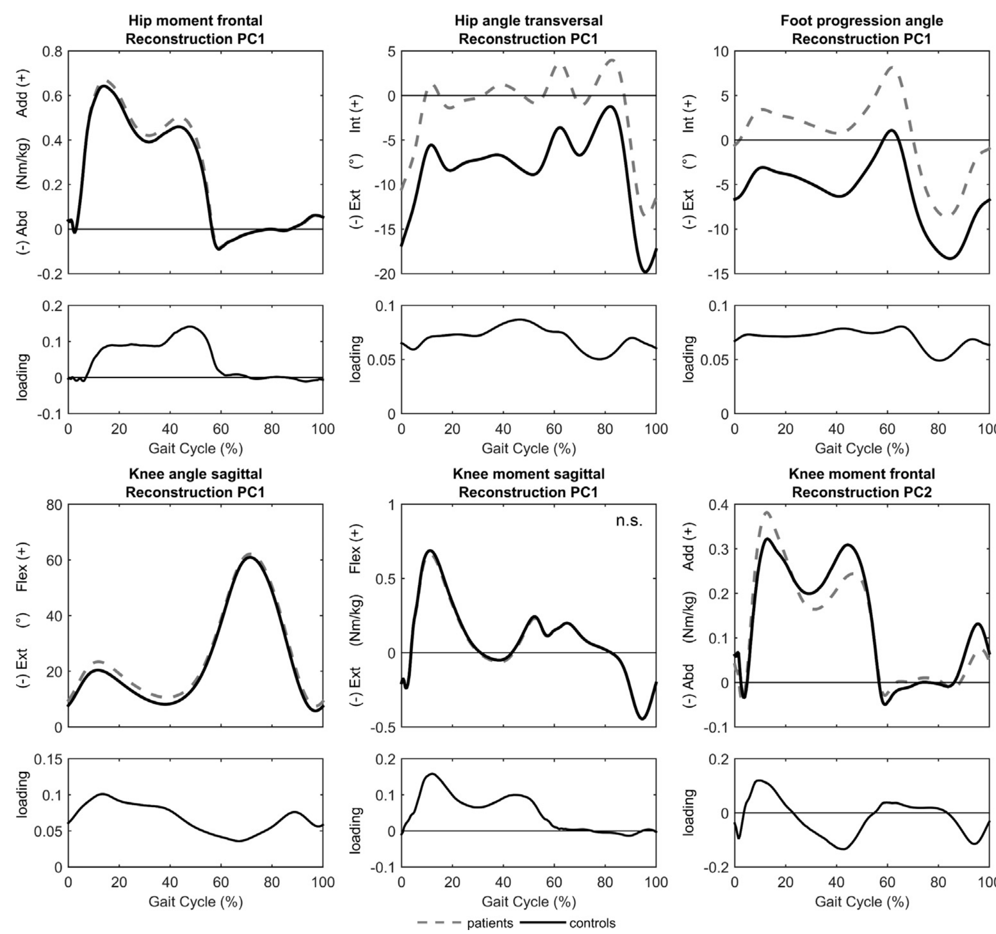

Fig. 2. Reconstructions of kinematic and kinetic waveforms as well as corresponding PC loadings are present for patients (grey dashed line) and TDC (black solid line). Except for the sagittal knee moment, all reconstructions showed significant differences between patients and TDC.

accepted in terms that this gait deviation might be an indicator for severity of increased femoral antetorsion.

Increased femoral antetorsion with clinically decreased external hip rotation led to patients walking with more internal hip rotation (Fig. 2). During standing, internal rotation of the lower extremity was shown to result in an anterior pelvic tilt as a direct result of bony approximation between the femoral head and the acetabulum (Duval et al., 2010). Bagwell et al. (2016) reported a consistent pattern of kinematic coupling of anterior pelvic tilt and internal femur rotation with 1.2–1.6° of internal femur rotation for every 5° of anterior pelvic tilting. In line with previous literature (Bruderer-Hofstetter et al., 2015) patients walked with increased anterior pelvic tilt and following greater hip flexion. In our patients the increased hip flexion led to an increased hip flexion moment in mid stance and decreased hip extension moment in terminal stance.

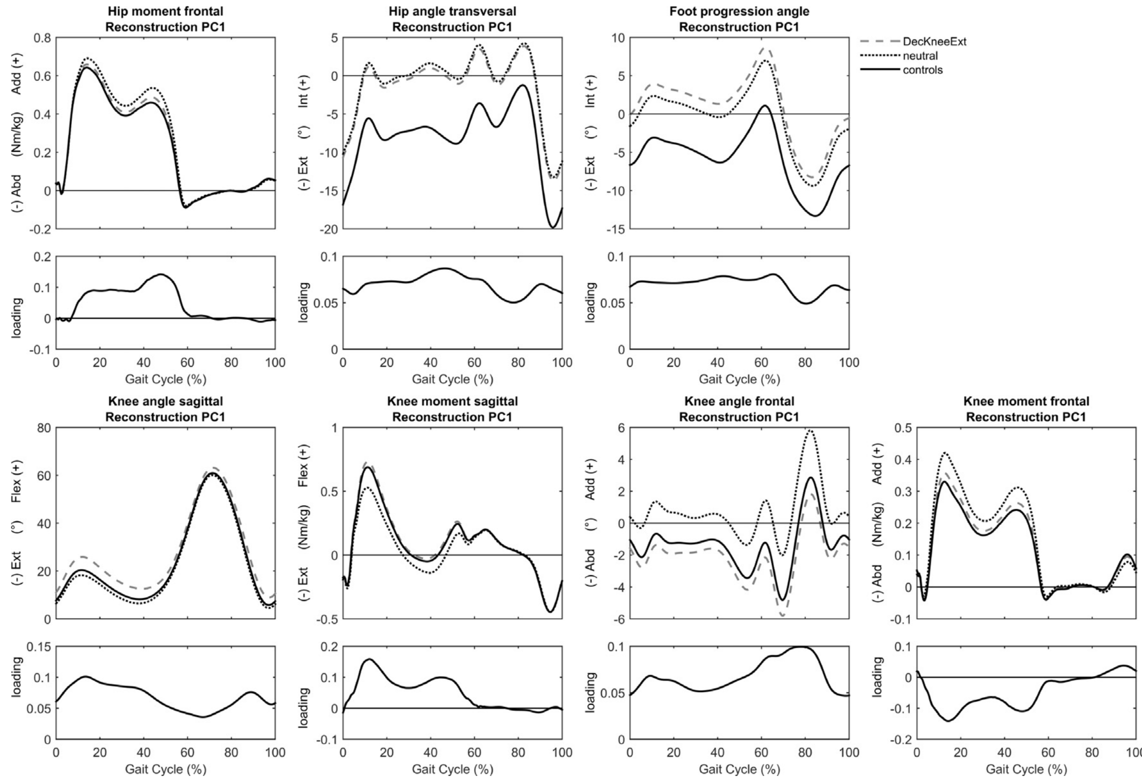

Fig. 3. Reconstructions of significantly different kinematic and kinetic waveforms as well as corresponding PC loadings for patients showing a decreased knee extension (*DecKneeExt*, grey dashed line), patients showing normal knee extension (*neutral*, black dotted line) and TDC (black solid line).

Table 4
Mean (standard deviation) anthropometric data, walking speed and clinical examinations for both patients groups DecKneeExt and neutral.

	DecKneeExt	Neutral
Femoral torsion [°]*	40.3 (7.2)	35.7 (3.7)
Age [years]	12.7 (1.8)	13.3 (2.2)
Side [left/right]	18/23	12/6
Height [m]	1.55 (0.09)	1.60 (0.10)
Mass [kg]	43.6 (7.6)	50.1 (12.5)
Walking speed [m/s]	1.25 (0.10)	1.25 (0.12)
External hip rotation [°]	13.5 (11.5)	19.4 (13.8)
Internal hip rotation [°]	66.8 (15.5)	62.8 (12.9)
Knee extension [°]	5.1 (4)	3.6 (4.1)

* Indicates significant difference between groups.

Furthermore, increased internal hip rotation during walking might be a compensatory mechanism to restore the moment arm of the hip abductors (Arnold et al., 1997). Passmore et al. (2018a) observed in a similar group of patients increased internal hip rotation and an increased force production of gluteus medius and minimus to maintain an internal hip abductor moment comparable to controls. In the current study increased hip adduction moments in terminal stance with no changes in frontal hip kinematics were found for patients. These indicate an increased generation of hip abduction muscle torque. In contrast to the study of Passmore et al. (2018a), who included patients with increased femoral antetorsion and increased external tibial torsion, the external foot progression angle in our patients was decreased. Toeing-in might have changed the lever arms of the acting forces possibly explaining the increased hip adduction moment.

Furthermore, increased knee flexion in mid stance and a decreased knee extension in terminal stance were not caused by a decreased joint range of motion since patients could achieve full knee extension with a bias towards knee hyper-extension as previously reported in the literature (Passmore et al., 2018a). A reduced efficacy of the plantar flexor-knee extension couple mechanism (Passmore et al., 2018a) as well as reduce hip stability due to the decreased abduction lever arm of gluteus medius (Akalan et al., 2013; Arnold et al., 1997) were previously reported as causes for compensatory hip and knee flexion. The efficacy of the plantar flexor-knee extension couple could be affected by the internal foot progression angle and altered external lever arms possibly being the reason for the observed decreased plantar flexion in pre and initial swing.

Another hypothesis for the reduced knee extension might be that gluteal muscles are limited in their capacity to extend the hip joint due to altered lever arms possibly leading to a compensatory contraction of the ischiocrural muscles as hip extensors. These muscles also have a flexion effect on the knee joint possibly leading to an increased knee flexion during gait. Literature reported that at initial contact, the ischiocrural muscles actively contract to flex the knee due to the lever arm situation at the knee under load, but as the leg is loaded the direction of activity changes and they become hip extensors (Brunner and Rutz, 2013). This hypothesis might be supported by unaltered knee joint moments in terminal stance, due to the support of the ischiocrural muscles. At the current state this, however, remains hypothetical and would need further verification for example by using electromyography.

Increased frontal plane hip and knee moments might also be attributed to the increased internal hip rotation gait and an altered lever of the ground reaction force due to internal foot progression. With in-toeing, however, an increased knee adduction moment in terminal stance might be expected (Krackow et al., 2011) rather than a decreased knee adduction moment as presented in the current study. Decreased knee adduction moments, however, are in agreement with the study of Passmore et al. (2018a).

In a next step, patients included in this study were divided into two subgroups based on their sagittal plane gait deviations in ter-

Table 5
Significant principal components for joint kinematics for subgroup analyses (two patient groups and TDC).

Principal component		Variance explained (%)	Group effect	DecKneeExt vs. neutral	DecKneeExt vs. TDC	Neutral vs. TDC	Interpretation
Sagittal pelvis angle	PC1	94.2	*0.000*	*1.000*	*0.000*	*0.000*	Both patient groups show more pelvic anterior tilt
Sagittal hip angle	PC1	79.9	*0.000*	*0.025*	*0.000*	*0.008*	Increased hip flexion: DecKneeExt > neutral > norm
Frontal hip angle	PC2	14.1	*0.002*	*0.003*	*0.035*	*0.698*	DecKneeExt: later start of hip adduction movement (loading response); longer/more hip adduction (terminal stance)
Transversal hip angle	PC1	72.1	*0.000*	*1.000*	*0.000*	*0.000*	Both patient groups show more hip internal rotation
	PC2	9.9	*0.000*	*0.009*	*0.000*	*0.800*	DecKneeExt: relatively more hip internal rotation (terminal stance) and lower hip internal rotation (mid swing)
Sagittal knee angle	PC1	47.8	*0.000*	*0.000*	*0.000*	*0.011*	DecKneeExt: higher knee flexion (mid & terminal stance) Neutral: decreased knee flexion (mid & terminal stance)
	PC3	12.4	*0.000*	*0.170*	*0.000*	*0.007*	DecKneeExt: later & less knee flexion (swing phase)
	PC4	8.7	*0.000*	*0.001*	*0.000*	*0.007*	Relatively more knee flexion in mid stance (TDC > neutral > DecKneeExt) and more knee extension in terminal stance (TDC > neutral > DecKneeExt)
Frontal knee angle	PC1	62.8	*0.000*	*0.000*	*0.247*	*0.000*	Neutral: more knee adduction (changes are more distinct in swing and loading response, than in terminal stance)
	PC2	18.2	*0.000*	*0.070*	*0.000*	*0.197*	Swing phase difference - no clinical relevance
Foot progression angle	PC1	77.6	*0.000*	*0.031*	*0.000*	*0.000*	Increased internal foot rotation DecKneeExt > neutral > TDC
	PC3	5.8	*0.006*	*0.104*	*0.006*	*1.000*	DecKneeExt increased foot rotation in swing (internal and external) compared to TDC

minal stance (decreased knee extension yes/no). A decreased knee extension in terminal stance was shown by 67.8% of the patients. *DecKneeExt* have on average 4.8° higher femoral antetorsion based on CT scans compared to *neutral*. Increased anterior pelvic tilt and increased hip internal rotation did not differ between subgroups. Similar hip rotation, even though differences in femoral antetorsion were found, is in agreement with literature showing a very weak correlation between femoral torsion and hip rotation due to considerable dynamic influence of compensation mechanisms during walking (Radler et al., 2010).

In *DecKneeExt* the increased knee flexion in mid and terminal stance did not alter knee joint moments, which might be explained by a knee extension support of ischiocrural muscles. Increased knee flexion might result in a relatively reduced limb length, but no differences in pelvic obliquity were found. Increased knee flexion in terminal stance might promote the tendency towards longer and increased hip adduction as well as slightly less knee varus compared to *neutrals*. We assumed that in *DecKneeExt* the combination of increased knee flexion, hip internal rotation and internal foot progression led to a hip adduction moment comparable to TDC.

In *neutral* the lower knee flexion moment in mid stance and higher knee extension moment in terminal stance might be due to the knee extension support of ischiocrural muscles. Increased hip adduction moments might be explained by the internal foot progression and more knee varus leading to the medialization of the force vector and therefore to an increased hip adduction moment. Furthermore, more knee varus also resulted in higher knee adduction moments, which might be associated with knee osteoarthritis (Telfer et al., 2017).

Differences between both patient groups might not only result from different femoral antetorsion but could also be due to different compensatory strategies. This might be supported by the fact that no differences in the clinical examination were found between both patient groups.

Observed differences of both patient groups in the frontal plane are within one standard deviation of TDC means. Increased knee adduction moments, however, were previously associated with knee osteoarthritis even with small increases in joint moments of on average 0.10% bodyweight * height (Telfer et al., 2017). Even though the effect of increased knee adduction moments in adolescents with increased femoral antetorsion on the onset of osteoarthritis later on are not know, the differences found might be clinically relevant.

Consistent with the previous published literature (Passmore et al., 2018a) reporting increased hip and patellofemoral joint forces, our results underline that increased femoral antetorsion leads to kinematic gait deviations and joint loading alterations. Joint moment alterations could present a risk for pathological joint loadings with the development of osteoarthritis. Furthermore, subgroup analyses showed that kinematic gait deviations were more pronounced in patients with higher femoral antetorsion (*DecKneeExt*), while deviations in joint moments were more pronounced in patients with lower femoral antetorsion (*neutral*). This is an important point and highlights that not only severely increased femoral antetorsion needs to be addressed, but also underlying compensation mechanisms seem important. As femoral antetorsion is reported to be a risk factor for femoroacetabular impingement (Audenaert et al., 2012; Ejnisman et al., 2013; Ito et al., 2001) further research has to analyse which factors predispose to early osteoarthritis of the hip. In the future, operative treatment should be considered depending on severity of gait deviations due to the possible wide-ranging consequences of compensation mechanisms and secondary effects.

Finally, following limitations should be considered. Hip rotation kinematics have been reported as least repeatable parameters in clinical gait analysis (McGinley et al., 2009), however, hip rotations are not the only gait deviations described in the current study. The used PiG model has limitations such as the hip joint center position (Sangeux et al., 2011) or that misplacement of the thigh markers lead to erroneous definition of the frontal plane of the femur (Kadaba et al., 1990). However, we are confident that these errors were kept to a minimum in this study due to the meticulous measurement protocol. To improve the quality of data the combination of gait analysis with the use of low dose biplanar radiographs (EOS imaging) to define anatomical landmarks has been proposed (Passmore et al., 2018b). Unfortunately, this is not yet available in our hospital. Decreased external hip rotation could also result from the anterior wall of the acetabulum blocking external rotation, even though radiologists checked for this in CT-measurements. Furthermore, patients with increased femoral antetorsion were included based on clinical evaluation of hip rotation, which was verified in CT-scans, possibly introducing a selection bias. The same clinical examination of torsional profile was performed for TDC but without verification in CT-scans. Accuracy of torsional profile clinical examination is reported controversial in the literature (Cordier and Katthagen, 2000; Sangeux et al., 2014), but it was suggested that clinical examination can be considered as screening technique (Sangeux et al., 2014). Therefore, we cannot preclude that TDC were possibly affected by increased femoral antetorsion.

Concluding, the current results help for a better understanding of gait deviations due to isolated increased femoral antetorsion. Those gait deviations also highlight the importance of 3D gait analysis additional to static measurements. Since surgery is the only effective treatment in patients with increased femoral antetorsion, research is needed to better understand which patients benefit most from surgical treatments, despite the existing strict indications to date. Resulting gait deviations should be taken seriously since effects on joint loading have been shown in the current as well as previous studies.

Conflict of interest

None of the authors had any financial or personal conflict of interest with regard to this study.

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