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Effects of varus knee alignment on gait biomechanics and lower limb muscle activity in boys: A cross sectional study

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ABSTRACT

Background: There is evidence that frontal plane lower limb malalignment (e.g., genu varus) is a risk factor for knee osteoarthritis development. However, only scarce information is available on gait biomechanics and muscle activity in boys with genu varus.

Research question: To examine the effects of knee varus alignment on lower limb kinematics, kinetics and muscular activity during walking at self-selected speed in boys with genu varus versus healthy age-matched controls.

Methods: Thirty-six boys were enrolled in this study and divided into a group of boys with genu varus (n = 18; age: 11.66 ± 1.64 years) and healthy controls (n = 18; age: 11.44 ± 1.78 years). Three-dimensional kinematics, ground reaction forces, loading rates, impulses and free moments of both limbs were recorded during five walking trials at self-selected speed. Surface electromyography was recorded for rectus femoris and vastus lateralis/medialis muscles.

Results: No significant between-group differences were found for gait speed. Participants in the genu varus group versus controls showed larger peak knee flexion ($p = 0.030$; $d = 0.77$), peak knee adduction ($p < 0.001$; $d = 1.63$), and peak ankle eversion angles ($p < 0.001$; $d = 2.06$). Significantly higher peak ground reaction forces were found at heel contact (vertical [$p = 0.002$; $d = 1.16$] and posterior [$p < 0.001$; $d = 1.63$] components) and at push off (vertical [$p = 0.010$; $d = 0.93$] and anterior [$p < 0.001$; $d = 1.34$] components) for genu varus versus controls. Peak medial ground reaction force ($p = 0.032$; $d = 0.76$), vertical loading rate ($p < 0.001$; $d = 1.52$), anterior-posterior impulse ($p = 0.011$; $d = 0.92$), and peak negative free moment ($p = 0.030$; $d = 0.77$) were significantly higher in genu varus. Finally, time to reach peak forces was significantly shorter in genu varus boys compared with healthy controls ($p < 0.01$; $d = 0.73–1.60$). The genu varus group showed higher activities in vastus lateralis ($p < 0.001$; $d = 1.82$) and vastus medialis ($p = 0.013$; $d = 0.90$) during the loading phase of walking.

Significance: Our study revealed genu varus specific gait characteristics and muscle activities. Greater knee adduction angle in genu varus boys may increase the load on the medial compartment of the knee joint. The observed characteristics in lower limb biomechanics and muscle activity could play a role in the early development of knee osteoarthritis in genu varus boys.

1. Introduction

Lower extremity deformities are among the most common diagnosed non-traumatic abnormalities in youth [1]. There is evidence that genu varus (GV) represents an important contributor to the progression of medial knee osteoarthritis (OA) in adulthood [2]. Therefore, it is important to understand risk factors that are associated with knee OA.

A previous study indicated that each degree of tibiofemoral malalignment resulted in a 51 N increase in medial compartment contact force during walking [3]. GV individuals aged 12–19 years compared with healthy controls showed significantly smaller knee extension angles and lower maximum knee extension moments during the terminal stance phase of walking [4]. The same authors revealed significantly higher maximum knee adduction moments during mid and terminal

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stance in youth with GV [4]. In addition, higher maximum hip abduction moments were found during the loading phase together with higher peak internal tibia rotation during the stance phase [4,5]. Taken together, these findings imply that GV does not only affect the frontal and sagittal planes during walking [4] but also the transverse plane [5].

More pronounced knee adduction moments are associated with higher knee joint loads in the frontal plane which could ultimately lead to the development of knee OA [6]. Changes in knee alignment (for example GV) affect the lever arm, thereby modifying the amplitude of ground reaction forces (GRF) and knee adduction moments. In the more distal part of the lower limb, GV-related malalignment causes a deviation of the tibial mechanical axis from perpendicular [7]. Thus, a possible consequence of varus angulation at foot strike is that the applied force is medially directed which results in increased lateral GRF [8]. Accordingly, GRF testing could be an adequate biomechanical means to evaluate the gait pattern of boys with GV.

A previous study demonstrated that males and females aged 18–35 years with GV exhibited greater peak knee adduction angles and moments, midstance knee flexion angles and excursion, and eversion angles and lateral ground reaction forces, but smaller peak hip adduction angles [8]. However, these studies did not report GRF characteristics such as three-dimensional peak amplitudes, time to peak (TTP), vertical loading rate, impulses in all axes, and peak positive or negative free moment (FM) amplitudes during walking in boys with GV. In addition, little is known on the effects of GV on quadriceps muscle activity during walking in boys with GV compared with healthy controls. Therefore, further studies should be conducted to assess quadriceps strength and how it is associated with loading rate or muscle activity during walking.

GRF is an approximate measure of lower-limb loading and relatively easy-to-administer [9,10]. The vertical GRF during heel strike in gait produces stress waves that are transmitted from distal to proximal segments of the lower extremities [11]. These stress waves appear to be harmful when applied over short periods of time [12]. Given that human tissue is viscoelastic, its loading response is time dependent (loading rates) and more prone to injury at higher loading rates [13]. The vertical loading rate is defined as the slope of the initial part of the vertical GRF-time curve (i.e., between the foot strike and the vertical impact peak) [14]. Previous studies reported associations between high vertical GRF loading rates and joint degenerative processes [12,13]. There is evidence that high loading rates and impact transients are related to overuse injuries sustained during walking (i.e., knee OA) [13]. However, the influence of GV on the medial and lateral GRF components in boys is not well-established. The lateral GRF may contribute to foot pronation which when excessive has been linked to lower limb (i.e., knee joint) pain [15]. Therefore, it is worthwhile to evaluate differences in biomechanical gait parameters between boys with GV and healthy age-matched peers to better understand GV-related biomechanical stress on joints of the lower limbs during walking. These information can be used to develop specific therapeutic means for individuals with GV. The FM represents the vertical moment applied to the center of pressure and is considered to be sensitive to the movement and the imbalances of the whole body in the transverse plane [16]. Higher FM amplitudes are associated with torsion between pelvis and the foot [17]. This mechanism appears to cause tibial torsional stress and may thus increase the risk of sustaining fractures during walking [18]. This highlights the importance of FM data for clinically relevant research questions on biological tissue loading during walking. Of note, impulse is the amount of force applied over a period of time and is associated with total body momentum [19].

It has previously been reported that muscles contribute to joint loading. Retrospective [20] and prospective [21] studies indicate that quadriceps weakness is a major contributor to knee OA development. Given that the quadriceps contributes to the attenuation of impact forces during the early stance phase of gait, quadriceps weakness could be partly responsible for high joint loading [22]. Several testing modes (e.g., isometric, isokinetic) and outcome measures (e.g., peak torque, rate of force/torque development [RFD/RTD]) have previously been used for the assessment of quadriceps weakness [23]. The detection of quadriceps strength markers (e.g., RFD, RTD) that are associated with impulsive loading during walking could provide important information for the identification of individuals who are at elevated risk of knee OA development. RFD/RTD are defined as the change in force/torque over change in time during an isometric contraction [24]. It indicates how rapidly a group of muscles can generate force/torque. RFD/RTD is a clinically relevant and rather easy-to-assess force production parameter that can be used in GV individuals to primarily detect deficits in force production capacity that may also affect GV individuals' gait characteristics. It has previously been showed that the extensor muscles encompassing a varus knee function less efficiently than those encompassing a normally aligned knee [25]. Given that the quadriceps is a major contributor to attenuate impact forces during the early stance phase of gait, this deficit in muscle activity could potentially result in impulsive or high loading rates [26]. The relationship between quadriceps strength and loading rate has yet to be evaluated in individuals with GV. Of note, an altered GV-related mechanical environment of the knee could cause modified activation patterns of the major muscles surrounding the knee joint. A previous study reported increased electromyography (EMG) amplitudes of quadriceps muscles in the presence of knee OA [27]. However, to date, no study has evaluated EMG activity of quadriceps muscles in boys with GV. Previous studies reported significant bilateral differences in cartilage volume [28], bone density (in participants aged 23–56 years) [29], and muscle strength as well as motor coordination abilities in the lower limb joints [30,31]. These asymmetries in lower limb anatomy may lead to different joint biomechanics in boys with GV.

Therefore, the purpose of this study was to examine the effects of varus alignment of the knee on kinematics and kinetics as well as lower limb muscle strength and activity in boys with GV compared with healthy age-matched peers. In this regards, GV was defined as a mechanical axis angle higher than 1.3° in both knees [5,32]. With reference to the relevant literature [4,33], we hypothesized that vertical (Fz) and mediolateral (Fx) GRFs during walking are higher in boys with GV compared to their healthy peers. In addition, we expected higher loading rates, mediolateral impulses, FM, quadriceps muscular activity and lower quadriceps strength and knee flexion angles in children with GV compared to their healthy peers which is indicative of an altered neuromuscular activation strategy.

2. Material and methods

Written informed consent was obtained from all participants and their legal representatives. Local ethics approval was granted and the study was conducted in accordance with the latest version of the declaration of Helsinki.

2.1. Participants

The power analysis (G*Power software) was computed with an assumed Type I error of 0.05, a Type II error rate of 0.20 (80% statistical power) and an effect size of 0.80 [33] for running kinetics (e.g., GRF and moments). The analysis revealed that 32 boys would be sufficient to observe large between-group differences. Due to potential drop outs, 36 boys aged 9–14 years were enrolled in this study. Eligible participants were divided into two groups, boys with GV and healthy age-matched peers (Table 1). Boys with GV were identified through clinical examinations (Table 1). Of note, boys with GV were included if they had a mechanical axis angle (MAA) higher than 1.3° in both knees [5,32]. Finally, 18 GV boys and 18 healthy boys were eligible for inclusion in this study. An individual was included in the control group if he did not show any signs of musculoskeletal, postural or neurological disorders. Exclusion criteria for the GV and the control group comprised musculoskeletal, neuromuscular, and/or othopedic disorders in the 6

Table 1
Group-specific means and standard deviations of all examined demographic characteristics.

Variable	Control	Genu Varus	p
Number	18	18	NA
Age (years)	11.44 ± 1.78	11.66 ± 1.64	0.701
Height (m)	1.40 ± 0.07	1.40 ± 0.08	0.950
Mass (kg)	36.39 ± 12.13	36.55 ± 11.44	0.968
BMI (kg/m^2)	18.08 ± 4.47	18.13 ± 4.04	0.974
Dominant MAA	0.6 ± 0.5	9.5 ± 6.1	0.001*
Non-dominant MAA	0.5 ± 0.5	9.2 ± 5.3	0.001*

MAA, Mechanical axis angle; BMI, Body mass index; NA, Not applicable; p, *t*-test significance. * Significant difference between both groups.

months prior to the start of the study (except for GV malalignment in the GV group), and limb length asymmetry ≥ 5 mm. In addition, we advised participants not to conduct strenuous exercise 48 h prior to testing. Inclusion and exclusion criteria were assessed by medical doctors. The dominant lower limb was determined using a kicking ball test [34].

2.2. Apparatus

Kinematic data were collected using a six-camera VICON motion capture system (Vicon system, Oxford Metrics, Oxford, UK) and 16 spherical reflective markers with a diameter of 15 mm at a sampling rate of 100 Hz. A Plug in Gait marker set from Nexus software was used to identify the bilateral pelvis, thighs, legs, and feet. A previous study reported acceptable reliability [35]. Two force platforms (Kistler, type 9281, Kistler Instrument AG, Winterthur, Switzerland) were used to record GRF data from each leg at a sampling rate of 1000 Hz. A portable EMG system (BTS FREE EMG 300, BTS Bioengineering, Italy) with three pairs of bipolar Ag/AgCl surface electrodes (circular in shape with 11 mm in diameter; 25 mm center-to-center distance; input impedance of 100 MΩ; and common mode rejection ratio of > 110 dB at 50–60 Hz) was used to record the activity of the vastus lateralis (VL), vastus medialis (VM), and rectus femoris (RF) muscles at a sampling frequency of 2000 Hz. All procedures related to skin preparation and electrode placement were conducted according to the European recommendations for surface EMG [36].

2.3. Gait analysis

Before the tests started, reflective markers were attached bilaterally to the anatomical landmarks of our participants based on Plug in Gait marker set [37]. Then, one static trial during standing was collected [35]. Five successful trials were recorded at preferred gait speed. A trial was accepted if subjects showed a normal stride pattern (i.e., without any loss of balance and without falling during walking). Trials were excluded if participants hit the edge of the force plate during heel strike or if they completely missed the force plate. Timing gates were applied to maintain consistency in walking speed between trials for each participant. GRF and kinematic data were filtered using a fourth-order low-pass Butterworth filter with a 20 Hz and 6 Hz cutoff frequency, respectively [38]. Both GRF and kinematic data were calculated during the stance phase of walking which was defined as the interval from ground contact (vertical GRF > 10 N) to toe off (vertical GRF < 10 N). Plug-in-Gait lower body model (Vicon Motion Systems) was utilized for data processing, and graphical reports were created in Polygon Authoring Tool (3.5.1, Vicon, Oxford, UK).

The kinematic variables were maximum ankle eversion angle, maximum knee flexion angle during loading response, and maximum knee adduction angle was assessed during the entire stance phase. Considered GRF variables were selected features of GRFs and their time of occurrence along the mediolateral (Fx), anteroposterior (Fy), and vertical (Fz) axes with respect to the ground. Peak GRF amplitudes, their TTP, vertical loading rate, and FM were calculated as described in Jafarnezhadgero et al. [33]. Forces were normalized to body weight (BW) and the corresponding timing was expressed as a percentage of stance-phase duration (% Stance). All FM waveforms were amplitude-normalized to the product of each individual's BW (N) and height (m).

EMG signals were processed as described in Farahpour et al. [37]. For EMG analysis, the peak root mean square (RMS) values (during loading response phase of walking) of the five trials were averaged and then normalized based on the peak RMS obtained by the maximum voluntary isometric contraction (MVIC) for the quadriceps muscles [37]. Ensemble average EMG for the 3 muscles was also calculated [37].

2.4. Radiographic methods

Radiographic assessment of lower extremity alignment was conducted as was introduced by Stief et al. [5]. For each radiograph, lower extremity mechanical axis angle (MAA) was calculated to quantify knee alignment in the frontal plane [5]. Radiographic assessment of lower limb alignment was conducted on the same day as the gait analysis [5]. Participants stood barefoot in a forward knee position with the patella centered over the femoral condyles with the feet pointing straight ahead to control for any foot rotation effects [39] and to attain a true full-length weight bearing anteroposterior radiograph [40].

2.5. Quadriceps strength assessment

Quadriceps strength was assessed bilaterally under isometric conditions [23]. For this purpose, participants were placed on a dynamometer (HUMAC NORM, CSMi, Stoughton, MA) with straps secured over the torso, thigh, and leg to isolate quadriceps contributions to the torque signal [23]. During isometric assessment, the knee and hip were in 90° and 85° flexed position, respectively [23]. Three trials were recorded (Torque data were sampled at 600 Hz and low pass filtered at 50 Hz) from which the maximal value for each outcome measure was used for further analysis.

The dependent variables used for further analyses were peak torque and RTD of the knee extensors. In accordance with Kline et al. [41], RTD was calculated as the slope of the torque-time curve from 20 to 80% peak torque, as well as the slope from 0 to 100 ms (RTD_100) following the onset of torque (i.e., torque ≥ 20 Nm). Peak torque and RTD were normalized to body mass (Nm/kg and Nm/kg·s-1, respectively).

2.6. Statistical analyses

After data normality assumption was verified with the Shapiro-Wilk test, a MANOVA test was used to detect between group differences. The significance level was set at $p < 0.05$. In addition, as an effect size, Cohen's d was computed. Relationships between measures of muscle strength and walking biomechanics (e.g., loading rate) were calculated using Pearson's product moment correlation. Correlation coefficients were defined as very weak (0 to 0.19), weak (0.2 to 0.39), moderate (0.4 to 0.59), strong (0.6 to 0.79), and very strong (≥0.8). The statistical package for the social sciences (SPSS version 22) was used for data analysis.

3. Results

There was no significant difference in walking speed between the GV (1.24 ± 0.05 m/s) and the control group (1.27 ± 0.12 m/s) ($p = 0.505$; $d = 0.35$).

The statistical analysis indicated significant between group differences with GV boys showing higher values in vertical GRF during heel contact (Δ19%; $p = 0.002$; $d = 1.16$), vertical GRF during push off

Table 2
Group-specific means and standard deviations of all examined kinetic parameters.

	Variable	Component	Genu varus		Healthy control		
			Dominant	Non-dominant	Dominant	Non-dominant	Group (*p*-value (Cohen's d))
GRF (%BW)	Fz	FzHC	154.9 (23.6)	150.4 (22.8)	128.1 (23.4)	128.4 (22.5)	*0.002 (1.164)
		FzDF	73.7 (16.1)	75.9 (12.1)	79.9 (15.3)	77.9 (10.4)	0.331 (0.339)
		FzPO	128.9 (13.6)	133.1 (14.9)	120.4 (17.6)	115.5 (18.9)	*0.010 (0.931)
	Fy	FyHC	63.7 (24.4)	55.4 (14.1)	39.2 (8.4)	40.1 (10.3)	* < 0.001 (1.636)
		FyPO	56.4 (10.4)	58.4 (11.9)	48.1 (8.4)	42.4 (10.4)	< 0.001 (1.344)*
	Fx	FxHC	13.9 (8.2)	13.5 (11.1)	7.6 (4.2)	7.7 (4.7)	0.013 (0.896)
		FxMS	15.1 (9.4)	16.2 (6.7)	11.2 (2.9)	12.2 (3.8)	*0.032 (0.766)
		FxPO	11.4 (6.3)	12.1 (8.1)	8.5 (3.9)	10.9 (3.1)	0.241 (0.408)
TTP of GRF (%stance)	Fz	FzHC	17.8 (6.2)	18.0 (4.4)	23.6 (1.6)	22.5 (2.4)	* < 0.001 (1.603)
		FzDF	42.8 (8.0)	45.5 (6.7)	47.3 (2.5)	47.2 (3.5)	0.039 (0.735)*
		FzPO	76.4 (4.9)	76.0 (3.1)	77.5 (1.2)	76.7 (1.5)	0.298 (0.364)
	Fy	Fy_{HC}	11.6 (4.8)	13.5 (3.8)	16.6 (4.1)	16.7 (2.7)	* < 0.001 (1.553)
		Fy_{PO}	86.4 (2.2)	84.3 (8.7)	88.3 (1.6)	86.7 (2.6)	0.111 (0.561)
	Fx	Fx_{HC}	5.0 (1.9)	4.7 (1.8)	6.3 (1.2)	6.8 (1.7)	*0.002 (1.170)
		Fx_{MS}	28.7 (13.5)	28.2 (13.7)	31.6 (12.0)	28.4 (5.6)	0.631 (0.168)
		Fx_{PO}	66.9 (9.4)	77.4 (11.5)	75.1 (5.6)	75.7 (6.3)	0.141 (0.514)
Impulse ((N × s)/kg)	Z		57.3 (17.7)	59.8 (20.7)	52.2 (7.8)	52.7 (7.4)	0.131 (0.532)
	Y		13.5 (3.2)	15.2 (5.1)	12.0 (4.1)	11.1 (2.1)	*0.011 (0.924)
	X		4.4 (2.6)	5.1 (2.6)	4.6 (1.3)	4.5 (1.2)	0.808 (0.090)
Loading rate (N/kg/s)	Z		19.9 (12.6)	17.5 (8.3)	9.6 (3.1)	9.5(2.8)	< 0.001 (1.529)*
Free moment (%Bw × height)	Negative peak		0.007 (0.004)	0.007 (0.004)	0.004 (0.001)	0.005 (0.002)	*0.030 (0.777)
	Positive peak		0.006 (0.002)	0.006 (0.002)	0.006 (0.001)	0.005 (0,001)	0.341 (0.333)

Note: Fz_{HC}, First peak vertical ground reaction force at heel contact; Fz_{DF}, Vertical ground reaction force amplitude at the mid stance; Fz_{PO}, Second peak vertical ground reaction force at push-off; Fy_{HC}, Braking reaction force; Fy_{PO}, Propulsion force; Fx_{HC}, Peak medial ground reaction force at heel contact; Fx_{MS} and Fx_{PO} are two consecutive negative peaks at the mid stance and the push-off phase, respectively. TTP, Time to reach peak.

(Δ11%; p = 0.01; d = 0.93), posterior GRF during heel contact (Δ50%; p < 0.001; d = 1.63), anterior GRF during push off (Δ27%; p < 0.001; d = 1.34), and medial GRF during midstance (Δ25%; p = 0.032; d = 0.76) compared with healthy controls (Table 2 and Fig. 1).

In terms of time to peak of GRF, GV boys showed a shorter time interval for vertical GRF during heel contact (Δ26%; p < 0.001; d = 1.60), vertical GRF during mid stance (Δ6%; p = 0.039; d = 0.73), posterior GRF during heel contact (Δ35%; p < 0.001; d = 1.55), and lateral GRF during heel contact (Δ26%; p = 0.002; d = 1.17). The limb specific analysis indicated a longer time interval to reach peak Fx_{PO} in the dominant (Δ8%; p = 0.005; d = 1.02) compared with the non-dominant limb, irrespective of the experimental group.

Significant between group differences were found for anterior-posterior impulse, loading rate and negative peak free moment. The GV group experienced higher values in anterior-posterior impulse (Δ25%; p = 0.011; d = 0.92), loading rate (Δ96%; p < 0.001; d = 1.52), and negative peak free moment (Δ60%; p = 0.030; d = 0.77) compared with healthy controls (Table 2).

The GV group showed significantly higher activity for the vastus lateralis (Δ42%; p < 0.001; d = 1.82), vastus medialis (Δ32%; p = 0.013; d = 0.90), and for ensemble average EMG of the 3 muscles (Δ26%; p = 0.002; d = 1.16) compared with controls (Table 3).

Participants in the GV group versus healthy controls showed significantly larger peak knee flexion angles (Δ12%; p = 0.030; d = 0.77), peak knee adduction angles (Δ157%; p < 0.001; d = 1.63), and peak ankle eversion angles (Δ85%; p < 0.001; d = 2.06) compared with healthy controls (Table 4).

Between group differences were only apparent for the parameter RTD_100. The analysis revealed a 16% lower RTD_100 value in the GV group compared with healthy controls (p = 0.002; d = 1.17) (Table 5). There was a weak negative correlation between peak extensor knee torque and LR (r = -0.36, p = 0.033).

4. Discussion

This study examined the effects of varus alignment of the knee on kinematics and kinetics as well as lower limb muscle strength and activity in boys with GV compared with healthy age-matched peers. Our results confirmed our hypotheses and revealed varus-specific gait characteristics in the form of greater free moments, peak knee adduction and ankle eversion angles together with higher quadriceps activity.

In accordance with the study of Stief et al. [5], no significant differences were found in self-selected walking speed in GV boys compared with healthy controls. Our study revealed that boys with knee GV demonstrated a greater amplitude in peak vertical ground reaction force during heel contact (Fz_{HC}) and lower RTD in both limbs compared with healthy controls. Our finding that lower RTD of the knee extensors is associated with impulsive loading (greater Fz_{HC} amplitude) during walking is in accordance with previous literature [22,23]. These results can most likely be ascribed to weaker quadriceps femoris muscle function (lower RTD) in boys with GV which is reinforced by higher EMG activity (Table 4) [42]. Decreased force production of the tensor fascia lata muscle was observed in those with GV deformity [43]. Since the tensor fascia lata acts as a knee extensor and as a synergist of quadriceps muscles during the early stance phase of gait [44], this deficiency may put extra load on the quadriceps muscles. The quadriceps activity may, as a compensatory mechanism, absorb and dispense high internal joint forces in the medial knee compartment of individuals with GV malalignment [25].

A previous study evaluated the effects of simulated GV in healthy individuals on GRF and subtalar joint function during walking [45]. Van Gheluwe et al. [45] demonstrated a significant increase in the mean peak mediolateral GRF component during simulated GV walking. Our study revealed that GV deformity causes an increase in subtalar eversion through a more laterally directed GRF at heel contact. Therefore, our findings are consistent with previous study [45]. As a consequence, the GRF vector is more laterally directed. Greater eversion in boys with GV is likely a compensatory mechanism for the greater inclination of the tibia [8].

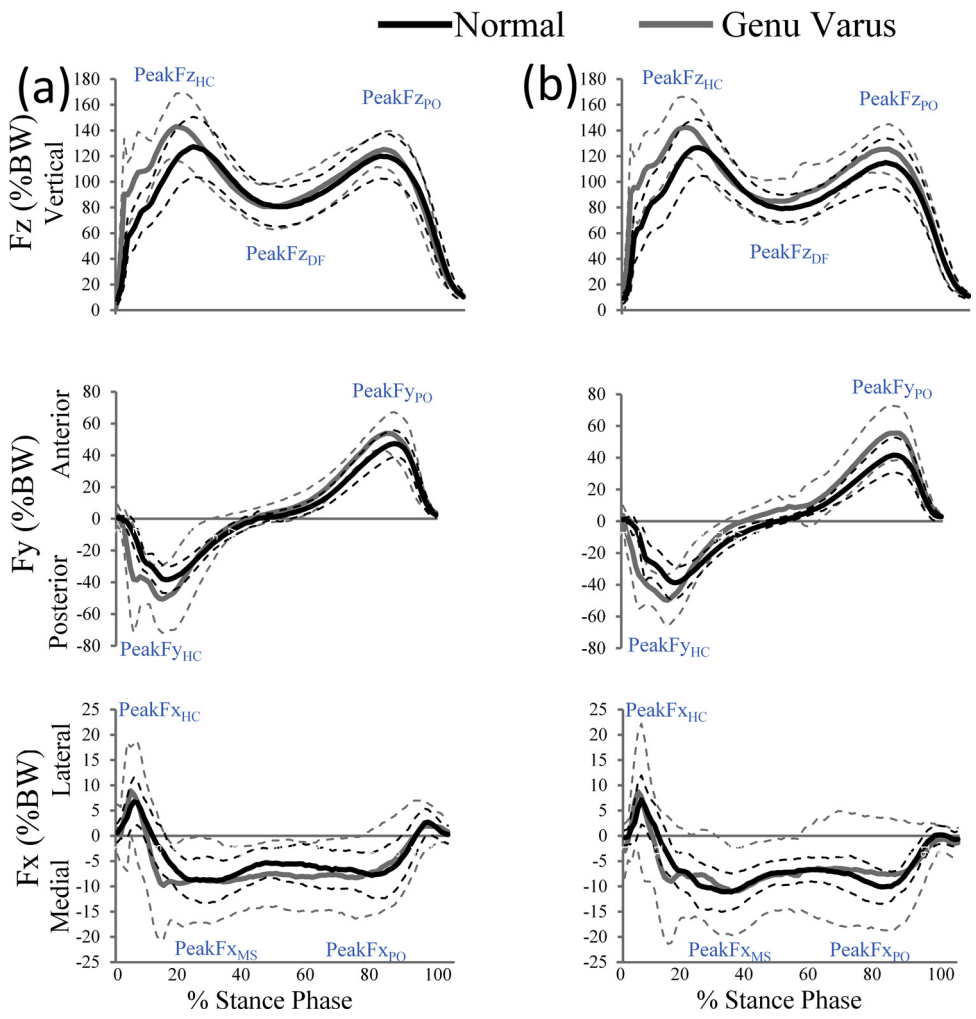

Fig. 1. Time-normalized traces of the ground reaction forces for both groups during the stance phase of gait cycle. Reaction forces in all axes for dominant (a) and non-dominant (b) lower limbs are displayed. The solid lines represent the mean values, whereas the dashed lines indicate standard deviation values.

Table 3
Group-specific means and standard deviations of EMG activity (% Maximum voluntary isometric contraction (MVIC)) of selected lower limb muscles.

Muscles	Genu varus		Healthy control		*p*-value (effect size Cohen's d)
	Dominant	Non-dominant	Dominant	Non-dominant	Group
m. rectus femoris	47.13 (20.76)	45.74 (17.89)	42.67 (17.63)	41.75 (16.37)	0.490 (0.238)
m. vastus lateralis	35.83 (5.61)	35.75 (5.57)	25.16 (6.47)	25.14 (6.39)	* < 0.001 (1.824)
m. vastus medialis	54.32 (15.01)	54.45 (14.77)	41.07 (15.95)	40.87 (15.55)	*0.013 (0.902)
Ensemble average	45.76 (10.47)	45.31 (9.73)	36.30 6.19)	35.92 (6.01)	*0.002 (1.164)

Table 4
Group-specific means and standard deviations of the examined kinematic parameters.

Joint angles	Genu varus		Healthy control		*p*-value (effect size Cohen's d)
	Dominant	Non-dominant	Dominant	Non-dominant	Group
Peak knee flexion angle (loading response phase)	17.78 (2.59)	17.75 (5.67)	15.87 (1.32)	15.73 (1.03)	*0.030 (0.777)
Peak knee adduction angle	5.43 (2.19)	5.43 (1.54)	2.42 (2.98)	1.81 (3.38)	* < 0.001 (1.633)
Peak ankle eversion angle	7.65 (1.63)	7.55 (1.75)	4.16 (1.78)	4.14 (1.75)	* < 0.001 (2.069)

In our study, boys with GV showed shorter times to reach peak GRF values than healthy age-matched controls. Shorter TTP values are associated with a higher rate of sustaining injuries such as stress fractures, articular soft tissue degeneration, osteoarthritis [14].

The observed kinematic changes in boys with GV in the present study are consistent with previous findings from Barrios et al. [8]. In fact, these authors revealed that adults aged 18–35 years with GV exhibited greater knee adduction, knee flexion, and ankle eversion than

Table 5
Group-specific means and standard deviations of the quadriceps strength indices.

Quadriceps Strength Indices	Genu varus		Healthy control		*p*-value (effect size Cohen's d)
	Dominant	Non-dominant	Dominant	Non-dominant	Group
Peak Torque (Nm/kg)	2.86 (0.45)	2.85 (0.42)	2.97 (0.51)	2.96 (0.53)	0.493 (0.238)
RTD (Nm/kg·s^{-1})	3.43 (1.38)	3.42 (1.20)	3.96 (1.23)	3.95 (1.22)	0.214 (0.434)
RTD_100 (Nm/kg·s^{-1})	7.44 (0.93)	7.53 (0.88)	8.63 (1.23)	8.67 (1.15)	*0.002 (1.173)

Note. RTD, rate of torque development;RTD_100, rate of torque development over the time interval of 100 ms after onset of torque.

healthy individuals [8].

In terms of loading rate during walking, results of this study confirm our hypothesis by showing that boys with GV demonstrated greater vertical loading rate than healthy age-matched controls. In contrast to our findings, Barrios et al. [8] observed no significant differences in loading rate between young adults with GV and healthy controls. This may be due to differences in the age range of participants in our study (9 to 14 years) compared with that of Barrios et al (18 to 35 years). Children compared with adults often adopt different neuromuscular coordination patterns when performing different types of motor tasks [46,47]. Immaturity of the neuromuscular system has been hypothesized as a potential reason for the differences observed in coordination patterns [46]. Besides maturity/immaturity, growth and thus rapid changes in anthropometrics may also play a significant role in neuromuscular coordination patterns when performing motor tasks [48]. Our results suggest that GV along with immaturity of the neuromuscular system during childhood [46] could also be responsible for the observed differences in quadriceps muscle activity and loading rate during walking in boys with GV compared with healthy controls. It has previously been demonstrated that lower RTD is associated with a greater vertical loading rate [23]. Our RTD-100 results confirmed the findings from Blackburn et al. [23]. Furthermore, as illustrated in Fig. 1, boys with GV demonstrated a heel strike transient phenomenon in their vertical GRF curve [49]. This phenomenon likely contributes to the high loading rate in boys with GV [49]. These data highlight the requirement of the quadriceps to generate knee extension torque rapidly following heel strike to allow appropriate impact force attenuation [23].

Higher loading rate during walking could be a risk factor for the development of knee OA [13]. This study revealed higher vertical loading rates due to lower limb malalignment in boys with GV compared with healthy controls. This could be associated with the long-term development of knee OA in individuals with GV. However, more longitudinal research is needed to verify cause and effect relations between higher vertical loading rates and the development of knee OA. A high loading rate contributes to a stiffening of the articular cartilage. This again may increase the risk of failure and injury [13]. It has been postulated that high loading rates during walking are caused by deficits in neuromuscular control, as well as muscle weakness [22]. There is evidence of muscle weakness in muscles encompassing the knee joint (e.g., quadriceps) in GV individuals [42]. However, it has to be noted that other studies were not able to show relationships between impact loading and quadriceps strength in individuals with knee OA [50,51]. Thus, other factors appear to play a role for impact loading during walking. A likely candidate, apart from muscle strength is neuromuscular control of the quadriceps [50]. From this it follows that future studies should explore the relationship between joint kinematics, muscle activation patterns, muscle co-activation synergies and loading while controlling for other covariates (e.g., neuromuscular control) in children with GV.

In this study, GV boys presented greater negative FM values in the dominant limb. The observed differences in negative FM during walking may be either a potential gait mechanism to compensate for excessive GV or may be necessary as a result of the altered MAA. Therefore, the increased internal knee rotation moments in GV boys may be attributed to modified gait patterns [4]. Consequently, it can be concluded that high FM amplitudes should be generally avoided because high FM amplitudes cause increased stress on biological tissues [52]. Therefore, orthotic devices should be designed to protect boys with GV from excessive loading of biological tissues during ground contact and to assist them to perform daily and sport activities. Apart from using passive stabilizers (i.e., orthotics), training (i.e., balance training) could be well suited as a means to treat lower limb malalignment in children with GV [53]. Granacher et al. were able to demonstrate that balance training not only improves balance performance but also RFD in adolescents [54]. Taken together, passive and/or active joint stability through orthotics and/or balance training appear to be well-suited means for GV treatment.

This study has some limitations that warrant discussion. First, participants were barefoot during walking trials and while this was done to generalize findings outside the laboratory, differences in shoes between individuals may somewhat distort results, as decreased impact loads are mentioned when walking barefoot compared to shoed walking [55]. Second, only boys and not girls were included in this study. Future studies should be conducted with girls to verify our findings.

5. Conclusions

When walking, boys with genu varus showed greater free moments, peak knee adduction and ankle eversion angles together with higher quadriceps activity compared with healthy age-matched controls. These findings illustrate typical gait characteristics of boys with genu varus that could be associated with the development of knee osteoarthritis. Accordingly, therapeutic interventions should be initiated at an early age to counteract long-term adverse effects of knee genu varus deformity.

Conflict of interest statement

None.

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