Inter-trial and test–retest reliability of kinematic and kinetic gait parameters among subjects with adolescent idiopathic scoliosis

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Abstract Gait analysis is actually used in subjects with scoliosis to determine the change in lower limb parameters after surgery, but the reliability of these parameters still remained unknown. The purpose of this study was to assess the repeatability of traditional gait parameters in subjects with adolescent idiopathic scoliosis (AIS) and to estimate the associated standard error of measurement (SEM). A test–retest design was used to assess the reliability of gait parameters at self-selected and fast speeds. A convenience sample of 20 girls aged from 12 to 17 years, with an idiopathic scoliosis (King classification: types I, II or III; Cobb angle 17–50°) participated in the study. Five good trials were recorded on two occasions. The time-distance, kinematic, and kinetic gait parameters were recorded using foot-switches in combination with a three-dimensional motion analysis system (Optotrak) and Advanced Mechanical Technologies Inc., (AMTI) Watertown, MA, USA; force plates. The coefficients of dependability and SEM derived from the generalizability theory were used to assess the

reliability. Inter-trial reliability was good for time-distance, kinematic, and kinetic (absolute and normalized) gait parameters except for the medio-lateral ground reaction force (GRF) component and the ankle dorsiflexor moment ($\phi = 0.60$ – 0.77). Test–retest reliability was higher for the kinetic than for the kinematic parameters. These coefficients ranged from 0.42 to 0.58 for the time-distance parameters; from 0.55 to 0.88 for the angular displacements and from 0.25 to 0.99 for the kinetic parameters. The SEMs were lower than 3.3° for the angular displacements and lower than 8 Nm (0.15 Nm/kg) and 36 W (0.54 W/Kg) for the joint moments and powers regardless of the speed. Several gait parameters are reliable among subjects with AIS and can be used to assess the evolution of the spinal modifications and the impact of treatment on their lower limb gait pattern.

Keywords Gait analysis · Adolescent idiopathic scoliosis · Reliability · Kinematics · Kinetics · Generalizability theory

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Introduction

Idiopathic scoliosis is a three-dimensional deformity of the spine occurring during the growth period [15]. About 10% of subjects with adolescent idiopathic scoliosis (AIS) require some form of treatment such as physiotherapy, a brace or surgery [32]. The pathogenesis of AIS still remains unclear but some studies have demonstrated that poor postural and balance control during walking might play an important role in the progression of scoliosis [3, 20].

It is now accepted that gait analysis can provide additional information about the impact of postural modifications and of vertebral surgery on functional

outcome among patients with scoliosis [12, 13, 19]. Several authors used gait analysis to document changes in gait parameters among adolescents and adults with idiopathic scoliosis before and after surgery [12, 13, 19]. However, to determine if change in gait pattern over time is attributable to real change, one must establish the natural variability of gait parameters from one trial to the other (inter-trial reliability) and how consistently the gait pattern is from one occasion to the other (test–retest reliability) [9].

Reliability studies of gait parameters for normal children have been reported [28, 33] as well as for normal young adults [17, 22, 36]. The time-distance parameters such as step length, cadence, and walking velocity are reproducible [17, 28] with the lowest reliability for the gait velocity parameter. The reliability of kinematic parameters is highest in the sagittal plane at the hip, knee, and ankle and highest in the frontal plane at the pelvis in normal children and normal adults [17, 28]. Movements showing the lower range of motion seemed to have less repeatability [28]. The kinetic parameter studies showed good inter-trial and test–retest reliability of the vertical, anterior-posterior and medio-lateral ground reaction force (GRF) components [17, 28] with lowest levels for the medio-lateral maximum peaks in normal children [33] and in normal girls [14]. Only few researchers have studied the repeatability of joint net moments during gait. According to Kadaba et al. [17] and Steinwender et al. [28], moment curve profiles are reproducible in normal and spastic children as well as in normal adults with better reliability observed in the sagittal plane.

Reliability studies of gait parameters are scarcely reported in children and youth with pathological conditions. Steinwender et al. [28] reported a lower level of reliability in spastic children compared to normal children for lower limb parameters at self-selected gait speed. Giakas et al. [14] showed higher intra and inter-subjects variability of the GRF in the frequency domain among subjects with AIS compared to the control group. Chan et al. [5] reported inconsistency of the within-day repeatability in trunk, pelvic, and shoulder motion during walking in subjects with AIS.

Although gait analysis has been suggested to assess function in subjects with AIS [6, 12, 19, 31], before making any inference about scoliosis and modifications of the gait parameters, a clearer picture of the variability of the gait patterns in AIS must be provided. The purpose of this study was thus to assess the reliability of the time-distance, kinematic, and kinetic parameters of gait in AIS and to estimate the standard error of measurement (SEM) associated with these parameters. A test–retest design with multiple trials at self-selected and fast speed was used to establish the inter-trial and test–retest reliability. The generalizability theory served as the framework to determine the sources of variance (generalizability study-G-study), the level of reliability

and errors expected for particular designs (decision study-D-study) [27].

Materials and methods

Participants

A convenience sample of 20 teenage girls aged from 12 to 17 with an idiopathic scoliosis of types I, II or III of King's classification and a frontal deformity between 17 and 50° (Cobb angle) participated in the study (Table 1). The diagnosis and classification of the spinal deformity was determined by an orthopedic surgeon (HL). Subjects having any other musculo-skeletal abnormalities or previous spine or lower limb surgery were excluded. All teenagers were recruited from the scoliosis clinics of two children's hospitals in Montreal (Sainte-Justine Hospital and Montreal Children's Hospital). Ethical approvals were obtained from the Research Ethics Committees. All subjects and their parents gave written consent before participating in this study. The gait evaluations were carried out at the pathokinesiology laboratory located at the Institut de réadaptation de Montréal (IRM).

Study design

A test–retest design was used to assess the reliability of the gait parameters. Each subject was evaluated on two occasions, 5–10 days apart at about the same time of the day. Five good trials, at self-selected and fast speed, were recorded for each participant. The same protocol was repeated at the second visit and the two evaluations were conducted by the same evaluator. The girls practiced well before the data were collected, until they felt that they performed the task naturally. They were asked to walk along the walkway at their own self-selected speed, the speed at which they felt most comfortable and then at their fast speed, the speed they would adopt if they were trying to catch up with someone without running. All girls wore their own running shoes/trainers.

Procedure and Instrumentation

Anatomical landmarks were identified before each evaluation and some reference points were noted to ensure better reliability of the location of anatomical landmarks in the second evaluation. The gait assessments included simultaneous recordings of time-distance, kinematic, and kinetic data. The time-distance parameters were recorded from three foot-switches located under each foot at the heel,

Table 1 Characteristics of the 20 adolescent girls with idiopathic scoliosis

Subjects	Age (years)	Height (cm)	Mass (kg)	Scoliosis		
				Cobb angle (°)		King types
				Thoracic	Lumbar	
1	12	160.0	46.7	21	23	I
2	12	150.5	43.8	31	24	II
3	12	152.0	36.8	17	25	II
4	12	157.8	59.4	27		III
5	14	174.0	60.0	43	38	I
6	14	163.0	46.3	23	16	II
7	14	161.0	86.1	37	27	II
8	14	169.0	71.0	50	39	II
9	14	164.0	44.0	33		III
10	15	166.0	55.8	38	38	I
11	15	159.3	52.5	17	20	I
12	15	163.0	40.0	50	37	II
13	16	161.5	45.3	46	42	I
14	16	175.8	73.0	46	46	I
15	16	159.0	46.0	40	27	II
16	16	155.5	49.0	46		III
17	17	161.5	50.5	28	31	I
18	17	166.0	49.5	41	32	II
19	17	158.0	49.4	33	25	II
20	17	166.0	53.2	33		III
Mean	14.8	162.2	52.9	35.0	30.6	Type I $n = 7$
(SD)	(1.77)	(6.35)	(12.0)	(10.5)	(8.6)	Type II $n = 9$
						Type III $n = 4$

metatarsal head, and first toe. The kinematic data were obtained using an Optotrak system (Northern Digital Inc., Waterloo, ON, Canada) sampled at 60 Hz. The three-dimensional coordinates of three noncollinear active infrared markers, placed on the feet (lateral heel, dorsum, and fifth metatarsal head), legs (lateral malleolus, mid-shank, and fibula head), thighs (greater trochanter, mid-thigh, and lateral femoral condyle), pelvis (left and right posterior superior iliac spines, left iliac crest), and trunk (C7, T8, and T12 spinous process and right and left at the level of T8) were acquired during gait. In addition, 12 specific anatomical points on the feet (heel, posterior point, and mid-toe anterior point), shanks (medial malleolus), thighs (medial femoral condyle), pelvis (right and left ASIS, right iliac crest), and on the trunk (left glenohumeral joint) were digitized with a probe to define precisely the segments and anatomical axes. The kinematic data were filtered with a fourth-order Butterworth zero-lag filter, with a cutoff frequency of 6 Hz. Using the analysis package from Mishac Inc. (Mishac Kinetics, Waterloo, ON, Canada), the relative angles were calculated from a rotation matrix using a Cardanic (x - y - z rotation) sequence of rotation such that the local x , y , and z axes corresponded,

respectively, to abduction–adduction, longitudinal rotation, and flexion–extension for the hip and knee joints, and eversion–inversion, transverse rotation, and dorsiflexion–plantarflexion for the ankle joint.

The GRF were collected at 600 Hz with three force platforms (AMTI Advanced Mechanical Technologies Inc., Watertown, MA, USA, Model OR6-7-1000) embedded in the 10-m walkway. The data were later filtered with a fourth-order Butterworth zero-lag filter, with a cutoff frequency of 10 Hz and resampled at 60 Hz to match the kinematic data. An inverse dynamic approach [4, 35] performed with Kingait3 software (Mishac Kinetics) was used to estimate the net moments at the ankle, knee, and hip joints. The net power at the ankle, knee, and hip joints was then computed by multiplying the joint angular velocity by the local net joint moment in the same plane of movement. Three-dimensional data from the right and left lower limbs were collected but the reliability of the gait parameters was assessed only for the frontal and sagittal planes of movement at the right side.

The gait cycle began upon initial contact of the right foot with the force plate and ended when the same foot contacted the ground. Five trials at comfortable and

maximal gait speed were recorded for each participant. Ensemble averages were calculated for the time-distance parameters and at each percent of the gait cycle for the angular displacements, the GRFs, the net moments as well as the powers. Specific maximum and minimum peak values were selected on the kinematic and kinetic curve profiles to assess the reliability of the gait parameters [33]. In the sagittal plane, the reliability was assessed for almost all the peaks usually reported in gait studies [1, 11, 34]. As for the sagittal plane, the reliability was assessed for known peaks in the frontal plane [1, 11]. A detailed nomenclature of these parameters is presented in Table 2.

Data analysis

Descriptive statistics [mean, standard deviation—(SD)] have been used to characterize subjects with scoliosis and the time distance parameters. The mean ($\pm$ SD) kinematic and kinetic (normalized) time-series profiles in the sagittal and frontal planes of movement are presented in Figs. 1 and 2 with the minimum and maximum peak parameters used in the reliability analyses.

The generalizability theory was used to evaluate the reliability of the gait parameters [27]. This theory can be considered as an extension of the intra-class coefficient (ICC) [16]. Based on the analysis of variance, the generalizability theory is divided into two parts, the first being the G-study and the second, the D-study. The G-study allows the magnitude of the variances attributed to the subject (S), to the systematic errors related to occasion (O), and trial (T), and to random errors associated with the interactions between occasion and trial (OT) and subject and occasion (SO) to be determined. The residual error is the interaction between all sources of error and, in this study, corresponds to the subject, occasion and trial variance (SOT) involved. Thus, in this study, six sources of error variance, S, O, T, OT, SO, and SOT, can be identified.

The D-study uses the information of the G-study to determine the reliability of a particular protocol. The G- and D-studies allow the dependability coefficient (ϕ) and generalizability coefficient to be calculated. The dependability coefficient is the ratio between the inter-subject variance (σ_s^2) and the sum of the inter-subject variance and all possible sources of error ($\sigma_s^2 + \sigma_{abs}^2$). The sum of the error variances (systematic and random) is called the absolute error variance (σ_{abs}^2). See Table 3 for formula details.

Like the ICC, the dependability coefficient ranges between 0 and 1: 0 is null reliability and 1, perfect reliability. Portney and Watkins [24] have suggested that values above 0.75 can be considered as good reliability,

those between 0.50 and 0.75 as moderate and those under 0.5 as poor. The formulas for the calculation of inter-trial and test–retest coefficients are given in Table 3.

To appreciate the errors in terms of the unit of measurement, the SEM, which is the root square of the absolute error variance, was computed [27]. In the present study, the SEMs were described according to the unit of each gait parameter. The generalizability analysis was done with the GENOVA program [8].

Results

The 20 girls with AIS had an average self-selected walking speed (mean $\pm$ SD) of 1.29 ± 0.16 m/s in their first assessment at the laboratory. Their stride length and cadence were 1.41 ± 0.10 m and 105 ± 7.8 steps/min, respectively. When asked to walk at fast speed, as a group, the girls' gait velocity increased to 1.82 ± 0.17 m/s and their stride length and cadence reached 1.67 ± 0.12 m and 125 ± 8.6 steps/min, respectively. By their second visit, the group's mean self-selected walking speed had increased by 9.4% (1.41 m/s vs. 1.29 m/s). This higher gait speed at the second visit was associated with an increase in both stride length (5%: 1.48 m vs. 1.41 m) and cadence (4%: 111 steps/min vs. 105 steps/min).

Reliability

Source of variance

For all gait parameters, the inter-subject variance was the major source of variance in the data (Table 4) followed by the interactions involving subjects with occasion (SO) and subjects with occasion and trial (SOT). For all parameters, the variance components associated with the “trial” (T) were very low, namely under 5%. At self-selected speed, a systematic effect associated with “occasion” (O) was observed for cadence (17%), speed and stride length (21%), as well as for absolute and normalized moment parameters ($\leq 16\%$) and normalized power parameters ($\leq 9\%$). There was no clear effect of the plane of movement on the variance components.

On average, for the angular displacements, the ankle showed the lowest inter-subject variance followed by the hip and the knee (not reported in Table 4). Among the kinetic parameters, the vertical GRF components showed the highest percentage of inter-subject variance (88–95%).

Normalization of the kinetic parameters (GRF, moments, and powers) to body mass dramatically decreased the percentage of inter-subject variance, about 30% for the vertical and antero-posterior GRF components.

Table 2 Nomenclature of the peak parameters used for the reliability studies

GRF: vertical component	GRF: antero-posterior component	GRF: medio-lateral component
FY1: first maximum peak	FX1: braking force; negative peak	FZ1: maximal peak at initial stance
FY2: minimum peak between first and second maximum peaks	FX2: propulsive force; positive peak	FZ2: first minimal peak
FY3: second maximum peak		FZ3: depression between first and second minimal peaks
		FZ4: second minimal peak
Angular displacements-sagittal		
Angular displacements-frontal		
Hip	Hip	
Ext: maximal extension in stance	ADD: maximal adduction at initial stance	
	ABD: maximal abduction in swing	
Knee		
FL1: maximal flexion in stance		
Ext: maximal extension in stance		
FL2: maximal flexion in swing		
Ankle	Ankle	
PF1: first plantar flexion at initial stance	ADD1: maximal adduction in stance	
DF: maximal dorsiflexion in stance	ABD: maximal abduction at transition from stance to swing	
PF2: maximal plantar flexion at push-off	ADD2: maximal adduction in swing	
Net joint moments-sagittal		
Net joint moments-frontal		
Hip	Hip	
Ext1: maximal extension moment at initial stance	ABD1: first maximal abduction moment	
FL1: maximal flexion moment at end of stance	ABD2: minimal abduction moment between first and second abduction moment	
Ext2: maximal extension moment at end of swing	ABD3: second maximal abduction moment	
	ADD: maximal adduction moment at the transition from stance to swing	
Knee	Knee	
Ext1: first maximal extension moment	ABD1: first maximal abduction moment	
FL1: first maximal flexion moment	ABD2: minimal abduction moment between first and second abduction moment	
Ext2: second maximal extension moment at transition from stance to swing	ABD3: second maximal abduction moment	
FL2: second maximal flexion moment at end of swing phase	ADD: maximal adduction moment at the transition from stance to swing	
Ankle		
DF: maximal moment in dorsiflexion at initial stance		
PF: maximal moment in plantar flexion at push-off		

Table 2 continued

Net joint powers-sagittal	Net joint powers-frontal
Hip	Hip
H1-S: maximal positive power by extensors at initial stance	H1-F: maximal negative power by abductors at initial stance
H2-S: maximal negative power by flexors at end of stance	H2-F: first positive power by abductors in stance
H3-S: maximal positive power by flexors at swing	H3-F: second positive power by abductors at the end of the stance phase
Knee	
K1-S: maximal negative power of extensor muscles at the beginning of stance	
K2-S: maximal positive power by knee extensors	
K3-S: maximal negative power of knee extensors at transition from stance to swing	
K4-S: maximal negative power by the knee flexors at end of swing	
Ankle	
A1: maximal negative power of plantar flexors in stance	
A2: maximal positive power of plantar flexors at push-off	

Inter-trial reliability

The inter-trial reliability of the kinematic gait parameters was good at self-selected speed and fast speed. The coefficients of dependability (ϕ) ranged from 0.94 to 0.97 for the time-distance parameters and from 0.85 to 0.98 for the angular displacements. The SEMs for stride length (SS and fast 0.02 m) and velocity (SS 0.03 m/s, fast 0.04 m/s) were low.

For the angular displacements, the inter-trial reliability in the sagittal plane (ϕ ranging from 0.88 to 0.98 for five trials) was comparable to that in the frontal plane (ϕ ranging from 0.85 to 0.98 for five trials) and the highest coefficients were observed at the hip, followed by the knee and the ankle. The SEMs of the angular displacements were lower than 1.7° (ranging from 0.7 to 1.6°) at SS speed and lower than 3.4° (ranging from 0.8 to 3.3°) at fast speed with highest value at the knee for flexion during the swing phase (flexion 2).

The inter-trial repeatability of the GRF at self-selected speed was high for the vertical ($\phi = 0.98$ –0.99) and antero-posterior ($\phi = 0.94$ and 0.97) components and slightly lower for the medio-lateral forces ($\phi = 0.79$ –0.88). The medio-lateral forces were more affected by gait speed (SS $\phi = 0.79$ –0.88, fast $\phi = 0.60$ –0.77) than the vertical and antero-posterior GRF components (lowest values of $\phi = 0.91$). Overall, the SEMs ranged from 3 to 16 N at self-selected speed and from 5 to 21 N at fast speed.

The inter-trial reliability for the moment and power parameters was also good at self-selected speed ($\phi = 0.93$ –0.99 and $\phi = 0.89$ –0.96, respectively) and the SEMs were lower than 3.2 Nm and 8.9 W. Walking at fast speed drastically decreased the inter-trial reliability of the ankle moment in dorsiflexion (SS $\phi = 0.92$ vs. fast $\phi = 0.60$) but did not affect other moment and power parameters. The SEMs of the moment and power parameters were lower at self-selected speed (moment <3.2 Nm, power <8.9 W) than at fast speed (moment <5.7 Nm, power <18.4 W).

In general, normalization to body mass reduced the inter-trial reliability of the kinetic parameters but the level remained good except for the medio-lateral force ($\phi = 0.63$ for normalized and 0.76 for absolute value).

Test–retest reliability

For the time-distance parameters at self-selected speed, the dependability coefficients revealed poor reliability (highest value of 0.58; Table 5). Walking at fast speed increased the reliability from moderate to good level and the SEMs showed a slight decrease (see details in Table 5). For the angular displacements, the dependability coefficients were poor to good at self-selected speed. In the sagittal plane, the knee showed the highest reliability ($\phi = 0.86$), followed by

Fig. 1 GRF components and angular displacement curve profiles (mean $\pm$ 1SD). **a** GRF components; **b** angular displacement curve profiles in sagittal plane; **c** angular displacement curve profiles in frontal plane

Fig. 2 Moments of force and power curve profiles (mean $\pm$ 1SD). **a** Moments of force curve profiles in sagittal and frontal plane; **b** power curve profiles in sagittal and frontal plane

the ankle and hip. In the frontal plane, only the maximal hip adduction angle at initial stance (ADD) showed a good reliability. The SEM-values were lower than 3.2° (see details in Table 5).

The test-retest reliability of the absolute GRF parameters at self-selected gait speed was moderate to good (Table 6). The coefficients of dependability were higher for the vertical component ($\phi = 0.92\text{--}0.99$) followed by the

Table 3 Equations for the calculation of reliability coefficientsDependability coefficient (ϕ) for random design

$$\sigma_S^2/\sigma_S^2 + \sigma_{abs}^2$$

Error absolute in random design (σ_{abs}^2)

$$\sigma_T^2/n_T + \sigma_O^2/n_O + \sigma_{ST}^2/n_T + \sigma_{SO}^2/n_O + \sigma_{OT}^2/n_On_T + \sigma_{SOT,e}^2/n_On_T$$

Test–retest coefficient

$$\frac{\sigma_S^2 + \sigma_{ST}^2/n_T}{(\sigma_S^2 + \sigma_{ST}^2/n_T) + \sigma_{abs}^{2of}}$$

Error absolute in mixed design with “trial” fixed (σ_{abs}^{2of})

$$\sigma_O^2/n_O + \sigma_{SO}^2/n_O + \sigma_{TO}^2/n_Tn_O + \sigma_{SOT,e}^2/n_Tn_O$$

Inter-trial coefficient

$$\frac{\sigma_S^2 + \sigma_{SO}^2/n_O}{(\sigma_S^2 + \sigma_{SO}^2/n_O) + \sigma_{abs}^{2of}}$$

Error absolute in mixed design with “occasion” fixed (σ_{abs}^{2of})

$$\sigma_T^2/n_T + \sigma_{ST}^2/n_T + \sigma_{TO}^2/n_Tn_O + \sigma_{SOT,e}^2/n_Tn_O$$

In this study all the coefficients were computed with $n_T = 5$ and $n_O = 1$

antero-posterior component ($\phi = 0.81$ and 0.82) and the medio-lateral component ($\phi = 0.72$ – 0.85). Walking at fast speed slightly decreased the reliability of the medio-lateral peak parameters (FZ2, FZ3, and FZ4). Overall, the SEMs were lower than 29, 13, and 6 N for the vertical, antero-posterior, and medio-lateral GRFs, respectively.

The test–retest reliability of the absolute net moments of force and power parameters was poor to good at self-selected gait speed (Table 6). The highest dependability coefficients were observed for the ankle plantar flexion moment in the sagittal plane ($\phi = 0.97$) and for the H1-F power parameter in the frontal plane ($\phi = 0.90$). The dorsiflexor moment, the first peak of hip extension moment

(Ext1) and the H1-S, H2-S, K1-S power parameters were the least-reliable parameters at self-selected speed. Walking at fast speed affected the hip extension moment (Ext1 $\phi = 0.55$ at SS vs. $\phi = 0.83$ at fast speed) and the reliability level of the K1-S and K2-S changed from poor to good. The SEMs of the net moment parameters ranged from 1.5 to 7.8 Nm in the sagittal plane and from 1.1 to 4.4 Nm in the frontal plane. The SEMs for the power parameters varied from 7.6 to 35.8 W in the sagittal plane and from 4.0 to 8.8 W in the frontal plane.

For several parameters, normalization of the kinetic data to body mass drastically lowered the level of reliability of the kinetic gait variables, even though some parameters still showed a good test–retest reliability (see Table 6).

Discussion

This study assessed the inter-trial and test–retest reliability of the gait parameters among subjects with AIS walking at self-selected and fast speeds. The most relevant results of the inter-trial reliability are discussed first followed by those of the test–retest reliability.

Inter-trial reliability

As previously reported by Giakas et al. [14], the inter-trial reliability was found to be lower for the medio-lateral GRF component at both speeds but more noticeable at fast speed. This could be associated with a more variable lateral

Table 4 Ranges of the percentage of the total variance components (%) for each facet and their interactions obtained from the G-study for the time-distance, kinematic and absolute kinetic gait parameters at self-selected and fast speed

Sources of variance	Gait parameters				
	Kinematic		Kinetic		
	Time-distance	Angle	GRF absolute	Moment absolute	Power absolute
Self-selected					
Subject (S)	39–71	39–78	46–95	46–94	23–83
Trial (T)	< 5	< 5	< 5	< 5	< 5
Occasion (O)	4–21	< 5	< 5	0–15	0–11
SO	13–30	5–42	0–12	2–26	4–57
ST	< 5	< 5	0–11	< 5	< 5
OT	< 5	< 5	< 5	< 5	< 5
SOT	8–12	4–30	2–45	4–23	9–24
Fast speed					
Subject (S)	60–84	32–70	32–92	19–91	30–86
Trial (T)	< 5	< 5	< 5	< 5	< 5
Occasion (O)	< 5	0–7	< 5	< 5	< 5
SO	6–15	2–34	< 5	3–15	2–48
ST	< 5	< 5	0–8	< 5	< 5
OT	< 5	< 5	< 5	< 5	< 5
SOT	10–22	7–62	5–65	5–67	4–34

Variances: SO subject-occasion, ST Subject-trial, OT occasion-trial, SOT subject-occasion-trial

Table 5 Test–retest reliability: dependability coefficients (ϕ) and standard error of measurement (SEM) for the time-distance and angular displacement parameters at self-selected (SS) and fast speed for five trials

Kinematics	Parameters		Self-selected speed		Fast speed		
			ϕ	SEM	ϕ	SEM	
Time-distance	Velocity (m/s)		0.53	0.12	0.86	0.07	
	Cadence (steps/min)		0.58	5.5	0.74	4.7	
	Stride length (m)		0.42	0.08	0.92	0.03	
Angular displacements (degrees)	HIP: sagittal	Ext	0.70 ^b	2.8	0.66	3.2	
		HIP: frontal	ADD	0.80 ^a	1.9	0.77	2.4
			ABD	0.54	2.4	0.61	2.2
	KNEE: sagittal	FL1	0.60	3.2	0.87	1.5	
		EXT	0.88	1.4	0.80	2.0	
		FLE2	0.81	2.1	0.67	2.8	
	ANKLE: sagittal	PF1	0.72	1.4	0.57	3.2	
		DF	0.74	1.8	0.71	2.2	
		PF2	0.67	2.7	0.71	2.4	
	ANKLE: frontal	ADD1	0.55	2.1	0.47	2.4	
		ABD	0.69	2.5	0.78	2.0	
		ADD2	0.77	2.1	0.72	2.4	

^a Values higher than 0.75, can be considered to represent good reliability

^b Values between 0.70 and 0.75, represent a close to good reliability

foot placement on the ground from one trial to the other [25]. According to Giakas et al. [14], the lateral deformity of the spine might affect the balance of the foot during the stance phase of gait and create higher variability in the medio-lateral GRF component from one trial to the other. This could also explain the lower inter-trial reliability of the medio-lateral GRFs at maximal speed because, as the walking speed increases, the discrepancy of the gait patterns can be amplified, possibly creating more variability [18].

Test–retest reliability

We found that the time-distance parameters, velocity, cadence, and stride length were not reliable at self-selected walking speed but were reliable at fast walking speed. The unreliability of the time-distance parameters originated from a systematic error associated with occasions, revealing a higher self-selected velocity at the second visit for almost all AIS subjects. This systematic error was not seen at fast walking speed, which increases the reliability for this condition. The generalizability theory indicates that a better level of test–retest reliability ($\phi = 0.68$) of the time-distance parameters at self-selected speed would be achieved using a protocol based on three trials repeated on three visits. If only one session is allocated, one alternative would be to assess AIS patients at fast speed of gait in addition to self-selected speed.

According to several authors [17, 21, 26, 28], angular-displacement test–retest reliability is influenced by the subjects, who can modify their gait pattern, the measurement system, differences in marker position and errors due

to soft tissue movement. In the present study, efforts were made to minimize the magnitude of these errors. First, a highly accurate system (Optotrak system) was used to track the marker positions placed on the subjects. This three-dimensional movement analysis system has a root mean square accuracy of 0.1 mm and a resolution of 0.01 mm. Second, the evaluator took measurements (e.g., distance between the ground and the marker placed on the great trochanter) in order to optimize the identical replacement of the markers on the subjects at the second visit. Finally, the same well-trained physical therapist performed all the marker identifications and placement. We also allowed the girls to practice until they felt that they performed the task naturally and we gave them the same instructions at each visit.

In this study, the reliability of angular-displacement parameters at self-selected speed was good in the sagittal plane for the knee (except for FL1) and moderate for the ankle and hip. This is in agreement with Maynard et al.'s findings [22]. These authors associate a better level of reliability at this joint with an easier identification of anatomical landmarks for the placement of markers on the knee. However, the SEM values for the ankle and hip were similar to those obtained for the knee (ranging from 1.4° to 2.8°). The hip angles were more reliable in the frontal (adduction movement) than in the sagittal plane. This could be explained by the higher inter-subject variance observed in the frontal plane than in the sagittal one.

Fast walking speed reduced the level of reliability of several parameters and increased the SEM-values, except for the first knee flexion (FL1). As already reported by Winter [34], knee flexion during the early stance phase was one joint angle that showed the greatest changes with an

Table 6 Test–retest reliability: dependability coefficients (ϕ) and standard error of measurement (*SEMs*) for the kinetic parameters at self-selected and fast speed

Kinetic parameters		Self-selected speed				Fast speed			
		Absolute values		Normalized values		Absolute values		Normalized values	
		ϕ	SEM	ϕ	SEM	ϕ	SEM	ϕ	SEM
GRF (N, N/kg)									
Vertical	FY1	0.95 ^a	28.3	0.49	0.56	0.97	22.3	0.84	0.45
	FY2	0.92	28.8	0.72	0.56	0.95	21.3	0.88	0.44
	FY3	0.99	14.8	0.80	0.29	0.98	19.7	0.78	0.42
Antero-post.	FX1	0.81	12.6	0.61	0.26	0.92	9.6	0.87	0.18
	FX2	0.82	12.1	0.51	0.23	0.93	9.1	0.72	0.17
Medio-lat.	FZ1	0.72 ^b	4.0	0.69	0.08	0.78	5.3	0.72	0.10
	FZ2	0.85	3.8	0.77	0.08	0.79	5.9	0.73	0.12
	FZ3	0.76	3.3	0.54	0.06	0.66	3.9	0.29	0.08
	FZ4	0.77	4.3	0.48	0.08	0.71	4.3	0.51	0.09
Moment (Nm, Nm/kg)									
HIP: sagittal	Ext1	0.55	7.8	0.53	0.14	0.83	7.1	0.81	0.13
	FL1	0.80	6.9	0.59	0.12	0.86	7.0	0.78	0.12
	Ext2	0.83	3.0	0.62	0.06	0.80	4.8	0.54	0.10
HIP: frontal	ABD1	0.95	3.5	0.72	0.07	0.95	3.6	0.75	0.07
	ABD2	0.93	3.5	0.78	0.06	0.96	2.3	0.87	0.04
	ABD3	0.91	4.4	0.57	0.08	0.88	5.5	0.66	0.10
	ADD	0.78	1.9	0.72	0.03	0.78	2.9	0.73	0.05
KNEE: sagittal	Ext1	0.79	6.7	0.61	0.12	0.91	5.2	0.83	0.09
	FL1	0.73	3.7	0.79	0.06	0.72	4.1	0.78	0.08
	Ext2	0.89	2.1	0.81	0.03	0.84	2.9	0.71	0.05
	FL2	0.84	1.5	0.54	0.03	0.88	1.6	0.67	0.03
KNEE: frontal	ABD1	0.82	3.0	0.79	0.06	0.83	3.2	0.84	0.06
	ABD2	0.86	1.5	0.75	0.03	0.87	1.4	0.80	0.02
	ABD3	0.89	1.4	0.87	0.02	0.86	1.8	0.82	0.03
	ADD	0.67	1.1	0.59	0.02	0.64	1.8	0.47	0.03
ANKLE: Sagittal	DF	0.53	2.6	0.57	0.05	0.40	3.2	0.53	0.04
	PF	0.97	2.7	0.85	0.06	0.96	3.3	0.85	0.06
Power (W, W/k kg)									
HIP: sagittal	H1-S	0.25	18.7	0.34	0.38	0.50	35.8	0.58	0.53
	H2-S	0.52	12.0	0.42	0.20	0.69	15.5	0.72	0.27
	H3-S	0.87	7.8	0.80	0.16	0.81	18.1	0.80	0.35
HIP: frontal	H1-F	0.90	8.8	0.83	0.15	0.95	8.4	0.88	0.17
	H2-F	0.84	4.0	0.78	0.08	0.82	5.7	0.77	0.11
	H3-F	0.83	4.3	0.71	0.09	0.76	6.7	0.68	0.14
KNEE: sagittal	K1-S	0.54	22.1	0.53	0.39	0.82	20.4	0.83	0.33
	K2-S	0.70	10.3	0.58	0.18	0.85	10.7	0.74	0.19
	K3-S	0.83	10.9	0.75	0.19	0.75	16.4	0.68	0.30
	K4-S	0.78	7.6	0.71	0.14	0.78	11.8	0.70	0.23
ANKLE: sagittal	A1	0.81	12.7	0.71	0.24	0.78	18.6	0.67	0.34
	A2	0.88	18.6	0.74	0.39	0.92	16.9	0.85	0.29

^a Values higher than 0.75, can be considered to represent good reliability^b Values between 0.70 and 0.75, represent a close to good reliability

increase in gait speed. Our data revealed that the variance associated with the subject factor was higher for the first knee flexion than for the other angular displacement parameters at fast speed.

The results of the D-study indicated a good test–retest reliability for several absolute kinetic parameters. For the GRFs, in agreement with previous studies [17, 28, 33, 34], we found a higher reliability for the absolute vertical and antero-posterior GRF components than for the medio-lateral component at self-selected and fast walking speeds. Kadaba et al. [17] stated that most of the variability for the GRF was attributable to the subject's physiological factors because the source of error of the force-plate measurement system was very small from one session to the other. In the present study, for the medio-lateral component, there was a considerable source of residual error, especially at fast speed, which can explain the lower reliability for this parameter. This source of residual error will need further examination to improve the reliability of the GRF parameter.

A good level of reliability was also observed at self-selected speed, in the sagittal plane, for absolute net joint moments and powers with the highest coefficient obtained for plantar flexion moment at the ankle and lower SEM-values at this joint. This agreed with the results of previous studies [17, 28, 36]. This fact might be related to the joint center, which was easier to estimate at the ankle than at the knee and hip from one session to the other [17]. Furthermore, during the stance phase, the hip and knee moments may be more variable than the ankle moment due to biarticular muscles crossing these joints that can change the motor pattern between the hip and knee joints [17, 36]. In the present study, the ankle moment in dorsiflexion and hip moment in extension at early stance were not reliable and corroborate with Maynard et al.'s findings [22]. Those moment parameters showed the lowest subject variance (46 and 49%) and had a systematic error associated with occasion of 15 and 14%, respectively. It is possible that the systematic error is due to a difference of 9.4% in the self-selected walking speed from Session 1 to 2.

The reliability of hip joint moments and powers was higher in the frontal plane than in the sagittal plane. Other studies have reported that reliability at the hip in the frontal plane was similar to or lower than that in the sagittal plane among normal children and normal adults [17, 28]. This can be explained by physiological factors [17] and by uncertainties in the calculation of joint moments in the frontal plane at the hip [10, 17]. Furthermore, Giakas et al. [14] previously reported higher inter-subject variability among subjects with AIS for the medio-lateral GRF component, which is used in the calculation of joint moments and thereafter for power parameters.

Overall, walking at fast speed did not affect to a great extent the level of reliability of joint moments and powers.

However, it increased the inter-subject variance of the extension moment (Ext1) at the hip and at the knee powers (K1 and K2) resulting in highest coefficients of dependability. According to Winter [34], the magnitude of extension moment at the hip (Ext1) is associated with the trunk position. If a subject adopts a forward lean instead of an erect trunk position, this results in a larger extension moment at the hip. Thorstensson et al. [30] did not report an increase in trunk flexion with an increase of speed but they did mention that the inter-individual variation for this movement was large, which can explain the higher inter-subject variability for the hip moment and the increased level of reliability found at fast speed. The results of the present study reveal that reliability in the frontal plane is not modified by an increase in the walking speed. Fast walking speed increased the SEMs-values of power parameters at the hip in the sagittal plane. These results might be associated with a more noticeable increase in the peak magnitudes of the power curve at the hip and corroborate with the findings of Chen et al. [7].

Many authors have suggested normalizing the kinetic parameters to body mass in order to compare inter-subject patterns [17, 22, 23, 28, 34] since it reduces the inter-subject variability [23, 34]. In the present study, this normalization measure reduced the level of reliability for several kinetic parameters, which can be explained by the reduction in inter-subject variance, which directly affects the value of the dependability coefficients. As reported by Strube and Delitto [29], ideally the universe score (inter-subject) variation must be quite high, suggesting that the score differences obtained are due largely to differences in the subjects' underlying universe score. The D-study revealed that the level of reliability for several normalized joint moment and power parameters, unreliable at self-selected speed, is not significantly increased even if the subjects are assessed at two different visits. These findings suggest that absolute data might be more appropriate to assess differences in kinetic gait parameters after treatment, particularly if the weight of the subjects is not modified during this period. Beck et al. [2] reported that changes in weight due to growth can be found in children of four to 16 years old, tested at intervals more than 3 months apart. This finding should also be taken into consideration when assessing subjects with AIS. For group comparison, it might be better to match subjects based on their anthropometric characteristics than to normalize kinetic values to body mass.

Limits of the study

The reliability of gait parameters can be assessed by different statistical methods. The generalizability theory and

SEMs were used to achieve this goal in the present study. However, even if the generalizability theory is a useful statistical tool for clarifying the various sources of error associated with gait parameter measurements, interpretation of lower coefficients of reliability must be handled with caution when the gait parameters do not have enough inter-subject variability. The reliability of the kinematic and kinetic gait parameters was reported for one side (right). According to Steinwender et al. [28], the reliability for the kinematic and kinetic parameters was not significantly different between the right and left legs. This will need to be verified in subjects with AIS.

Conclusion

Several gait parameters are reliable among subjects with AIS. Future studies will demonstrate if these findings are useful to quantify the impact of treatment or of the progression of scoliosis on gait parameters. The inter-trial reliability is higher than the test–retest reliability for all gait parameters but the test–retest reliability was higher for the kinetic than for the kinematic parameters. Gait speed affects test–retest reliability of several parameters and must be taken into consideration. Normalization to body mass reduces the test–retest reliability for the kinetic parameters, which suggests that it might be more appropriate to assess the reliability with absolute gait data in a specific group if there are no growth effects. In this study, the generalizability theory proved to be a useful statistical tool for clarifying the various sources of error associated with gait parameter measurements and to determine the error of measurement for each parameter.

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