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Review article

Dynamic balance assessment during gait in spinal pathologies – A literature review

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ABSTRACT

The role of the spine as a gait stabilizer is essential. Dynamic assessment, while walking, might provide complementary data to improve spinal deformity management. The aim of this paper was to review spine dynamic behavior and the various methods that have been used to assess gait dynamic balance in order to explore the consequences of spinal deformities while walking. A review was performed by obtaining publications from five electronic databases. All papers reporting pathological or non-pathological spine dynamic behavior during gait and dynamic balance assessment methods were included. Sixty articles were selected. Results varied widely according to pathologies, study conditions, and balance assessment techniques. Three methods assessing dynamic stability during gait were identified: local-orbital dynamic stability, tri-axial accelerometry, and dynamic stability margin. Data from conventional gait analysis techniques were established essentially for scoliosis and low back pain, but they do not assess specific consequences on gait dynamic balance. Three techniques investigate gait dynamic balance and have been validated in normal subjects. Further investigations need to be performed for validation in spinal pathologies as well as the value for clinical practice.
Level of evidence: Level IV.

1. Introduction

Given that walking is the most frequent activity of daily living, it is conceivable that a dynamic assessment of spinal motion and the consequences on gait should be performed regarding spinal deformities. Spine function is inseparable from gait function. The trunk is not a passive "passenger unit" of the human body. The trunk movement assists with active control of whole body mechanics in order to achieve the goal of efficient locomotion. Therefore spinal deformities modify the natural balance of walking. Gait is a challenging balance task, which can be assessed by different methods including mathematical algorithms and physical parameters. Three techniques exist to assess gait stability. The first method is the local (the maximum Lyapunov exponent λ_{max}) and orbital (Floquet multiplier FM_{max}) dynamic stability that quantify dynamic behavior of a system after infinitesimally perturbations. Each one considers different properties of a system dynamics [1]. Local stability (short-term λ_S and long-term λ_L exponents of the λ_{max}) can be defined as the system's sensitivity to respond continuously and in real-time [2]. In other words, this is how quickly a system converges toward or diverges away from the original trajectory. An increase λ_{max} indicates a decrease in local dynamic stability. Orbital stability quantifies the tendency of a system's state to return to the periodic limit cycle orbit. The FM_{max} is calculated at a single point during the cycle and quantifies how much small perturbations grow or decay after one subsequent stride. An FM_{max}'s magnitude below one indicates orbital stability. The second method consists of tri-axial accelerometry that measures segment accelerations during gait. It is based on the concept that gait balance has the capacity to minimize upper body sways [3]. The lack of smoothness in trunk acceleration represents a suggestive assessment of gait dysfunction and imbalance. The third method, the dynamic stability margin (DSM), is defined as the shortest distance between the extrapolated center of mass (xCoM) and the boundaries of the base of support (BoS), and corresponds to the divergence that a system can handle before losing balance [4–6]. The xCoM is a function from the CoM vertical projection and velocity, the gravity, and trochanteric height [4,7,8]. A positive distance indicates stability [5,9].

The aim of this paper was to review spine dynamic behavior and the various methods that have been used to assess gait dynamic balance in order to explore the consequences of spinal deformities while walking.

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2. Materials and methods

The electronic search of databases with retrieval of published reports indexed from 1959 to 2014 was limited to MEDLINE, EMBASE, Cochrane Library, CINAHL, and SCOPUS. Only full-texts in English or French were selected from the electronic databases. Titles and abstracts included the following combination of words: (spine OR trunk OR back OR upper body OR spine deformity) AND (gait OR walking OR locomotion OR motion AND balance OR stability) AND dynamic. Complete procedure and exclusion criteria are summarized in Fig. 1. Additional manual search was supplemented by screening relevant articles based on the reference list of the retrieved articles.

3. Results

A meta-analysis was not possible due to the fact that the studies were widely variable in terms of the dynamic balance assessment methods and in terms of pathologies.

3.1. Search yield

Initially the electronic searches yielded to a total of 7164 titles. The title and abstract screening process eliminated 7031 articles. Following the application of inclusion and exclusion criteria, 44 articles were selected for review. Sixteen articles were retrieved from the reference lists, yielding a final total of 60 articles.

3.2. Study characteristics

Study design retrieved 23 comparative observational studies, 35 non-comparative studies, and 2 reviews. Nineteen studies were performed on a treadmill, 37 were performed over ground, and two studies included both conditions. Thirty-three studies were performed at a self-selected normal walking speed only, 20 at different speeds, and in 5 studies the walking speed was not reported.

3.3. Participants

The reviewed articles tested participants with varying numbers of subjects (5 to 282), varying ages, and varying physical characteristics. Among the non-comparative studies, only five included pathological subjects. Among the comparatives studies, there were no significant differences in ages for 20 studies. Minors were tested exclusively in articles related to adolescents with idiopathic scoliosis [10–22].

3.4. Spine/trunk dynamic behavior during gait

Nine non-comparative studies reported trunk dynamic behavior during gait as is summarized in Table 1 [23–31].

3.4.1. Kinematics

All studies used a different spine kinematic model since marker placement and trunk segments were variable and demonstrated a high variability in range of motion (ROM) values between subjects (Table 1) [23–26,30]. However, patterns of motion were similar and the upper trunk demonstrated higher ROM than the lumbar spine [24,30,32,33]. The high decrease of horizontal acceleration confirms the role of the trunk as a gait progression stabilizer, as well as the opposite movement between lumbar spine and pelvis and between the head-neck unit and the trunk [26,27].

3.4.2. Kinetics

Few authors reported trunk normal kinetics during gait. The trunk principal forces during gait act in the antero-posterior and medio-lateral planes and peaks of forces occur simultaneously and increase with velocity [28]. Natural asymmetries of human walking are well emphasized by ground reaction force (GRF) data [10–12,24].

3.4.3. Electromyography (EMG)

While walking, spinal extensors and abdominal muscles play a fundamental role to control inertial and gravitational forces. Muscular activity increases with walking velocity in order to stabilize the spine around a neutral position [31]. The erector spinae muscles have a phasic activity pattern with peak activity at heel-strike [26,30]. Given to phasic activity between abdominal muscles and ES, results are controversial. Only one study used fine-wires and emphasizes a near constant activity of lateral abdominal muscles during gait [31].

3.5. Dynamic behavior in spinal pathologies

3.5.1. Scoliosis

Thirteen studies reported gait anomalies related to scoliosis and are presented in Table 2 [10–22,34]. Patients' characteristics regarding the deformity were variable (Table 3).

Temporal-distance parameters are controversial (Table 4). AIS subjects have about 7% shorter step length but authors do not agree on the decrease of gait velocity [10,17,19,21,22]. Opposite results are reported regarding the influence of the severity of the deformity on time-distance parameters [19,21]. One study assesses time-distance parameters in adult scoliosis and demonstrates a reduced velocity due to a reduced cadence [34].

AIS subjects do not show side-to-side 3D kinematic trunk asymmetry except for in the transverse plane [17,19,22]. Lower limb kinematics showed a reduction in ROM of the pelvis, hips, and shoulders in the frontal plane; a transverse plane reduction of hip ROM; and a sagittal plane reduction of knee ROM [19,21]. Regarding the influence of the severity of the deformity, reported results are controversial for pelvis, hip, and knee sagittal ROM [19,21]. Adults with scoliosis have an increased trunk flexion and restricted ROM in the frontal and transverse plane, with a slight tilt and external rotation on the right shoulder [34]. Lower limb kinematics are not affected [35].

Kinetics are relevant and reliable to assess gait compensatory mechanisms [10,14,15,20,22]. Tri-dimensional GRF are asymmetrical and higher impulses on the concave side is reported as a control mechanism [10,11,13]. Given to the severity of the deformity and the main side of the primary curve, the lack of correlation with GRF asymmetries has been proved [13,15,20].

Only two reports assessed paraspinal muscular activity during gait and do not agree on the pattern of activity [16,19]. However, paraspinal muscle activity is extended during gait. Abdominal and lateral abdominal muscle activity was not reported.

3.5.2. Low back pain (LBP)

The ten comparative studies assessing spinal behavior in LBP during gait are summarized in Table 5 [36–45].

Patients exhibit significant slower walking speeds, shorter strides and slower cadence (Table 4) [36,38–40,42,43]. Slow walking is a voluntary choice. LBP patients are able to voluntarily increase their walking velocity with controversial results regarding the level of significance [36,38,40].

LBP patients exhibit similar bi-phasic kinematic ROM compared to control subjects without LBP [38,41–43]. Nevertheless, the timing is different, even more with increased velocity. LBP patients do not achieve physiological anti-phase trunk-pelvis rotation [41–43]. They have higher harmonicity confirming more stiffness of the transverse plane trunk motion. Conversely, in the frontal plane, subjects show more variability and less tight movements [41].

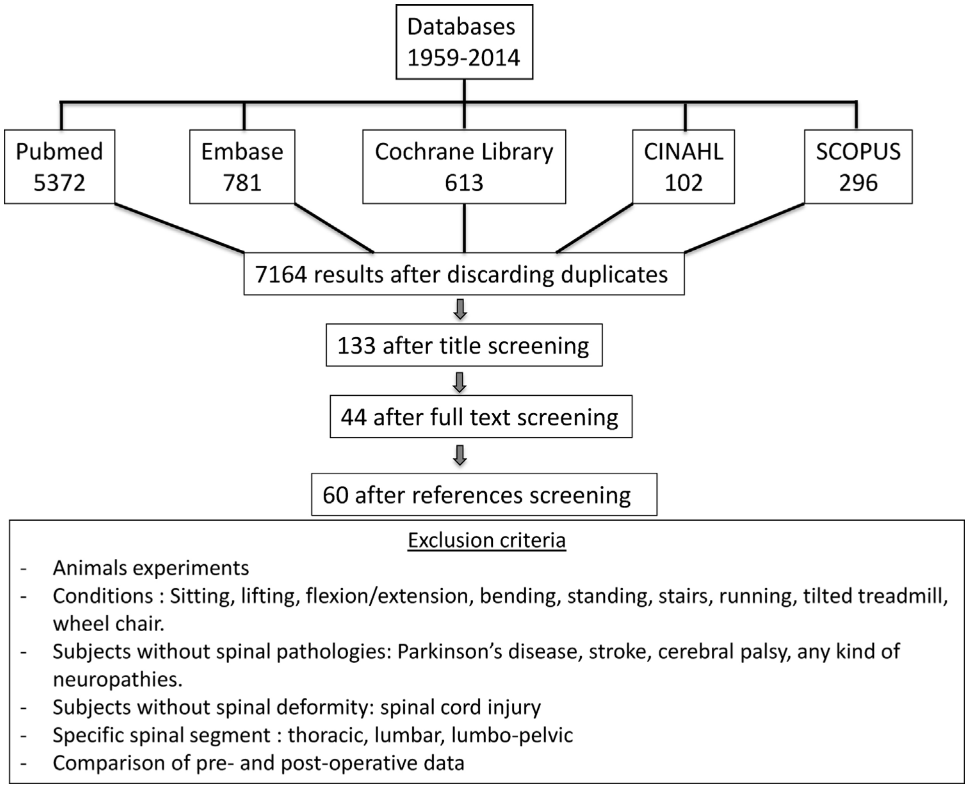

Fig. 1. Flow-chart reporting studies selection and exclusion criteria.

Three-dimensional GRF in patients with LBP are reduced but results differ whether there are or are not significant differences from normal subjects [36,39,40].

Erector spinae (ES) have an earlier onset with higher and extended activity during gait [38,41,45]. Studies considering average values from consecutive trials do not enhance different patterns [38,45]. Once the variability is considered, pattern of activity enhanced less coordination for ES activation in LBP [41]. The rectus abdominis (RA) has the same abnormal pattern described above for ES [45]. Obliquus externus muscles (OE) activity does not differ from normal subjects.

3.5.3. *Postoperative flatback syndrome*

All data suggest a compensatory mechanism to achieve more stability during the stance phase, in response to the anterior position of the center of mass (CoM) in only one study [44]. Subjects have shorter step length, increased step width, and slower velocity. Kinematic data reveal an increased anterior tilt of the torso and a decrease ROM. Patients also exhibit increased knee flexion and hip external rotation. Kinetics showed decrease vertical GRF.

3.5.4. *Spondylolisthesis and ankylosing spondylitis*

As sagittal imbalance is evident in these pathologies, no study reported specific results from spinal dynamic anomalies and imbalance during gait.

3.6. *Assessment of dynamic balance*

3.6.1. *Local and orbital dynamic stability*

Eight non-comparative studies were reviewed for local and orbital dynamic stability and are summarized in Table 6 [1,2,32,46–50].

To achieve reliable precision of the λ_L measurement, a minimal time of five minutes is required [1]. However, too large data series lead to overlapped data for the orbital dynamic stability and the λ_S exponent [47,49]. Treadmill walking reduces both components of the local stability and, to a lesser extent, the orbital dynamic stability [2,51]. Increase of walking velocity affects local-orbital dynamic stability measurements by increasing the sensitivity to local perturbations [32,48]. The influence of walking speed on the λ_S is still unclear as studies report opposite results [46,48]. Changes in stride width and length influenced local and orbital dynamic stability especially the λ_S exponent, and did not confirm the theory according to which shorter step length and wider step width increased gait stability [50].

3.6.2. *Tri-axial accelerometry*

Seven studies (one review, two comparatives studies, and four non-comparative studies) were reviewed and are summarized in Table 7 [3,52–54,56–58].

Unstable subjects make more frequent, but less consistent, adjustment to stabilize their trunk that leads to a non-smooth pattern [56,59,60]. Nineteen to twenty strides are sufficient to provide reliable data [3]. Test-retest reliability shows high intraclass correlation coefficients (ICC: 0.78–0.93) with low coefficients of variation (C_V: 3.1% to 6.8%) [53,54]. Determination of gait parameters also has good test-retest reliability (ICC: 0.94–0.96 and C_V: 1.37% to 1.64%) [52]. Tri-axial accelerometers are also accurate devices for the estimation of the CoM displacement during gait [58].

3.6.3. *Dynamic margin of stability (MoS)*

Seven studies (two comparative and six non-comparative) were reviewed and summarized in Table 8 [9,46,61–66].

CoM determination can be performed by three different methods: the sacral marker method, the segmental analysis method, and a kinetic method [63]. The determination of the BoS boundaries can either be performed by kinematics, or by a foot pressure measurement device [66]. Two consecutive gait trials are sufficient for

Table 1
Spine normal dynamic behavior during gait.

Study	Participants characteristics	Tool for dynamic stability assessment	Instrumented gait analysis data	Gait conditions
Leardini et al., 2009 [23]	10 healthy subjects 5 M/5 F Mean age 25.2 Y Range [23–30] Y	3 planes ROM	*Kinematics* 14 markers 5 pelvis 9 trunk (C7, T2, T8, T10, L5) 4 segments	Self-selected speed
Stokes et al., 1989 [24]	8 healthy subjects 5 M/3 F	ROM Frontal 9 deg Sagittal 6 deg Transversal 5 deg	*Kinematics* (SELSPOT system, LED) 2 triangular plates: 1 pelvis, 1 thorax Trunk: thorax + pelvis	Treadmill 5 speeds 3 trials 6 consecutive strides/15 sec
Crosbie et al., 1997 [25]	108 healthy subjects 50 M/58 F [20–82] Y	ROM Frontal 7 deg Sagittal 2.5 deg Transversal 4.0 deg	*Kinematics* (Expert Vision, Motion Analysis Corp.) Trunk 13 markers 3 segments Upper/lower/lumbar trunk	3 trials
Cromwell et al., 2001 [26]	8 healthy subjects 6 M/2 F Mean age 25.6 Y Range [22–30] Y	Trunk angular position Surface electrodes: Erector spinae and rectus abdominis	*Kinematics* (Peak Performance Technologies) Trunk: 2 markers C7-S1 *EMG* (Therapeutics Unlimited)	5 trials Self-selected speed
Winter et al., 1993 [27]	5 healthy subjects	Thoracic and pelvis ROM and accelerations Sagittal and Frontal GRF and moments (hip) Surface electrodes: Stride to stride EMG Values average	*Kinematics* Trunk 1 segment (C7-T1 to L3-L4) *Kinetics* 2 force platforms *EMG*: Erector spinae (L3-L4 and T9)	Self-selected speed 10 trials
Cappozzo, 1983 [28]	5 healthy males subjects	Trunk forces and couples	*Kinematics* (Stereo photographic method and active markers) Trunk C1-L4 and bilateral shoulder joint center	Photographic records 4 speeds 4 trials
Herzog et al., 1989 [29]	62 healthy subjects 33 M/29 F Mean age 26 Y	GRF	*Kinetics* (Kistler) 1 force platform	Self-selected speed 10 acceptable trials (5/side)
Saunders et al., 2005 [30]	7 health subjects 6 M/1 F	ROM Frontal 13 deg Sagittal 5 deg Transversal 7 deg Fine-wires: transversus abdominis, obliquus internus and externus Surface electrodes: erector spinae, rectus abdominis	*Kinematics* (VICON, Oxford Metrics) Trunk 1 segment T12-sacrum *EMG* (Aloka SSD) Right side	Treadmill 2 walking speed Kinematics: 10 sec data sampling EMG: 30 sec data sampling
Hu et al., 2012 [31]	12 healthy females Mean age 27.5 Y	ROM L5-S1 moments estimation Fine-wire: Transversus abdominis Surface electrodes Obliquus internus et externus	*Kinematics + kinetics* (Optotrak) 9 Cluster 3 LED. Trunk 2 segments (head-T6, T6-pelvis) *EMG* (Porti, TMS international) *Time-distance parameters*	

Y: years; ROM: range of motion; GRF: ground reaction force.

reliable analyses of the data [65]. There are no significant differences between over ground and treadmill walking [64]. Influence of voluntary change in step width or length and cadence on the MoS are similar to the changes in local and orbital dynamic stability [5,50,61]. Individuals can increase their MoS by wider and longer steps in more ways than shorter and narrower steps can decrease the MoS. Medio-lateral MoS is not affected by walking speed [46]. ICC are high and over 0.75 [65]. It is thought that there is no significant side-to side asymmetry of the MoS during normal walking, however this is controversial [61,64,67].

4. Discussion

The diversity of study participants, methods, and modeling techniques precludes a simple conclusion about dynamic spinal balance assessment while walking. There were a number of domains that were not addressed adequately by any study, for example, most studies did not perform sample size calculations or state the sampling strategy used. Non-comparative studies were mainly tested on healthy subjects and further clinical application needs to be performed for spinal pathologies.

4.1. Spine dynamic behavior during gait

The trunk is a combination of many joints that cannot be assessed individually and is usually considered as a single rigid body. However, according to the level, spine segments exhibit different behaviors during gait [6]. Many kinematic models have been developed but none of them have been validated [23]. A lack of correlation between static and dynamic spine measurement was demonstrated by Lenke et al. [18]. However, both assess different conditions.

The upper body represents 2/3 of the body mass and is the most responsible for the GRF.

Table 2
Dynamic behavior in scoliosis.

Study	Participants characteristics	Mathematical/physical model	Tool for dynamic stability assessment	Instrumented gait analysis data	Gait conditions
Kramers de Quervain et al., 2002 [17]	10 AIS females Mean age 14.4 Y	"Torsional offset" = (minimal + maximal pelvic trunk rotation)/2 GRF	ASI	*Kinematics* (VICON system, Oxford metrics) Trunk 1 segment (T1-S1) *Kinetics* (Kistler) 2 force platforms *Time-distance parameters*	Self-selected speed 5 trials
Yang et al., 2013 [22]	20 AIS subjects 2 M/18 F Mean age 14.9 Y 20 matched controls 5 M/15 F Mean age 14.4 Y	ROM GRF	Correlation coefficient (CC)	*Kinematics* (Motion Analysis Corp.) Trunk 1 segment (T1-Sacrum) *Kinetics* (Bertec FP4060) 2 force platforms *Time-distance parameters*	5 trials
Mahaudens et al., 2009 [19]	41 AIS females Mean age 14.5 Y 3 subgroups according to the severity 13 control females Mean age 16.5 Y	ROM	Surface electrodes: Quadratus Lumborum, Erector Spinae	*Kinematics* (Elite system BTS) 22 markers Trunk: shoulder motion *EMG* (Telemg BTS) *Time-distance parameters*	On a treadmill, 1 speed ($4\,km.h^{-1}$) 20 sec sample, 10 successive strides
Lenke et al., 2001 [18]	30 AIS subjects 2 M/28 F Mean age 14 Y Ranges [12–18]		Kinematic gait spine alignment: Sagittal vertical alignment (SVA), coronal vertical alignment (CVA), acromion-pelvis angle (APA)	*Kinematics* (HiRes Motion Analysis Corporation system) Trunk: 3 segments C7-T10 T10-L4 L4-S2 *Time-distance parameters*	5 to 7 trials
Syczewska et al., 2012 [21]	63 AIS females Age range [12–17] years 3 subgroups according to the severity	ROM		*Kinematics* (VICON) Lower limb, Helen Hayes protocol *Time-distance parameters*	6 trials
Chockalingam et al., 2008 [14]	9 AIS 1 M/8 F Mean age 15.3 Y	A-P and M-L CoP	Symmetry index $SI = [(X_1 - X_2)/0.5 \times (X_1 + X_2)] \times 100$	*Kinematics* (Ariel Dynamics) 4 segments trunk model (C7-T6, T6-T12, T12-L4, L4-S2) *Kinetics* (Advanced Mechanical Technology) 1 force platform	6 trials
Giakas et al., 1996 [15]	20 AIS females Mean age 13 Y 20 control females Mean age 13 Y	GRF Harmonic analysis (frequency domain parameters)	ASI	*Kinetics* (AMTI system) 1 force platform	40 trials (20 per side)
Bruyneel et al., 2009 [10]	10 AIS females Mean age 13.8 Y 15 controls females Mean age 12.6 Y	GRF	AI (asymmetry indicator) = [(right–left)/right × 100]; 5% threshold for dynamic asymmetry	*Kinetics* (AMTI) 2 forces platforms	Gait initiation: forward and lateral steps 5 trials of each
Chockalingam et al., 2004 [13]	16 scoliosis (13, AIS, 2 neurofibromatosis and 1 congenital) 4 M/12 F Mean age 11 Y Range [9–19] Y	A-P and vertical GRF	Symmetry index $SI = [(X_1 - X_2)/0.5 \times (X_1 + X_2)] \times 100$	*Kinetics* (Advanced Mechanical Technologies) 1 force platform	Self-selected speed 3 trials per side
Bruyneel et al., 2010 [11]	10 AIS females Mean age 13.8 Y 15 controls females Mean age 12.6 Y	GRF		*Kinetics* (AMTI) 2 forces platforms	Gait initiation 2 series of 5 trials
Schizas et al., 1998 [20]	21 AIS 1 M/20 F Mean age 16.1 Y	Vertical GRF		*Kinetics* (Kistler) 1 force platforms	Several trials until 5 "clean" foot strikes
Fortin et al., 2008 [14]	20 females with AIS Mean age 14.8 Y Ranges [12–17] Y	GRF		*Kinematics* (Optotrak) Trunk: 3 segments (C7-T8, T8-T12, T12-pelvis) *Kinetics* (AMTI) 3 forces platform	2 speeds: comfortable and maximal 5 trials
Engsberg et al., 2001 [34]	8 adults with scoliosis (idiopathic or degenerative) Mean age 46 Y 6 control females Mean age 44 Y		Kinematic gait spine alignment	*Kinematics* (HiRes Motion Analysis Corporation system) Trunk: 1 segment (C7-S2) *Time-distance parameters*	5–7 trials/subjects

AIS: adolescent idiopathic scoliosis; ASI: absolute symmetry index; A-P: antero-posterior; M-L: medio-lateral; CoP: center of pressure.

Table 3
Adolescent idiopathic scoliosis studies' characteristics.

Authors	Thoracic	Thoraco-lumbar	Lumbar	Double curves	Left side	Right side	Cobb angle Mean/median range	Number of patients	Control group
Yang et al. [22]	x	x	x	x	x	x	[7°–34°]	20	x
Mahaudens et al. [19]		x	x		x	x	20° ≤ Cobb ≥ 40°	41	x
Kramers de Quervain et al. [17]	x		x		x	x	[3°–67°]	10	
Bruyneel et al. [10,11]	x					x	33.4° [17°–78°]	10	x
Giakas et al. [15]							42° [25°–62°]	20	x
Syczewska et al. [21]	x	x	x		x	x	36° [20°–60°]	63	
Chockalingam et al. [13,14]	x	x	x	x	x	x	61° [47°–85°]	9	
Schizas et al. [20]	x	x			x	x	42° [18°–90°]	21	
Fortin et al. [14]				x	x	x	33° [17°–50°]	20	
Lenke et al. [18]	x	x	x	x	x	x	57° [48°–77°]	30	

Back muscles and abdominal muscles have an active part in gait balance by controlling the antero-posterior trunk sways and by controlling rotational movements of the shoulder girdle [26,27,30,31]. Whether abdominal muscles coordinate activity with the erector spinae during gait is not clear [26,30]. Given the most recent study from Saunders et al., abdominal muscles have a lower and antagonist activity compared to paraspinal muscles [30].

4.2. Dynamic behavior in spinal pathologies

4.2.1. Scoliosis

Heterogeneity of the subjects might explain the dissimilar time-distance parameters from this review (Tables 2 and 3). Adolescents with idiopathic scoliosis have shorter steps and certainly a reduced walking speed. In their study Kramers de Quervain et al. do not enhance abnormal time-distance parameters [17]. However, half of their subjects were previously treated by brace with no mention for the treatment duration. As demonstrated by Mahaudens et al., gait parameters are normalized after six months of wearing a brace [68]. The study from Mahaudens et al., could not enhance lower velocity as the subjects walked at the same selected speed on a treadmill [19]. The decrease of the step initiation velocity enhanced progressive adaptive control mechanisms to normalized gait velocity [10].

Considering the dissimilar results from transverse plane kinematics, two studies suggested that AIS subjects have similar ROM amplitude compared to normal subjects, with no precision about frontal balance [19,22]. Trunk rotation is considered to be essential for the assertion of the horizontal line of pelvis and head progression, which are symmetrically oriented in normal subjects. The two studies assessing the influence of the severity of the deformity on kinematics were conducted into different walking conditions [19,21]. Treadmill walking reduces natural gait variability and therefore kinematics might be influenced by the severity of the deformity [32,33,51]. Limitation of lower limb ROM is a compensatory mechanism to maintain frontal plane upper body balance. Results from adult populations suggest age related modifications but the lack of data cannot confirm this assumption [34,35].

Kinetics are reliable and can detect the influence of spinal deformity on gait, but they do not directly assess dynamic stability [14]. GRF asymmetries suggest compensatory mechanisms, with no correlation to the severity of the deformity [10,11,13,15,20,22].

The consequences of the spinal deformity on muscular length explain the asymmetrical pattern of activity demonstrated by Hopf et al. [16]. However, since the direct measurement of muscular force is not possible yet, the role of the extensor muscles as a potential cause of spinal deformity must be interpreted cautiously [69,70].

Table 4
Time-distance parameters in spinal deformities.

Pathology	Authors	Velocity	Cadence	Step length	Stride length	Step width	Stance phase	Swing phase
Scoliosis	Engsberg [34,35] (adults)	↘	↘		Normal	Normal		
	Kramers de Quervain [17]	Normal	Normal	Normal	↘			
	Bruyneel [10,11]	↘						
	Mahaudens [19]	Normal		↘			↘	
	Syczewska [21]	↘	↘	↘				
	Yang [22]	Normal	Normal	↘(right side)	Normal		↗ (right side)	↘ (right side)
Low back pain	Lee [36]	↘ (self-selected speed)						
	Vogt [37,38]	↘		↘	↘			
	Khodadadeh [39]	↘	↘	↘	↘		↗	↗
	Simmonds [40]	↘ (self-selected speed)						
	Lamoth [41,42]	↘			↘			
	Crosbie [43]	↘						

Table 5
Dynamic behavior in low back pain.

Study	Participants characteristics	Tool for dynamic stability assessment	Instrumented gait analysis data	Gait conditions
Sarwahi et al., 2002 [44]	21 postoperative flat back syndrome (3 M/18 F) Control group	ROM Moments GRF	*Kinematics* (Motion Analysis Corp.) Trunk 1 segment *Kinetics* (Bertec) 2 forces platform *Time-distance parameters*	Self-selected speed 3 trials
Khodadadeh et al., 1993 [39]	30 LBP subjects 13 M/17 F Mean age 47 Y 20 controls 9 M/11 F Mean age 40 Y	Moments	*Kinetics* (Kistler) 1 force platform *Time-distance parameters*	6 trials
Simmonds et al., 2012 [40]	20 LBP subjects Mean age 46 Y 20 leg pain + LBP subjects Mean age 46.1 Y 20 healthy controls Mean age 46.4 Y	GRF	*Kinetics* (AMTI) 1 force platform *Time-distance parameters*	3 trials 2 gait speeds: preferred and fastest Absolute and normalized velocity on stride length, body height and stride time
Lee et al., 2007 [36]	20 LBP subjects Mean age 46 Y 20 leg pain + LBP subjects Mean age 46.1 Y 20 healthy controls Mean age 46.4 Y	Vertical GRF	*Kinetics* (AMTI) 1 force platform *Time-distance parameters*	3 trials 2 speeds: preferred and fastest
Vogt et al., 2001 [38]	34 LBP subjects 16 M/22 F Mean age M/F: 34.8/29.4 Y 22 controls 21 M/13 F Mean age M/F: 36.3/32.1 Y	ROM	*Kinematics* (3D ultrasonic system Zebris CMS 70) Trunk: T12-S1	Treadmill 4.5 km/h 30 sec recorded
Vogt et al., 2003 [37]	17 LBP males Mean age 36.3 Y 16 males controls Mean age 33.7 Y		*EMG* (Biovision®) Erector spinae	Treadmill ≥ 20 strides
Lamoth et al., 2002 [42]	39 LBP subjects 12 M/27 F Mean age 38 Y 19 controls 10 M/9 F Mean age 41 Y	Trunk ROM Index of Harmonicity (IH)	*Kinematics* (Selspot, Selcom) Active markers Trunk: 1 segment (T4-Sacrum) *Uniaxial accelerometers* (Coulbourn) Heel-strike moments	Treadmill Velocity sequentially increased from 1.4 to 5.4 km.h^{-1}
Crosbie et al., 2013 [43]	19 LBP subjects 7 M/12 F Mean age 34 Y 19 controls 6 M/13 F Mean age 28.6 Y	Index of Harmonicity (IH) Lower trunk, lumbar and pelvis ROM	*Kinematics* (Motion star wireless) Electromagnetic active markers Trunk 2 segments T6-L1 L1-S2	2 walking speeds Self-selected Fastest comfortable Midstance and heel-strike
Lamoth et al., 2006 [41]	22 LBP subjects 9 M/13 F Mean age 38 Y 17 controls 9 M/8 F Mean age 31 Y	Transversal and frontal plane thoracic, lumbar and pelvis ROM	*Kinematics* (Optotrak) Trunk 2 segments T3-L2 L2-Sacrum *EMG* (Blue-Sensor, Medicotest) Erector spinae T12, L2, L4	Treadmill Self-selected speed and sequential increments
Van der Hulst et al., 2010 [45]	63 LBP subjects 33 M/30 F Mean age 41 33 controls 16 M/17 F Mean age 40 Y		*Kinematics* (VICON, Oxford Metrics) *EMG* (Biotel 99, Glonner Electronic) Surface electrodes Erector spinae L1 and L4 Rectus abdominis Obliquus externus abdominis	Treadmill ≥ 20 consecutive strides Sequential increase velocity

Table 6
Gait stability by local and orbital dynamic stability.

Study	Participants characteristics	Tool for dynamic stability assessment	Instrumented gait analysis data	Gait conditions
Hak et al., 2013 [46]	9 healthy subjects 6 M/4 F Mean age 21.9 Y	Local dynamic stability λ_S M-L and backward MoS	*Kinematics* (Optotrak®)	Treadmill 25 trials, 4 min Comfortable walking speed 5 strides frequencies 5 strides lengths
Dingwell et al., 2006 [32]	12 healthy young subjects 6 M/6 F	Local dynamic stability (λ_S–λ_L)	*Kinematics* (VICON 612, Oxford Metrics) 1 marker T1	Treadmill 4 speeds Percent of the preferred walking speed
Dingwell et al., 2007 [2]	10 healthy subjects 5 M/5 F Mean age 27.1 Y	Orbital (max FM) and local dynamic stability (λ_S–λ_L)	*Kinematics* *Tri-axial accelerometry* (Kistler) Sternum	Treadmill and over ground Self-selected speed
Bruijn et al., 2009 [47]	9 healthy males Mean age 25.5 Y	Orbital (max FM) and local dynamic stability (λ_S–λ_L) Bootstrap procedure	*Kinematics* Active 3D movement registration (Optotrak®) 2 cluster of 3 LED T6 Right foot	Treadmill 2 walking speeds At least 20 min for each speed
England et al., 2007 [48]	19 healthy subjects 6 M/13 F Mean age 22.5 Y	Local dynamic stability (λ_S–λ_L)	*Kinematics* (VICON, Oxford Metrics) 21 markers sacrum to feet	Treadmill 5 walking speed 4 trials of 30 strides
Bruijn et al., 2010 [49]	9 healthy males	Orbital (max FM) and local dynamic stability (λ_S–λ_L)	*Kinematics* Active 3D movement registration (Optotrak®) Cluster of 3 LED (T8) *Inertial sensors* (3D gyroscopes, magnetometers, accelerometer)	Treadmill 3 walking speeds 5 min each
Kang et al., 2006 [1]	20 healthy adults Mean age 40 Y Range [18–73]	Orbital (max FM) and local dynamic stability (λ_S–λ_L)	*Kinematics* (VICON 612 Oxford Metrics) Trunk T1-T10 6 markers	Treadmill 3 trials 5 min
McAndrew et al., 2012 [50]	14 healthy subjects 7 M/7 F [18–35] Y	Orbital (max FM) and local dynamic stability (λ_S–λ_L)	*Kinematics* (VICON, Oxford Metrics) 1 marker C7	Treadmill 3 trials/3 min 6 c $\geq$ 120 consecutive strides

FM: Floquet multiplier; λ_S–λ_L: short- and long-term Lyapunov exponents.

Table 7
Gait stability by tri-axial accelerometry.

Study	Participants characteristics	Accelerometer positioning	Instrumented gait analysis data	Gait conditions
Henriksen et al., 2004 [52]	20 healthy subjects 4 M/16 F Mean age 35.2 Y Range [18–57] Y	Lower trunk L3	*Tri-axial accelerometry* (Mega Electronics) *Time-distance parameters*	3 speeds Back and forth 10-m walkway
Moe-Nilssen et al., 1998 [53,54]	21 healthy subjects 4 M/17 F Mean age 22.9 Y Range [21–26] Y	Lower trunk “Over L3 region”	*Tri-axial accelerometry*	5 walking speeds 10-m walkway Wooden and uneven floor
Iosa et al., 2014 [55]	Elderly subjects	Upper body	*Tri-axial accelerometry* Review	
Auvinet et al., 2002 [3]	282 subjects 138 M/144 F Range from 20 years to > 70 years	Lower trunk L3-L4	*Tri-axial accelerometry* (Locometrix) *Time-distance parameters*	Self-selected speed 40 m walkway 20.48 sec trials (19–20 gait cycles)
Yack et al., 1993 [56]	20 elderly subjects over 65 Y 19 controls Mean age 24 Y	Upper trunk T2	*Tri-axial accelerometry* (EGAXT-F10, Entran Devices)	30-foot walkway 40 sec data 10 consecutive strides analyzed
Iosa et al., 2012 [57]	15 strokes 9 M/6 F Mean age 61.4 Y 15 healthy elderly 10 M/5 F Mean age 65.1 Y 15 healthy young 7 M/8 F Mean age 29 Y	Lower trunk L2-L3	*Tri-axial accelerometry* (FreeSense®, Sensorize)	Self-selected speed 10 m walkway
Floor-Westerdijk et al. [58]	8 healthy subjects Mean age 62.3 Y Range from [50–75] Y	Sacrum Sacral inertial sensor Compensated sacral inertial sensor	*Kinematics* (VICON, Oxford Metrics) Trunk 1 segment Sacral marker and Segmental analysis *Tri-axial accelerometry* (MT9, Xsens Technologies)	Self-selected speed “Walk repeatedly” Center of mass determination

Table 8
Gait stability by the dynamic margin of stability.

Study	Participants characteristics	Tool for dynamic stability assessment	Instrumented gait analysis data	Gait conditions
Hak et al., 2013 [46]	9 healthy subjects 6 M/4 F Mean age 21.9 Y	Local dynamic stability λ_S M-L and backward MoS	*Kinematics* (Optotrak®)	Treadmill 25 trials, 4 min Comfortable walking speed 5 strides frequencies 5 strides lengths
Lugade et al., 2011 [9]	20 healthy young subjects 10 elderly healthy subjects 10 elderly fallers subjects	MoS	*Kinematics* (3 foot markers) *Kinetics* (Advanced Mechanical Technologies) 2 force platforms	4 gait cycle phases: heel-strike, midstance, heel off, toe-off
McAndrew et al., 2012 [61]	13 healthy subjects 7 M/6 F Age ranges [18–35] Y	MoS A-P and M-L directions xCoM	*Kinematics* (VICON MX, Oxford Metrics) 77 markers on the whole body 2 foot markers	Treadmill 3 trials 3 minutes 6 experimental conditions Normal Normal metronome Wider steps Narrower steps Shorter step Longer step
Hahn et al., 2003 [62]	6 elderly subjects with balance disorders 1 M/5 F Mean age 76 Y 9 elderly healthy controls 7 M/2 F Mean age 72 Y	M-L excursion of CoM	*Kinematics* (Motion Analysis Corp.) 27 markers *Time-distance parameters*	Obstacle crossing
Gard et al., 2004 [63]	10 healthy subjects 5 M/5 F	Vertical whole body CoM excursion Comparison of 3 methods The sacral method The segmental analysis The force plates	*Kinematics* (Eagle Digital Real-Time Motion system) 21 markers 12-segment model 5 trunk and arms 7 lower limbs *Kinetics* (AMTI) 6 force platforms	4 walking speeds 0.8–2.0 m/s
Rosenblatt et al., 2010 [64]	10 healthy subjects Mean age 24.4 Y	MoS	*Kinematics* (Motion Analysis) 10-segment rigid body *Time-distance parameters*	Treadmill and over ground Self-selected comfortable walking speed
Süptitz et al., 2012 [65]	11 healthy females Mean age 25.5 Y	MoS Absolute Symmetry Index (ASI) $\mid X_R - X_L \mid / (0.5 \mid X_R + X_L \mid)$	*Kinematics* (VICON, Oxford Metrics) 12-segment body model	Treadmill 6 walking speeds 10 trials per speed

MoS: margin of stability; CoM: center of mass; xCoM: extrapolated center of mass; A-P: antero-posterior; M-L: medio-lateral.

4.2.2. Low back pain

Identification of homogeneous groups of subjects remains difficult in LBP.

The decrease in gait velocity and step length suggests a cautious pattern to reduce and avoid pain, and is confirmed by the ability to increase gait velocity [36,38–40,42,43]. Although subjects' characteristics were dissimilar, both demonstrate the analgesic effect of walking. Significant differences in level of walking velocity result from the age and pain intensity.

LBP has no influence on the amplitude of angular spine movement [38,42,43]. High inter-individual variability can be explained by the lack of homogeneous groups. However, considering harmonicity, results from transverse and frontal plane suggested a compensatory mechanism to avoid pain. As transverse plane motion is smoothened, frontal plane motion increases to maintain the role of the trunk as a gait stabilizer as much as possible.

Patients with LBP walk slowly to reduce their GRF and therefore are more comfortable [36,39,40].

Lamoth et al., by considering motor variability, enhanced controversial results [41]. Most studies considered motor variability as a random phenomenon and therefore averaged the data over strides [37,45]. Motor variability is not simply a reflection of random noise but contains features and regularities that may provide insight into both normal and pathological conditions. However, the extended and earlier activity of the ES and of the RA confirmed their role in the cautious mechanisms to avoid pain and transverse plane stiffness [37,45]. Similarly, OE activity confirmed the highest variability of motion in the frontal plane.

4.3. Assessment of dynamic balance

4.3.1. Local and orbital dynamic stability

Local and orbital dynamic stability are mathematical tools based on the predictability of the next movement after small perturbations. They quantify two different aspects of stability: local stability applies to strongly aperiodic systems and orbital stability to strictly periodic systems [2,71]. As human gait is neither strictly aperiodic, nor strictly periodic, both parameters should be conducted when studying walking. Another issue should also be taken into account regarding markers positioning [1,2,49,50]. Local and orbital stability of one segment does not necessarily assess stability of the adjoining segment, especially if associated with spinal deformities [6].

4.3.2. Tri-axial accelerometry

Body-segments move in a coordinated way with a proper speed to minimize body accelerations and explain the use of accelerometry to assess dynamic stability. Tri-axial accelerometry has many advantages: handheld, light-weight, small dimension, and low cost [3,53,55–57]. Furthermore, the device can be used in real-life environment that cannot be recreated in laboratory experiments [58]. However, modifications and variations of accelerations during gait give only an indirect measure of gait stability. Another limitation is the influence of accelerometers positioning (Table 7) [56]. As demonstrated by Winter al. trunk acceleration is different if considering the upper or the lower trunk [6]. The lack of influence of sensor lumbar positioning has been demonstrated but further investigation needs to be performed to compare with cervical and thoracic positioning [72].

4.3.3. The dynamic margin of stability (MoS)

The MoS provides a direct measure of gait dynamic as spine's angles for static measurements. Difficulties remain in the more reliable way to determine the CoM position [62,63,72,73]. Hahn et al. demonstrated that the whole body CoM is a more sensitive tool to individualize gait imbalance that the individual segments CoM excursion [62]. The choice between methods remains an issue as each method has advantages and disadvantages [63,73]. Segmental analysis and kinetic methods are not influenced by gait velocity. Both methods showed equivalent results and are usually used. Recently, accuracy of tri-axial accelerometers using the compensated sacral inertial sensors was recently demonstrated. It can be performed in real-life conditions but still remains the calculation of the BoS, which cannot be dispensed without kinematic or kinetic measurement in a gait laboratory [58,66].

4.3.4. Techniques comparison

Tri-axial accelerometry and MoS need less time to be accurate than local and orbital dynamic stability [3,65]. Multiple trials of 150 consecutive strides are necessary to assess local and dynamic stability as demonstrated by Bruijn et al. [47]. When dealing with disabled children or adults, amputees, and elderly, it might be difficult for them to achieve the minimal time.

Test and retest studies were performed for tri-axial accelerometry [52,55]. Experimental studies assess accuracy and reproducibility of the local and orbital stability, and the MoS [47,65]. However, as the study of Bruijn et al. was conducted on more than 20 minutes of walking, fatigue may have influenced their results [47]. Treadmills reduce the natural variability of gait but do not affect orbital stability and MoS values [2,67]. However, studies conducted on normal subjects and pathological situation might produce errors that need to be assessed. Both local-orbital dynamic stability and MoS were influenced by voluntary changes in gait parameters (step length and width) [50,61]. Hak et al. did not enhance this difference for local stability, but data recording was conducted on a single two-minute trial that did not reach the minimal time recording for data reliability [46]. The influence of gait velocity was only performed for the local and orbital stability components and increased gait balance if slower [48].

5. Conclusion

Measurements from radiographs while standing are essential to assess postural spinal balance in patients with spinal deformity. However, a dynamic approach to assess the dynamic consequences on gait balance might provide further information to improve the management and treatment of spinal deformities. Standard gait parameters do not reveal a clear understanding of the relationship between gait characteristics in spinal deformity and gait dynamic balance. Three methods are currently validated to assess gait stability in normal subjects. Clinical validation and reproducibility of these measurements in patients with spinal deformity need to be performed. Their additional value to radiographic measurements also needs to be assessed for daily clinical practice.

Disclosure of interest

The authors declare that they have no conflicts of interest concerning this article.

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