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Review

Center of mass in analysis of dynamic stability during gait following stroke: A systematic review

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ABSTRACT

Background: The Center of mass (CoM) analysis reveals important aspects of gait dynamic stability of stroke patients, but the variety of methods and measures represents a challenge for planning new studies.
Research question: How have the CoM measures been calculated and employed to investigate gait stability after a stroke? Three issues were addressed: (i) the methodological aspects of the calculation of CoM measures; (ii) the purposes and (iii) the conclusions of the studies on gait stability that employed those measures.
Methods: PubMed and Science Direct databases have been searched to collect original articles produced until July 2017. A set of 26 studies were selected according to criteria involving their methodological quality.
Results: A compromise between accuracy and feasibility in CoM calculation could be reached using the segmental method with 7–9 segments. Regarding their purposes, two types of studies were identified: clinical and research oriented. From the first ones, we highlighted: the margin of stability (MoS) in the mediolateral (ML) direction, and the angular momentum in the frontal plane could be indicators of dynamical stability; the MoS in the anteroposterior (AP) direction might be able to detect the risk of falls and the symmetry of vertical CoM displacement could be used to analyze energy expenditure during gait. These and other CoM measures are potentially useful in clinical settings, but their psychometric properties are still to be determined. The research oriented studies allowed to clarify that stability is not improved by widening the step in stroke patients and that the impaired control of the non-paretic limb might be the main source of instability.
Significance: This review provides recommendations on the methods for estimating CoM and its measures, identifies the potential usefulness of CoM parameters and indicates issues that could be addressed in future studies.

1. Introduction

Deficits in motor control following a stroke compromise the performance in several daily living activities, with negative impacts on individual independence [1]. Mobility is affected due to reduced gait velocity, muscle weakness, spasticity, and impaired sensitivity [2]. Such impairments make this population more prone to falls in comparison to healthy counterparts [3,4].

Indeed, post-stroke individuals living actively in the community have a high incidence of falls during walking [3]. Along with gait, transfer tasks are also associated with a high risk of falls [5], as they also demand accurate postural control during lower limb action [1]. Actually, regardless of the presence of stroke, mobility relies on the integrity of the postural control system, which is responsible for maintaining body stability during static and dynamic tasks [6].

Body stability, in turn, is achieved when the movement of the center of mass (CoM) relative to the base of support (BoS) is performed in a controlled manner to prevent a fall, even when the former is outside the latter [7]. Therefore, analysis of the relative movement between the CoM and BoS by considering their position and velocity has been employed in body stability studies [8]. For instance, parameters related to the position and velocity of the CoM have been calculated according to the equations of motion of an inverted pendulum model and applied to body stability assessment during gait [9–11]. In addition, when

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investigating the motor control mechanisms underlying body stability during this task, the CoM has been considered as one of the most important variables to be controlled by the central nervous system [12,13].

The analysis of CoM movement during gait has been determined to verify the relationship between kinematic patterns and the risk of falls of older adults [14] and to obtain information on the mobility recovery after a stroke [15]. Since the characteristics of the CoM movement can be considered as indicators of body stability during gait [16], several studies were devoted to understand how the young [17], elderly [10], and persons with stroke [15,18] control the CoM.

Although these studies focused on the same issue, they have applied diverse methods to measure and calculate the CoM, which may lead to different conclusions. This diversity makes designing a study challenging, particularly with stroke patients, when a tradeoff between accuracy and feasibility must be met. For instance, the classical segmental method to estimate CoM in 3D demands a hard decision: Should one prioritize accuracy using a detailed anthropometric model with a large number of body markers, or prioritize the well-being of the volunteers, reducing the number of segments and sacrificing accuracy? Another example is the choice of the CoM parameters among the ones existent in the literature: Which parameters serve as outcome measures in clinical evaluations? Which CoM measures are more adequate to discriminate groups? What conclusions can be drawn from the CoM measures?

Considering this scenario, we conducted this literature review to examine (a) the methodological aspects involved in the calculation of CoM and the subsequent determination of stability indicators; (b) the purpose of analysis of CoM in studies of dynamic stability after stroke; and (c) the main conclusions about gait stability drawn from studies analyzing CoM during gait.

2. Material & methods

The analysis was conducted by two reviewers (GFD, RCDB) by applying the same search strategy. Then, the final searches were combined and the duplicates excluded. All studies were screened by title and abstract according to the inclusion and exclusion criteria. The full texts were analyzed, a quality assessment was performed and the inclusion/exclusion criteria was applied once again. Finally, the data extracted from the selected studies were analyzed. Disagreements between the reviewers were resolved by discussion until a consensus was reached.

2.1. Search strategy

An electronic search was performed using PubMed and Science Direct databases for original articles until July 2017. The search identified articles using the following terms: stroke, gait, center of mass and stability, combined as follows: gait OR walk AND stroke AND center of mass; gait OR walk AND stroke AND stability; gait OR walk AND stroke AND stability AND center of mass. To ensure that all potentially relevant publications were found, the expression “center of mass” was written in 4 different forms: center of mass, centre of gravity, center of mass, and center of gravity. The search was conducted considering title, keywords and abstract. In addition, the same strategy was applied to identify studies referenced by each selected article. The search codes are provided in the supplementary material.

2.2. Quality assessment

The 2 reviewers extracted key details from each selected study using a customized form, and the following data were extracted: study design, aims, participant demographic and anthropometric characteristics, clinical assessments, task description, specifications of instruments, parameters of CoM, main results and conclusions.

A customized checklist was developed based on the model proposed by Downs & Black [19] and the guidelines for critical review of quantitative studies [20] were used to assess the methodological quality of the selected articles. Each question was graded using the following criteria: 2 (satisfying description or justification), 1 (limited details), or 0 (no information). The 12 items used in the quality checklist in this review were: Q1: Were the research objectives clearly stated? Q2: Was the study design clearly described? Q3: Were the eligibility criteria specified? Q4: Were demographic and anthropometric characteristics of participants adequately described? Q5: Were clinical characteristics of participants adequately described? Q6: Was at least a classic functional mobility and balance measure (Berg Balance Scale [BBS], Fugl-Meyer Assessment Score [FMAS], Functional Ambulation Category [FAC], or Timed Up and Go [TUG] test) reported? Q7: Was a dynamic task clearly defined? Q8: Were the instruments and set-up appropriately described? Q9: Was the biomechanical model and marker attachment method clearly described? Q10: Were the center of mass calculations clearly specified? Q11: Were the main results easily interpretable? Q12: Did the conclusion achieve the research objective?

Q9 was applied only for those studies that performed kinematic analysis. Each study was evaluated independently by the 2 reviewers (GFD and RCDB) and in the case of discrepancy, the original article was assessed again by other two raters and a consensus was reached after a discussion.

The percentage score was calculated as the ratio between the achieved score and the maximum possible score, multiplied by 100%. The maximum score was 24 points, or 22 points when Q9 was not applied. The overall quality of papers was rated based on the criteria suggested by Hootman et al., 2011 [21]: low-quality (score < 33.3), medium-quality (score ranging from 33.4 to 66.7%) and high-quality (score > 66.8%).

2.3. Inclusion and exclusion criteria

Articles were included if they were published in English, peer-reviewed, have analyzed the center of mass, involved post-stroke subjects performing a gait-related task, and reached a high-quality score (i.e., > 66.8%). Articles were excluded if they were literature reviews, conference abstracts, did not measure center of mass parameters, involved only orthostatic posture, involved only simulations or robotics, were not performed with humans, did not involve stroke or examined mixed pathologies, included only healthy subjects, or did not establish a relationship between CoM and stability.

2.4. Data analysis

The form used for quality assessment was the same as that used in the analysis performed to answer the question stated in the introduction.

3. Results

3.1. Article selection

A total of 294 articles were found and 210 remained after excluding duplicates. Following title and abstract screening, 74 articles were submitted to a full text assessment. As a result, 25 studies remained. One additional study was included after checking the reference lists. Thus, 26 studies satisfied the inclusion criteria. The results of the article selection process and the exclusion criteria are summarized in Fig. 1.

3.2. Quality assessment

The results of quality assessment of the 26 included papers are shown in Table 1. Two studies were classified as medium-quality as they did not include the aim of the study, present the eligibility criteria, unclear design description, incomplete anthropometric or clinical

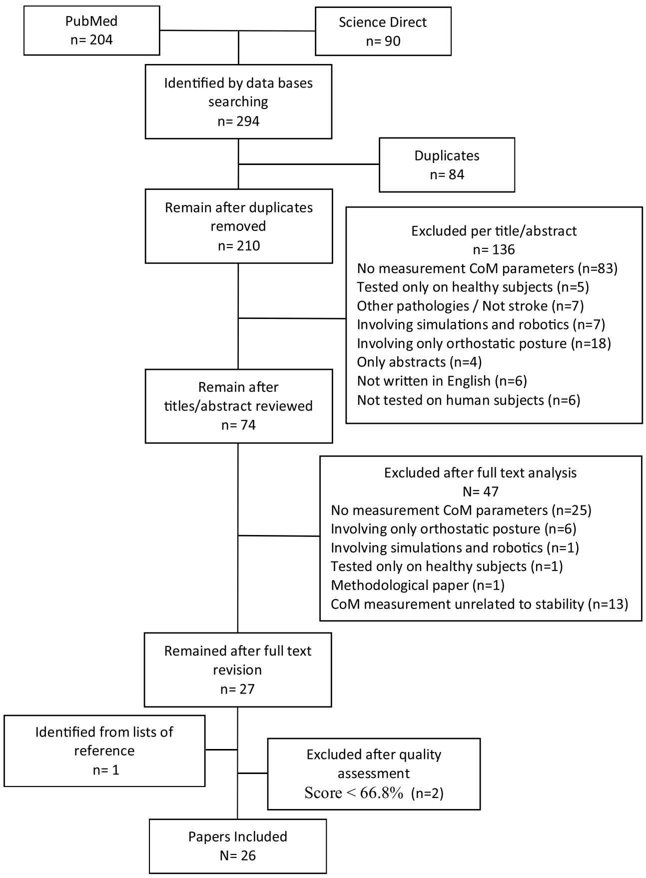

Fig. 1. Flow chart showing the review process.

characteristics and not presenting specifications for CoM calculation.

3.3. Characteristics of reviewed articles

The majority of the reviewed articles (20/26) were published in the last 5 years: 4 were published in the period between 2008–2012 and only 2 were published in the 1990′s. Twenty-four studies that evaluated gait itself, 58% (14 studies) addressed over-ground walking and 46% (11 studies) addressed treadmill walking. One study evaluated gait initiation [25], two studies assessed gait performance during the TUG test and one of them assessed gait performance both during the TUG test and over-ground walking.

Regarding the sample characteristics, most of the studies (17/26) were conducted with chronic stroke survivors: 6 with individuals in the subacute phase and 3 with a mixed sample including individuals in both phases. The mean age of stroke survivors ranged from 36 years [22] to 69 years [23]. The sample size was less than 10 individuals in 5 selected reviewed papers, including 1 study with only 1 stroke participant [24]; 17 studies included 10 to 30 participants and only 4 studies presented samples larger than 35. More details on the samples are provided in Table 2.

3.4. Methods for estimating the CoM the parameters derived from it

A summary of the instruments and the methods used to obtain the CoM, along with the CoM measures employed in the reviewed papers is provided in Table 3.

The methods employed to calculate CoM can be classified as kinematics-based and kinetics-based methods. Among the 22 studies that applied kinematics-based methods to estimate the CoM, 16 used full-body anthropometric models, 5 adopted simplified models and one study did not report the model adopted [15]. The most detailed anthropometric models used 15 segments [23,26] and the most simplified

Table 1
Results of the quality assessment of the included articles.

Included Papers	Q1	Q2	Q3	Q4	Q5	Q6	Q7	Q8	Q9*	Q10	Q11	Q12	Total score	Percentage score
Bonnyaud et al., 2015 [32]	2	1	2	2	2	2	2	1	2	2	2	2	22	92%
Bonnyaud et al., 2016 [33]	2	1	2	2	2	2	2	2	2	2	2	2	23	96%
Dean e Kautz, 2015 [28]	2	1	2	1	1	0	2	1	2	2	2	2	18	75%
Do Carmo et al., 2015 [34]	2	1	2	2	2	2	2	2	2	2	2	2	23	96%
Hak et al., 2013 [18]	2	1	1	1	1	2	2	2	2	2	1	2	19	79%
Hak et al., 2015 [38]	2	1	2	2	2	2	2	2	2	2	2	1	22	92%
Hesse et al., 1997 [25]	2	2	2	1	2	0	2	2	NA	2	2	2	19	86%
Hsiao et al., 2017 [45]	2	1	2	1	2	2	2	1	1	2	2	1	19	79%
Kajrolkar et al., 2014 [36]	2	1	2	2	2	2	2	1	2	2	2	1	21	88%
Kajrolkar et al., 2016 [35]	2	1	2	1	2	2	2	2	2	2	2	2	22	92%
Kao et al., 2014 [15]	2	1	1	1	1	2	2	2	2	0	2	2	18	75%
Kobayashi et al., 2012 [22]	2	2	2	2	2	2	2	2	2	1	2	2	23	96%
Lamontagne et al., 2010 [23]	2	1	2	2	2	0	2	2	2	1	2	2	20	83%
Lan et al., 2013 [27]	2	2	2	2	2	2	2	2	2	2	2	2	24	100%
Massaad et al., 2010 [29]	2	2	2	1	1	0	2	2	NA	2	2	2	18	82%
Nott et al., 2014 [41]	2	1	0	1	1	2	2	2	1	0	2	2	16	67%
Papi et al., 2015 [24]	2	2	0	2	1	0	1	2	1	2	2	2	17	71%
Prassas et al., 1997 [37]	2	2	0	1	1	0	2	2	2	2	2	1	17	71%
Punt et al., 2017a [42]	2	2	2	2	1	2	2	2	2	2	2	2	23	96%
Punt et al., 2017b [43]	2	2	2	2	1	2	2	2	2	2	2	1	22	92%
Said et al., 2008 [39]	2	1	0	1	2	0	2	1	2	2	2	2	17	71%
Van Meulen et al., 2016a [30]	2	2	2	2	2	2	2	2	NA	2	2	2	22	100%
Van Meulen et al., 2016b [31]	2	1	2	2	2	2	2	2	NA	2	2	1	20	91%
Vistamehr et al., 2016 [40]	2	1	1	1	2	2	2	2	1	0	2	2	18	75%
Walker et al., 2016 [44]	2	1	1	1	2	2	2	2	2	2	2	2	21	88%
Zhu et al., 2016 [26]	2	2	2	2	1	0	2	2	2	2	2	2	21	88%

Q1: Were the research objectives clearly stated? Q2: Was the study design clearly described? Q3: Were the eligibility criteria specified? Q4: Were demographic and anthropometric characteristics of participants adequately described? Q5: Were clinical characteristics of participants adequately described? Q6: Was at least a classical functional mobility and balance measure reported? Q7: Was dynamic task clearly defined? Q8: Were the instruments and set up appropriately described? Q9: Was biomechanical model and markers attachment method clearly described? Q10: Were the center of mass calculations clearly specified? Q11: Were main results easily interpretable? Q12: Was the conclusion responds the research objective? Grade: 2 (satisfying description or justification), 1 (limited details), or 0 (no information). NA: not applicable; * Only for those studies that performed kinematic analysis.

Table 2
Sample characteristics of the included articles.

Included Papers	Stroke Subjects				Body Able Subjects		
	Sample	Gender	Age (years)	Time since stroke	Sample	Gender	Age (years)
Bonnyaud et al., 2015 [32]	n = 29	18 M; 11 F	54.2 ± 12.2	7.9 ± 5.7 yrs	n = 25	11 M; 14 F	51.6 ± 8.7
Bonnyaud et al., 2016 [33]	n = 29	18 M; 11 F	54.2 ± 12.2	7.9 ± 5.7 yrs	n = 25	11 M; 14 F	51.6 ± 8.7
Dean e Kautz, 2015 [28]	n = 16	13 M; 3 F	Low fall risk group: 66.0 ± 7.0; Higher fall risk group: 59.0 ± 13.0	Low fall risk group: median 29 (19 to 116); Higher fall risk group: median: 26 (9 to 201) months	n = 19	6 M; 13 F	62.0 ± 8.0
Do Carmo et al., 2015 [34]	n = 14	14 M	53.0 ± 10.3	8 ± 3.8 yrs	n=14	14 M	51.0 ± 5.0
Hak et al., 2013 [18]	n = 10	NR	60.8 ± 8.4	2.4 ± 1.8 yrs	n = 9	NR	57.3 ± 7.2
Hak et al., 2015 [38]	n = 10	6 M; 4 F	57.6 ± 15.4	range 1 to 93 months	-	-	-
Hesse et al., 1997 [25]	n = 14	6 M; 9 F	mean 62.8, range 51 to 84	mean 3.7 (range 2.1 to 8.3) months	n = 10	5 M; 5 F	mean 57.3 (range 46–67)
Hsiao et al., 2017 [45]	n = 36	23 M; 13 F	58.2 ± 12.5	3.7 ± 5.3 yrs	-	-	-
Kajrolkar et al., 2014 [36]	n = 10	7 M; 3 F	range 40 to 64	range 4.5 to 16.2 yrs	-	-	-
Kajrolkar et al., 2016 [35]	n=20	14 M; 6 F	Paretic group: 58.67 ± 5.36; Non-paretic group: 56.80 ± 7.15	Paretic group: 6.10 ± 6.87 yrs; Non-paretic group: 8.79 ± 6.40 yrs	-	-	-
Kao et al., 2014 [15]	n = 9	5 M; 4 F	60.8 ± 9	3.4 ± 3.3 yrs	n = 9	5 M; 4 F	61.7 ± 10.0
Kobayashi et al., 2012 [22]	n=5	5 M	36.0 ± 8.0	16 ± 11 months	-	-	-
Lamontagne et al., 2010 [23]	n = 10	8 M; 2 F	69.0 ± 11.0	mean 12.6 (range 1 to 51) months	n = 11	NR	NR
Lan et al.,2013 [27]	n = 20	11 M; 9 F	55.3 ± 5.9	mean 9.7 (range 5 to 15) weeks	-	-	-
Massaad et al., 2010 [29]	n = 6	2 M; 4 F	47.0 ± 13.0	159 ± 98 months	-	-	-
Nott et al., 2014 [41]	n = 48	29 M; 19 F	58.3 ± 12.0	5.1 ± 3.1 yrs	n = 20	4 M; 16 F	65.1 ± 10.4
Papi et al, 2015 [24]	n = 1	1 M	86	2 months	n = 6	3 M; 3 F	29.8 ± 6.7
Prassas et al, 1997 [37]	n = 8	7 M; 1 F	69.6 ± 11.0	range 2 to 24 months	-	-	-
Punt et al., 2017a [42]	n = 38	18 M; 20 F	Non fallers group: 55.0 ± 12.2; Fallers group: 65.4 ± 6.7	Non fallers: 73.8 ± 53.0; Fallers: 104.0 ± 89.0 months	-	-	-
Punt et al., 2017b [43]	n = 40	16 M; 24 F	Non fallers group: 58.4 ± 14.3; Fallers group: 64.6 ± 8.5	Non fallers: 71.8 ± 65. Fallers: 113.0 ± 109.0 months	-	-	-
Said et al, 2008 [39]	n = 12	NR	65.1 ± 16.6	median 62 (range 15 to 360) days	n = 12	NR	64.3 ± 16.7
Van Meulen et al., 2016a [30]	n=13	8 M; 5 F	64.1 ± 8.7	2.4 ± 1.8 yrs	-	-	-
Van Meulen et al., 2016b [31]	n = 10	7 M; 3 F	63.2 ± 8.9	2.6 ± 2.0 yrs	-	-	-
Vistamehr et al., 2016 [40]	n = 19	NR	62 ± 11	24.0 ± 16.0 months	-	-	-
Walker et al., 2016 [44]	n = 10	5 M; 5 F	range 54 to 65	range 7 to 317 months	n = 10	5 M; 5 F	range 54–66
Zhu et al. 2016 [26]	n=22	16 M; 6 F	Experimental group: 59.18 ± 7.34; Control group: 58 ± 6.97	Experimental group: 3.9 ± 0.83; Control group: 3.72 ± 0.78 months	-	-	-

M: male; F: female; NR: not reported; yrs: years. Values are expressed as mean ± standard deviation, unless stated otherwise.

Table 3
Summary extracted data from included papers.

Included Papers	Dynamic task specification	Measures derived from the CoM	Instruments specifications and set-up	CoM calculations
Bonnyaud et al., 2015 [32]	Overground - 3 TUG tests at a comfortable speed.	Amplitude of CoM displacement in ML and vertical directions. Maximum velocity of CoM displacement.	100 Hz; Motion Analysis Corporation; 34 markers (Helen Hayes marker set);	12- segment model; CoM calculations according to Dempster´s anthropometric table and Hasan et. al (1996)[a] method.
Bonnyaud et al., 2016 [33]	Overground - 3 TUG tests under standardized conditions.	The trajectory of the CoM was analyzed by calculation of the global trajectory length, Hausdorff Distance and Dynamic Time Warping	8 cameras (100 Hz); Motion Analysis Corporation; 34 markers (Helen Hayes marker set);	CoM calculations according to Dempster´s anthropometric table.
Dean and Kautz, 2015 [28]	Instrumented treadmill - walking at a self-selected and fastest-comfortable speeds.	ML position and velocity of the swing heel, relative to CoM, at the initial and at the end of the step. Average value of these parameters during the first half of the swing phase.	100 Hz; Vicon Motion Systems; Markers placed on the sacrum, left and right heels;	The sacrum marker was used as an estimate of mediolateral CoM position
Do Carmo et al., 2015 [34]	Overground - walking at a comfortable self-selected speed.	Maximum CoM displacement, in the stance and swing phases, in ML and vertical directions. Difference between peak displacement at stance and swing phases. Amplitude of CoM displacement in ML, AP and vertical directions.	4 video cameras (75 Hz); D Video kinematic analysis system;	13-segment model based on Zatsiorsky et al., (1990)[b] and De Leva et al., (1996) [51].
Hak et al., 2013 [18]	Instrumented treadmill - walking in combination with Virtual Environment (at self-paced speed, unperturbed walking, continuous perturbation, and virtual obstacle avoidance).	MoS in ML and AP (backward) direction at the instant of initial contact.	12 cameras (120 Hz); Vicon Motion Systems; 16 markers (lower body Plug-in-Gait);	CoM position estimated as the average of pelvis markers.
Hak et al., 2015 [38]	Instrumented treadmill - walking in combination with Virtual Environment (at different combinations of stride frequency and length).	MoS in ML and AP (backward) directions, at the moment when MoS reached its minimum, within the step.	12 cameras (120 Hz); Vicon Motion Systems; 16 markers (lower body Plug-in-Gait);	CoM was estimated as the average of the pelvis markers.
Hesse et al., 1997 [25]	Overground - after an acoustic signal, participants stated to walk at self-adopted speed five times with each limb.	Gains in CoM speed in AP direction during each of three sections of the swing phase.	2 triaxial force platforms (100 Hz);	Integration of ground reaction forces divided by body mass, yielding the velocity of the CoM
Hsiao et al., 2017 [45]	Instrumented treadmill - at comfortable walking speeds and maximal walking speeds on a split-belt treadmill.	The minimum CoM - CoP distance in the ML direction during single limb stance.	8 cameras; Motion Analysis Corporation;	Model created on Visual 3D version 5.0 (C-Motion Inc, Germantown, MD), based on Hanavan (1964)[c].
Kajrolkar et al., 2014 [36]	Overground - After establishing the baseline walking, a slip was introduced without warning (two trials of perturbation on non-involved leg).	Stability measure based on CoM position, velocity and a simulated threshold related to loss of balance (Pai and Iqbal, 1999)[d].	6 cameras; Motion Analysis Corporation; 24 markers;	12-segment representation of the body calculations according to De Leva (1996)[51].
Kajrolkar et al., 2016 [35]	Overground - A slip was introduced without warning after establishing the baseline walking (one slip trial on the paretic or nonparetic limb).	Stability measure based on CoM position, velocity and a simulated threshold against loss of balance (Pai and Iqbal, 1999)[d], at the instant of touchdown of the slipping limb, right before the slip onset, and the post-slip instant right before contralateral limb touchdown.	8 cameras (120 Hz); Motion Analysis Corporation; 24 markers (full body);	12-segment representation of the body based on De Leva (1996) [51].
Kao et al., 2014 [15]	Treadmill - walking at four different speeds: 60%, 80%, 100% of preferred walking speed and fastest attainable speed.	MoS calculated in AP and ML directions, at heel strike for each foot.	8 cameras (120 Hz); Motion Analysis Corporation; 46 markers (lower body, trunk and C7 vertebra);	CoM calculations was not reported.
Kobayashi et al., 2012 [22]	Overground - walking at comfortable self-selected speed with and without ankle foot orthosis.	The height of the peaks during the stance phases of the affected and non-affected legs.	6 cameras (120 Hz); Motion Analysis Corporation; 12 markers (acromion, femoral trochanter, femoral condyle, malleolus, fifth metatarsal, and heel);	CoM was calculated by referring to anthropometric data for each body segment of the rigid-linked model defined by the markers.
Lamontagne et al., 2010 [23]	Overground - walking at a comfortable speed in a large open space, while visualizing a virtual environment.	Angular orientation of CoM trajectory, of 2 m to 3 m length, along the horizontal plane.	10 cameras (120 Hz); Vicon Motion Systems; Plug-In-Gait model, except for the head (represented by a 3-marker model on the helmet-mounted display);	15-segment kinematic model and the anthropometric characteristics of the participants.
Lan et al., 2013 [27]	Overground – walking as fast as possible with and without ankle foot orthosis.	Amplitude of CoM displacement in the vertical and lateral directions during gait cycle.	6 cameras (50 Hz); Quatisys System; 18 markers (shoulders, pelvis, anterior and lateral of knees, ankles, heels, metatarsus, 12 thoracic vertebrae, sacrum)	The CoM was located just anterior to the second sacral vertebra
Massaad et al., 2010 [29]	Instrumented treadmill- walking at a comfortable self-selected speed in a split belt treadmill.	Amplitudes of the vertical CoM displacement over a walking stride, over the paretic and nonparetic steps.	Instrumented treadmill (100 Hz); Elite System;	A double mathematical integration of the vertical CoM acceleration according to Cavagna (1975)[e].

(*continued on next page*)

Table 3 (*continued*)

Included Papers	Dynamic task specification	Measures derived from the CoM	Instruments specifications and set-up	CoM calculations
Nott et al., 2014 [41]	Instrumented treadmill - walking at a comfortable self-selected speed in a split belt treadmill.	Change in the angular momentum (H) during each of 6 sections of the gait cycles. Step to step variability of the time derivative of H and of the foot placement, defined as the ML distance between the paretic foot and the CoM at the onset of the first half of the paretic stance phase.	12 cameras (100 Hz); Vicon Motion Systems;	13-segment customized model created in Visual3D.
Papi et al., 2015 [24]	Overground - walking at comfortable speed with and without ankle foot orthosis.	Stability index based in the variance of the CoM and of the joint angle trajectories (Sholz and Schöner, 1999)[74] in the sagittal plane, during stance of the paretic limb.	12 cameras (100 Hz); Vicon Motion Systems;	CoM estimated as a fixed point (intersection of the diagonals pelvis markers).
Prassas et al., 1997 [37]	Overground - walk on the walkway without rhythm in a comfortable speed and with rhythm as pacemaker.	Amplitude of CoM displacement in ML and vertical directions and horizontal velocity. No information on the parametrization of vCoM is given.	2 cameras (60 Hz); Ariel Performance Analysis System; 12 markers;	11 rigid link system (Dempster data) based on Hay (1993)[f] procedure.
Punt et al., 2017a [42]	Instrumented treadmill - walking under six types of gait perturbations in ML and AP directions at 0.41 m/s.	MoS in AP (forward) and ML directions at the instant of foot contact (Hof et al., 2005)[9].	10 cameras; Vicon Motion Systems; 47 markers;	14-segment model based on Zatsiorsky (1998)[g].
Punt et al., 2017b [43]	Instrumented treadmill - walking on a dual-belt treadmill at a preferred speed.	MoS in AP (forward) and ML directions at the instant of foot contact (Hof et al., 2005)[9].	10 cameras; Vicon Motion Systems; 47 markers;	14-segment model based on Zatsiorsky (1998)[g].
Said et al., 2008 [39]	Overground - unobstructed walking trials at comfortable speed and walking trials for the 4 cm high obstacle and a wide obstacle.	AP CoM velocity at 6 critical instants; peaks of ML CoM velocity in 2 intervals, one before and other after obstacle crossing. CoM-CoP and CoM-BoS (stance heel) distances in the AP direction, at lead toe clearance.	6 cameras; Vicon Motion Systems; 21 markers;	7-segment Body Builder 1 model according to Winter (1990)[h].
Van Meulen et al., 2016a [30]	Overground timed 10-meter walking test at a self-selected comfortable pace, while wearing instrumented shoes.	Shortest distance from the xCoM to the front line of the BoS in the AP direction.	A pair of instrumented Xsens Force Shoes (50 Hz);	CoM calculations according to Schepers et al., (2009)[i] method.
Van Meulen et al., 2016b [31]	Overground - TUG test and 10 m walking tests at a comfortable speed, while wearing instrumented shoes.	Percentage of the time when the MoS in AP (forward) direction was positive; symmetry of MoS in the ML direction. Both parameters were calculate during the walking and the turning phase of the TUG.	A pair of instrumented Xsens Force Shoes (50 Hz);	CoM calculations according to Schepers et al., (2009)[i] method.
Vistamehr et al., 2016 [40]	Instrumented treadmill - walking on a split belt treadmill at self-selected walking speed.	Minimum distance between CoM and the BoS in the ML direction, at each step; peak to peak range of angular momentum in the frontal plane over the gait cycle.	12 cameras (100 Hz); Vicon Motion Systems;	13-segment inverse dynamics model (C-Motion, Inc., Germantown, MD).
Walker et al., 2016 [44]	Instrumented treadmill- walking on a split-belt treadmill under six conditions altering the amount and type of visual information.	Amplitude of CoM displacement in the frontal plane (CoM sway), over the gait cycle; ratio of the step width to CoM to the amplitude of CoM displacement.	6 cameras (100 Hz); Vicon Motion Systems; 15 markers (Plug-In-Gait model) with 7 additional markers (shoulders, C7, head);	8-segment model, according to Winter, (2009)[53].
Zhu et al., 2016 [26]	Overground - walking at a self-selected speed.	Peak values of the CoM in the vertical direction during paretic and non-paretic stance phases; amplitude of CoM displacement in the ML direction.	16 cameras (100 Hz); Motion Analysis Corporation; 15 markers	15-segmento model according to Dempster anthropometric data of segments and Schepers et al. (2009)[i] method.

TUG: Timed Up and Go; CoM: Center of Mass; CoP; Center of Pressure; ML: Mediolateral; AP: Anteroposterior; MoS: margin of stability; xCoM: extrapolated CoM; BoS: base of support.

[a] S.S. Hasan, D.W. Robin, D.C. Szurkus, D.H. Ashmead, S.W. Peterson, R.G. Shiavi. Simultaneous measurement of body center of pressure and center of gravity during upright stance. Part I: Methods. Gait Posture. 4 (1996) 1–10.

[b] V.M. Zatsiorsky, V. Seluyanov, L. Chugunova. In vivo body segment inertial parameters determination using a gammascanner method. Biomechanics of Human Movement Application in Rehabilitation, Sports and Ergonomics. Worthington, 1990.

[c] E.P. Hanavan. A mathematical model of the human body. AMRL-TR-64-102. Aerospace, 1964.

[d] Y.C.Pai, K. Iqbal. Simulated movement termination for balance recovery: can movement strategies be sought to maintain stability in the presence of slipping or forced sliding? J. Biomech. 32 (1999) 779–786.

[e] G.A.Cavagna. Force platforms as ergometers. J Appl Physiol. 39 (1975).174–179.

[f] J.G. Hay. The Biomechanics of Sports Technique. Englewood Cliffs, Prentice Hall. 1993.

[g] V.M. Zatsiorsky. Kinetics of human motion. 1998.

[h] D.A. Winter. Biomechanics and motor control of human movement. New York: John Wiley & Sons, Inc.; 1990.

[i] H.M. Schepers, E. Van Asseldonk, J.H. Buurke, P.H. Veltink. Ambulatory estimation of center of mass displacement during walking. IEEE Trans Biomed Eng. 56 (2009):1189–1195.

one used a single marker in the sacral region as an estimate of the CoM position [27,28]. The studies that adopted kinetics-based methods employed either force platforms [25,29] or custom-made instrumented shoes [30,31] and estimated the CoM displacement by double mathematical integration of the ground reaction forces by applying Newtonian dynamics.

The majority of the studies that employed kinematics-based methods filtered marker data before calculating the CoM [18,26,32–37]. Except one study [37] all the others employed Butteworth filters of second [35,36], third [26] or fourth orders [18,34,38]. Cut-off frequencies also varied among studies: 6 Hz [26,32–34,37], 10 Hz [18,38] and some used optimal cut-off frequencies ranging between 4.5 Hz–9 Hz [35,36]. Some studies did not provide the filtering specifications, while others did not mention filtering.

The time series of the CoM displacement, i.e. the CoM coordinates over time, was used to derive other continuous variables, such as the CoM instantaneous velocity, (vCoM), in 6 studies, the angular momentum (H), in 2 studies, the extrapolated CoM (xCoM) and the Margin of Stability (MoS), in 8 studies. The vCoM has been calculated as the central finite difference of CoM position [39], or simply not performing a second integration of the ground reaction forces, when the kinetics-based method was employed [25]. The other studies did not report the algorithm for calculating vCoM. The whole-body angular momentum was calculated as the summation of the angular momentum of all segments about the CoM [40,41]. Finally, the MoS is defined as the distance between a reference point in the BoS and the extrapolated CoM (xCoM), which depends simultaneously on the position and velocity of CoM [9]. The reference point in the BoS was considered either as the position of the CoP [30,31,40] or as the position of some anatomical landmark on the foot [15,18,38,42,43].

In order to obtain discrete measures, the time series of the CoM or of other continuous variables, were parametrized as described shortly in the third column of Table 3. Several studies have calculated displacement parameters (peaks or amplitudes) in both the ML and vertical directions [26,27,32,34,37], while other studies considered a single axis in the analysis [22,28,29,44]. Only three studies considered the CoM or its velocity as vectors. Two of them analyzed the absolute value of the CoM velocity vector in three dimensions [31] and in the horizontal plane [37] and one [23] analyzed the direction of CoM movement in the horizontal plane. The parametrization also differed regarding the time interval or the instant at which the values of the variables were taken. Among the studies which analyzed the MoS, some used the smallest value along the whole gait cycle [18,38] or along the stance phase [30,31,40], others used the value at foot contact [15] and others analyzed the symmetry between MoS of paretic and non-paretic limbs [31].

3.5. Purposes and conclusions of the reviewed papers

Two prevailing profiles were identified among the included papers: clinical (12/26) and research (14/26) oriented studies. Clinical-oriented studies were those which focused primarily on the clinical applicability of the CoM parameters, while the research-oriented ones were devoted primarily to understand and/or describe the mechanisms underlying gait stability. We further divided the clinically-oriented studies into 2 categories: those that aimed at analyzing discriminative power of CoM measures and those that employed them to evaluate the clinical intervention outcomes. The purposes of these studies and their main conclusions are summarized in Table 4. Similarly, the research-oriented studies were classified according to other 3 categories: those that compared the CoM measures between groups, those that compared the paretic and non-paretic limbs and those that analyzed the responses to external stimuli. Their purposes and conclusions are summarized in Table 5.

It is important to point out that this categorization was only based on the main conclusions of the papers and did not consider all the results, which could have spanned more than one category. This decision was made to be consistent with our third objective, which focused on the main conclusions related to CoM. However, we have reported here only the conclusions supported by statistically significant findings.

3.6. The use of clinical measures

In the order of frequency of appearance, the clinical scales or instruments most frequently used to characterize the participants were: the Berg Balance Scale (BBS) [18,22,38,40,41,43–45,27,30–36] the Timed Up and Go (TUG) test [31–33,35,36,43], the Fugl-Meyer Assessment Scores (FMAS) [15,34,40,44] and the Functional Ambulation Category (FAC) [18,27,38,42,43].

4. Discussion

In this systematic review, 26 high-quality studies were selected, 10 of which were published between 2016 and 2017, indicating an increasing interest in dynamic stability during gait after a stroke. This review intended, primarily, to provide support for those who are designing experiments and/or therapies related to gait stability of stroke patients and are interested in studying CoM measures.

4.1. Methodological aspects of the calculation of CoM and stability indicators

The calculation of CoM stability indicators involves several steps: signal processing of raw data, calculation of CoM, calculation of continuous variables derived from the CoM and calculation of discrete measures. The methodological issues regarding some of these steps are discussed in this section.

To estimate the CoM position, most of the reviewed papers adopted the classical segmental approach [46], whose accuracy improves as the number of body segments is increased. This approach requires the correct identification of anatomical landmarks and the proper placement of body markers. To accomplish this, the experimenter has to manipulate the subjects position, e.g., transferring the weight from one limb to another or passively moving the relevant body segments [47]. This procedure is particularly challenging when dealing with stroke patients as they might resist to passive movement of their limbs due to spasticity. Stroke patients also exhibit difficulty to execute motor commands due to impairments in the proprioceptive system, weakness or cognitive deficits. In addition, if the procedure lasts too long, the subjects may get tired and not perform well during the gait tests. To deal with these practical challenges, one can adopt a number of simplifications identified among the reviewed papers: (i) to reduce the number of segments to 7 [39] or 8 [44], (ii) to approximate the CoM by a single point defined by the average of the pelvis markers [18,24,38] or by some anatomical landmark in the sacral region [27,28]. It is important to analyze how these simplifications compromise the reliability of the results.

One of the included papers [39] addressed this issue by comparing the CoM obtained with the 7-segment simplified model with the original 12-segment model, for three healthy subjects. They found differences below 4 mm in the AP and ML directions, which were deemed as acceptable. In a work not included in this review, Tisserand and colleagues [48] analyzed the accuracy of a model with nine body segments, reconstructed from 13 markers. They compared the CoM and the xCoM of healthy adults in three dimensions with those obtained from a 16-segment model and reported that the mean distance between the CoM estimates (about 8 mm) was comparable to the error due to tissue artifacts and marker positioning. It was concluded that the simplified model is reliable.

The use of the pelvis segment, solely, to estimate the CoM [18,24,38] or even a unique marker placed at the sacrum is very practical but the impact on the accuracy is critical, as pointed out by

Table 4
Summary of the purposes and conclusions of clinically-oriented papers.

Classification of purpose	Included Papers	Study purposes	Conclusions related to CoM parameters
Analysis of discriminative power of CoM measures	Bonnyaud et al., 2015 [32]	To analyze correlations between TUG performance time, CoM displacement and foot clearance in stroke patients and healthy subjects and to compare these parameters between fallers and non-fallers.	ML CoM velocity was different between stroke and healthy subjects. Vertical CoM velocity during the turn distinguished fallers from non-fallers stroke patients. There were moderate correlations between TUG performance time and: ML CoM displacement and velocity for the "turn" phase; and vertical CoM velocity for the "return" phase.
	Bonnyaud et al., 2016 [33]	To compare the measures length of CoM Trajectory, Hausdorff Distance and Dynamic Time Warping (DTW), driven from CoM trajectories between patients with stroke at left and right; fallers and non-fallers. To analyze the correlation with clinical measures.	The measures distinguished between healthy and stroke group. No differences were found between patients with right and left stroke. DTW in the "go" phase distinguished fallers and non-fallers. There was negative correlation between the BBS HD and DTW.
	Nott et al., 2014 [41]	To compare frontal-plane angular momentum between stroke and healthy controls and verify whether it can serve as a quantitative measure of dynamic balance performance during walking.	The behavior of the frontal-plane angular momentum was different between stroke in healthy controls during the first half of the single (paretic) leg stance. There were inverse correlations between the change in angular momentum and clinical tests (DGI and BBS).
	Punt et al., 2017a [42]	To identify clinical and biomechanical measures that could be good fall predictors.	The MoS AP of both paretic and non-paretic limbs were different between fallers and non-fallers, but they cannot be fall predictors by themselves, they must be associated to other measures.
	Van Meulen et al., 2016a [30]	To relate balance metrics obtained from customized instrumented shoes to standardized clinical measures.	AP MoS and step length symmetry were not significantly correlated with Berg Balance Scale score.
	Van Meulen et al., 2016b [31]	To propose methods to objectively evaluate balance during functional walking	Participants with higher BBS were able to move their CoM more towards their affected side and their AP MoS indicated that they are more frequently unstable during walking, more similar to normal gait.
	Vistamehr et al., 2016 [40]	To investigate the relationship between the peak to peak range of whole body angular momentum with Berg Balance Scale (BBS), Dynamic Gait Index (DGI) and MoS.	CoM based measures (ML MoS and frontal-plane angular momentum) are inversely correlated with the clinical measures (BBS and DGI).
Results of clinical intervetions on CoM measrues	Kobayashi et al., 2012 [22]	To investigate the effect of ankle foot orthosis (AFO) on the sagittal plane displacement of the CoM	The use of an AFO could increase the peak of vertical CoM displacement during stance phase of the paretic limb.
	Lan et al., 2013 [27]	To verify if an AFO helps in the correction of abnormal motion of the body CoM and the pelvis.	With the AFO, the CoM displacement in the vertical direction was larger and the one in the ML direction was smaller, in comparison to the non-AFO condition. There was correlation between gait velocity and the CoM displacement.
	Massaad et al., 2010 [29]	To verify if the visual feedback of CoM trajectory in the frontal plane, during gait training, is able to reduce the walking energy costs in hemiparetic patients.	After the gait training the patients reduced the amplitude of vertical CoM displacement and the energy consumption.
	Prassas et al., 1997 [37]	To investigate the effect of auditory rhythmic cuing on gait kinematic parameters of stroke patients.	CoM vertical displacement was smaller in the presence of rhythmic cues.
	Zhu et al., 2016 [26]	To qualify the improvements of lower limb modified constraint-induced movement therapy (m-CIMT) by assessing the CoM displacement and the basic gait parameters.	The m-CIMT intervention increased the CoM displacement in sagittal plane and decreased in frontal plane.

CoM: Center of Mass; CoP: Center of Pressure; DTW: Dynamic Time Warping; MoS: Margins of Stability; UCM: Uncontrolled Manifold; ML: Mediolateral; AP: Anteroposterior; AFO: Ankle Foot Orthosis; BBS: Berg Balance Scale; DGI: Dynamic Gait Index; m-CIMT: modified constraint-induced movement therapy (CIMT). Only conclusions supported by statistically significant findings were included.

several studies [48–50]. Tisserand et al. [48] demonstrated that estimating the CoM using a single sacral marker introduces errors 2.5 times larger than using a simplified 9-segment model [48]. Huntley and colleagues, evaluated two different simplified models in a work performed with post stroke individuals, using a three-segment model (head-trunk-pelvis) and one-segment model (pelvis) and compared their results against a 13-segment model [49]. The performance of the simplified models did not differ in the ML direction and both approaches exhibited a root mean square error (rms) of 1.5 cm. In the AP direction, the three-segment model performed much better than the pelvis model, which generated errors of about 5 cm. Similarly, Yang and Pai [50] analyzed older healthy adults and found larger rms errors in the AP direction than in the vertical and ML directions (about 3 cm) when comparing the position of the sacral marker with the CoM estimated by a 13-segment model [51]. Actually, besides the inaccuracy, the use of a single point to estimate the CoM is conceptually wrong, as it cannot grasp the behavior of a multi-segment structure [46]. One could wonder if the three-segment model (head-trunk-pelvis) analyzed by Huntley et al. [49] would be a more practical and accurate alternative. However, the error associated is relatively large, probably because the coordinates of the lower limbs are not considered. This is particularly critical after a stroke given the between-limb asymmetry. Therefore, although it may constitute a good model from the practical standpoint, its accuracy is not acceptable.

The kinectis-based methods to calculate the CoM impose other challenges and generate other sorts of inaccuracies. From the practical standpoint, the main disadvantage of using force platforms is the need for at least two consecutive steps on two different platforms. It is not always easy to obtain one step in each platform, especially for those with a pathological gait pattern [16,34]. Therefore, the use of instrumented shoes seems to be an attractive alternative to evaluate the CoM under more realistic conditions that may extrapolate the laboratory environment [30,31]. On the other hand, the use of instrumented shoes may influence the walking pattern, as they are usually heavy and bulky [16]. All these disadvantages could be overcome by using other technologies such as instrumented pathways, but none of the reviewed papers have employed them yet. Besides the practical issues, the kinetics-based method relies on calculations based on the underlying

Table 5
Summary of the purposes and conclusions of research-oriented papers.

Nature of the Analysis	Included Papers	Study purpose	Conclusions related to CoM parameters
Comparison of CoM behavior between groups	Dean e Kautz, 2015 [28]	To verify if the typical strategy for mediolateral stabilization, employed in normal gait, is disrupted after stroke.	Stroke participants classified as having low risk of falls, in general, preserved the neuromechanical strategy of normal gait, namely, controlling the foot placement to accelerate the CoM toward the midline, while those with large risk of falls, did not.
	Hsiao et al., 2017 [45]	To compare the weight transfer characteristics between slow and fast post-stroke ambulators.	Slow walkers had an increased minimum CoM-CoP distance during single stance of the paretic limb and delayed weight transfer to the paretic side, and more lateral paretic limb foot placement relative to the CoM compared to fast walkers.
	Kao et al., 2014 [15]	To directly quantify walking stability in stroke survivors and healthy controls in four different gait speeds; and to determine which stability measures would reveal the changes in walking stability following stroke.	In all gait speeds, stroke survivors exhibited smaller average AP MoS and greater MoS variability both in AP and ML directions, compared to healthy controls. The variability of ML MoS was greater at the affected leg compared to the nonaffected one.
	Papi et al., 2015 [24]	To investigate the feasibility of the uncontrolled manifold (UCM) approach to analyze kinematic gait data of healthy and neurologically impaired individuals.	The analysis demonstrated that the person with stroke used differently the available degrees of freedom at joint level, but all achieved a stable CoM movement during stance phase.
	Punt et al., 2017b [43]	To verify whether characteristics of perturbed gait differ between fallers and non-fallers stroke survivors.	Both groups have a preserved ability to cope with external gait perturbations, but the forward MoS, two steps after the perturbation was different between them, with the fallers presenting a reduced forward MoS in response to an ipsilateral perturbation of the paretic limb.
	Hak et al., 2013 [18]	To examine whether post stroke subjects employ the same strategies as healthy individuals to preserve margins of stability (MoS) during unperturbed walking and under two gait manipulations: (i) ML translations of the walking surface and (ii) an adaptability task, namely, to hit a virtual target by raising the knee.	Post-stroke participants were able to regulate their ML MoS, during all experimental conditions, to the same degree as the able-bodied subjects, but this required a larger step width and a relatively high step frequency. During the gait adaptability task, post stroke patients exhibited a significantly smaller backward margin of stability, compared to healthy controls.
Comparison between limbs	Do Carmo et al., 2015 [34]	To identify and to analyze the alterations of CoM trajectory during gait cycles of both affected and unaffected limbs, of post-stroke patients comparing to healthy subjects.	The main alterations at CoM trajectory were observed in the single support phase of the affected side (higher lateral displacement, lower vertical and forward displacement).
	Hesse et al., 1997 [25]	To compare the symmetry of gait initiation between healthy and hemiparetic subjects.	When starting with the nonaffected leg, the CoP displayed marked mediolateral sway with no corresponding initial movement of the CoM; when starting with the affected leg the movement pattern of CoP and CoM was comparable to that of normal gait.
	Kajrolkar et al., 2016 [35]	To compare the differences between the non-paretic and paretic sides in dynamical stability and protective stepping strategies under external perturbation.	The nonparetic-side slip group exhibited higher values of the stability index at recovery step touchdown, resulting from lower perturbation magnitudes compared to the paretic-side slip group.
	Said et al., 2008 [39]	To examine the impact of stroke on balance during obstacle crossing.	At affected lead toe clearance, stroke subjects exhibited greater CoM-CoP distance compared to healthy counterparts at the same speed, while at affected toe-off the CoM AP velocity was smaller, minimizing the instability.
Response to external stimuli	Hak et al., 2015 [38]	To investigate how adjustments in stride frequency and length influence the backward and ML MoS.	Post-stroke individuals have difficulty in adjusting to step frequencies higher than the self-selected one, and their capacity to increase the backward and ML MoS was limited when this higher frequency was required.
	Kajrolkar et al., 2014 [36]	To examine the control of dynamic stability and the characteristics of the compensatory stepping responses to an unexpected slip under the nonaffected limb and to examine adaptative changes on a second slip.	People with stroke were able to reactively control CoM state stability to decrease fall-risk upon a novel slip. Prior exposure to a slip did not significantly alter feedforward control but improved the ability to use the reactive feedback control for improved slip outcomes.
	Lamontagne et al., 2010 [23]	To examine the ability of individuals with stroke to control their walking direction immersed in a virtual environment describing translational optic flows expanding from different directions.	Stroke patients displayed altered steering behaviors characterized either by an absence of CoM trajectory corrections, multiple errors in the heading direction, or systematic deviation to the nonparetic side.
	Walker et al., 2016 [[44]]	To quantify deficits in dynamic balance control during walking, and to evaluate the influence of different conditions of visual feedback.	Stroke survivors walked with larger CoM sway and wider step widths compared to controls. Despite these baseline differences, both groups walked with a similar ratio of step width to CoM sway. When required to keep a reference of body movement inside a stationary target, the stroke group reduced CoM sway.

CoM: Center of Mass; CoP: Center of Pressure; MoS: Margins of Stability; UCM: Uncontrolled Manifold; ML: Mediolateral; AP: Anteroposterior. Only conclusions supported by statistically significant findings were included.

assumption that the body behaves as a simple inverted pendulum [52] which is not true, irrespective of the gait phase. Moreover, the double integration and filtering lead to the accumulating experimental errors. Therefore, it seems that the kinetics-based method for CoM estimation would not be recommended due to its inaccuracy associated with difficulties in its practical implementation.

Considering the balance between accuracy and experimental feasibility and taking to account the well-being of the subjects, one of the best alternatives to calculate the CoM is the segmental approach using a small number of segments (from 7–9) [39] [44], [48]. However, determining the performance of these models to estimate the CoM, with a significant sample of stroke patients is still required and may provide a valuable contribution.

Before calculating the CoM, filtering of kinematic data should be performed. However, this was not mentioned or not well described in some of the reviewed studies. Among those which described filtering, the most commonly adopted cut-off frequency was 6 Hz, what is not surprising, as it is recommended in the classical literature [53]. The use of an optimal cut-off frequency for each marker [35,36] seem to be a more indicated procedure as 6 Hz has been established for steady-state gait and not for a non-stationary response to perturbation. The authors did not describe the way such optimal frequencies were defined. The residual analyses method, as described by Winter [53], could be a good alternative.

Considering the calculation of the discrete measures derived from CoM, the reviewed studies differ in several methodological aspects. One aspect worth discussing, is the number of strides, which ranges from one [34] to over 60 [42,43]. The relatively high intra-subject gait variability of stroke patients [54], compromises the reliability of parameters calculated from few steps or strides. However, this issue is delicate as the well-being of the volunteers plays a role. The number of repetitions must be carefully chosen considering, the strength of the statistical power, the comfort of the participants and the nature of the question under investigation.

4.1.1. Main findings of studies with clinical focus

Several CoM measures have shown significant correlations with clinical scores, but not all of them. The geometrical characteristics of the CoM trajectories defined by Bonnyaud et al., 2016, were inversely correlated with BBS [33], the frontal plane angular momentum was inversely correlated with the BBS and the Dynamic Gait Index [40,41], as well as the ML MoS [40]. In one study, the authors pointed out that the AP MoS differs between participants with higher and lower BBS [31], but the sample was too small and no statistical analysis was conducted in the study. On the other hand, no correlation between BBS and AP-MoS was found [30]. These findings suggest that the CoM parameters in the frontal plane might be more suitable to identify the instability in stroke patients than those in the sagittal plane. Actually, the parameters in the ML direction have been reported in the literature as good dynamical balance indicators [44]. Additionally, it has been pointed out that imbalance in the frontal plane is a major consequence of stroke [55].

Regarding the discriminative power the CoM measures, they have been able to distinguish healthy from stroke individuals, which is not surprising. Additionally, some of them were even able to distinguish fallers from non-fallers: the Vertical CoM velocity during the turn phase of the TUG test [32], DTW in the go phase of the TUG test [33], and the AP-MoS in both paretic and non-paretic limbs [43]. Thus, the TUG test seems to be a relevant locomotor task to analyze risk of falling, especially in the turning phase [32].

The only study which compared right and left hemiparetic subjects did not report differences in the CoM measures [33]. The authors suggested that this unexpected result stems from characteristics of their sample, which was mildly impaired and uneven, regarding the number of left and right hemiparesis. This issue deserves further attention in future studies, once differences between gait of individuals with right and left hemiparesis are well documented in the literature [56–58] and should be captured by the CoM stability measures.

When contrasting the findings reported by Van Meulen et al. [43] with those presented by Bonnyaud et al. [30], it is possible to observe that AP-MoS was able to discriminate fallers from non-fallers. Interestingly, AP-MOS did not correlate with BBS measures. This might be related to limitations of the BBS that exhibits high ceiling effects and is not recommended for screening [59]. Therefore, one may consider the usefulness of the AP-MoS as an indicator of risk of falls or as an outcome measure able to detect if a therapy is able to reduce the risk of falling.

The clinical interventions presented in the reviewed studies employed different therapeutic approaches (ankle-foot orthosis, [22,27], visual feedback of CoM [29], modified constraint-induced movement [30] and rhythmic cuing [37]). They aimed at reducing the energy expenditure during gait, which is an important clinical concern, as the high energy cost of hemiparetic gait [60,61] compromises the patient functionality and community participation. In all studies [22,27,29,30,37], the outcome measures derived from the CoM demonstrated that the therapeutic goal was achieved because the distance between peaks of vertical CoM during the paretic and the non-paretic steps was reduced. Thus, the analysis of the CoM measures confirmed that the therapies were able to treat the cause of the high energy expenditure, i.e., the substantial muscle work of the non-paretic lower limb, which excessively lifts the CoM against gravity [62].

These intervention studies, despite of the reduced number, indicate a potential usefulness of the CoM parameters as outcome measures in clinical interventions. The CoM measures may not only provide information about effectiveness of the therapies, but also about the changes in the motor strategies applied by the patients. Therapists could take advantage of this information to customize their approach according to the needs of each patient.

In summary, the points drawn from analyzing the studies with clinical focus were: (i) the ML MoS and frontal plane angular momentum, as they correlate with clinical measures, could be employed to characterize the dynamical stability; (ii) the AP-MoS may be able to detect the risk of falls with more sensitivity than the BBS and (iii) symmetry of the CoM displacement in the vertical direction might be used as outcome measure to get information on the energy expenditure. In a broader sense, it was concluded that the CoM measures are promising to provide clinically relevant information about gait stability of stroke patients.

These point are relevant because the scientific community has been searching for summary measures [63] able to quantify gait performance and clinical evolution from data collected in gait laboratories. Examples of such measures are the Gait Deviation Index (GDI) [64], the Gait Profile Scores (GPS) [65], GDI-Kinect [66]. In spite of their applicability, sound statistical construction and repeatability for stroke patients [67,68], the equations of these measures are somewhat artificial. The CoM parameters, on the other hand, “summarize” kinematic and kinetic data with the help of mathematical equations derived from physical laws. Thus, the information provided by the CoM measures may be useful as it indicates not only if changes have occurred, but also how these changes are related to the gait dynamics and stability.

Even though the CoM parameters are potentially useful for clinical application, the importance of clinical measures must be recognized (e.g., BBS), as they are widely accepted, with well-known and adequate psychometric properties for the stroke patients [59,69,70]. On the other hand, the psychometric properties of the CoM measures, such as their repeatability [71] and the minimal clinically important change [72] remain to be determined.

4.1.2. Conclusions of studies with research focus

The studies with research focus (Table 5) were devoted primarily to investigate the strategies for postural control in the presence of stroke. To do this, comparisons between groups (healthy vs. stroke, stroke with different functionality levels) or between limbs were performed. They also submitted the stroke individuals to external stimuli or perturbations.

Contrary to expected, ML MoS was not able to differentiate healthy individuals from stroke patients [15,18]. This may be related to the facty that stroke patients try to maintain ML stability by increasing the step width [18], similarly to older adults with high risk of falls [73]. On the other hand, the MoS was smaller among stroke patients during unperturbed gait [15] and during a gait adaptability task [18], revealing poor stability in the sagital plane. Among stroke patients, those with higher risk of falls [28] and the slower walkers [45], were less able to maintain the CoM near the midline due to impairments in the control of paretic limb during foot placement. Therefore, increasing the step width does not improve body stability during gait after a stroke. This strategy differs from that observed in healthy adults [17], possibly due to the impaired control of the paretic limb.

Additional knowledge was provided by studies that compared the behavior of the CoM parameters between the paretic and the non-paretic limbs [25,34,35,39]. The difficulty in moving the CoM during the stance phase of the affected limb [25,34,39] resulted in a larger distance between the CoM and CoP in the AP direction [39], in larger lateral and shorter vertical and forward amplitudes [34] than during the non-affected stance. When a gait slip was induced on the affected limb, the poor reactive control was evidenced by a lower CoM stability [35].

When challenged to walk faster, the individuals with stroke had difficulty in increasing their stride frequency [38] due to their limited capacity to increase the backward and ML MoS. On the other hand, after a previous slip experience, stroke survivors adopted reactive strategies of CoM control [36] and were able to reduce the CoM sway when required to maintain the head oriented towards a stationary target [44]. Therefore, future studies employing CoM measures, could help to clarify the issues related to the development of new motor strategies when strokes patients are exposed to perturbation.

An interesting finding of Kao et al. [15] is that, average ML MoS were similar between healthy and stroke subjects, but its variability was larger in the first group. Although this have appeared in a single work of this review, variability of CoM measures should be investigated in future studies. This point of view stems from the fact that spatio-temporal gait parameters of stroke patients exhibit larger variability than those of healthy controls and the amount of variability correlates with severity of stroke [54]. Thus, poorer motor control may be related to larger variability. This is in accordance with other finding of Kao et al. [15], that the variability of ML MoS was significantly larger at the affected leg compared to the unaffected one.

The relationship between variability and motor control was also explored in the study of Papi et al., 2015 [24], who employed the UCM approach [74] to analyze the relationship between the variability of the CoM trajectory and the joint angles. This pilot study revealed that, although the variability of the CoM and angle trajectories were larger in the presence of stroke, the stroke patients were able to coordinate the many joints similarly to healthy individuals, exhibiting similar stability index [24]. The mechanisms behind this finding could be addressed in future studies considering not only the CoM as a priority variable [12,13,75,76], but also the xCoM or the MoS.

In summary, the studies with the research focus allowed to clarify that (i) widening the step does not improve stability in stroke patients; (ii) the main reason for instability might be the impaired control of the non-paretic limb during both the stance phase (when it is responsible for foot placement) and in support phase (when it is moving the CoM in relation to the CoP). Moreover, to gain a better understanding of motor control after stroke, future studies should address the variability of CoM measures.

5. Conclusion

In this review, (a) the methodological aspects involved in the calculation of the CoM and subsequent determination of stability parameters; (b) the purpose of analysis of CoM in studies of dynamic stability after stroke; and (c) the main conclusions about gait stability drawn from studies analyzing the CoM during gait were examined.

Regarding the methodological aspects of the CoM calculation, it was concluded that the segmental method with few body segments (7–9) may allow a compromise between accuracy and feasibility. The importance of choosing carefully the filtering techniques according to the motor task and of using an appropriate number of repetitions due to intra-subject variability was highlighted.

Two prevalent profiles of the included papers, regarding their purpose were identified: clinical and research-oriented studies.

The main points drawn from the studies with clinical focus are that the ML MoS and frontal plane angular momentum may be good parameters to indicate dynamical stability; the AP-MoS may be able to detect the risk of falls and the symmetry of CoM displacement in the vertical direction and is a possible outcome measure related to energy expenditure. The CoM measures are considered as potentially useful for clinical application, but future studies are necessary to determine their psychometric properties.

The use of CoM measures in the research-oriented studies allowed to clarify that the wider steps do not improve stability and the impaired control of the non-paretic limb may constitute the main source of instability. Moreover, future studies should address the variability of the CoM measures, in a similar way as other classical gait parameters, as it may provide important information on the stability and postural control.

In this review, it was possible to: to make recommendations on the methods for calculating the CoM and its measures; to identify the potential usefulness of the CoM parameters for clinical and research applications and to indicate issues that could be addressed in future studies related to gait stability measures.

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